An Explorative Study into the Mechanics of Projectile Impacts to the Head

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Abstract There has been little research focused on the mechanics of high-velocity, low-mass projectile impacts to the head. The little work that has been conducted has focused solely on linear acceleration, despite the evidence linking rotational acceleration to the development of brain injury. The aim of this study was to explore the presence of rotational acceleration in projectile impacts and investigate the influence of impact location. A pressurised air cannon was used to project a BOLA™ ball at 22 and 28 $m.s^{-1}$ towards a BSEN 960:2006 headform positioned to elicit impacts at frontal and lateral locations. High-speed video and accelerometer measurements were used to investigate differences in contact duration, ball deformation and average linear and rotational acceleration during loading.

Contact duration was found to be independent of impact location or speed. Greater ball deformation was observed in frontal impacts, despite no differences in time to maximum deformation. Average linear acceleration was observed to be greater during the loading phase in the frontal impacts then in the lateral impacts, potentially due to differences in surface geometry, resulting in differences in ball deformation. Average rotational acceleration was greater in lateral impacts potentially due to differences in the moments of inertia of the headform. Rotational acceleration was found to be higher than previously published injury thresholds and therefore a potentially important factor in projectile impacts, warranting further research.

Keywords Impact characteristics, Impact location, Projectile impacts, Rotational acceleration.

I. INTRODUCTION

The response of a head to an impact has been characterized through the assessment of the observed acceleration experienced by the head during the impact duration. There has been substantial research linking the acceleration of the head with the loading and deformation experienced by the brain and therefore the development of injury [1-3]. In regard to the aetiology of concussion, linear acceleration of the head has been shown to correlate highly with intracranial pressure gradients, leading to coup and contre-coup injury [3,4]. On the other hand rotational acceleration of the head has been shown to correlate with the development of shear strains within the brain [5-7] – particularly at core regions of the brain [8]. It has been stated that the brain is particularly susceptible to damage from shear strains [5], and due to this, it has been suggested that these are the root cause of concussion [8-9]. However, other research has stated that concussion is likely a result of a combination of linear and rotational acceleration of the head [4, 10, 11].

When considering head impacts in sports, there has been substantial research into the mechanics of high-mass, low-velocity collisions like those seen in American Football or Rugby [12,13]. As a result, the observed accelerations of the head during these types of collisions are relatively well understood. Conversely there has been little work into the mechanics of low-mass, high-velocity head impacts that may occur in sports such as Cricket, Baseball and Hockey, among others. Head injuries including lacerations and concussions occur in these sports despite the widespread use of helmets [14-16] The limited research into the mechanics of these types of collisions has mainly focused on the performance of personal protective equipment (PPE), drawing conclusions based solely on the linear acceleration observed during an impact [17] and has often utilized potentially inappropriate drop tests [5]. As a result the specific response of the head to these types of collisions remains unclear and in need of further investigation [18]. There are many factors that influence the response of a head during an impact and in order to investigate this in a controlled, laboratory setting, headform surrogates that are instrumented with accelerometers are commonly used. In this study, as in various British Standards, the BS EN 960:2006 headform [19] is used. Research has shown that the results of drop tests are influenced by the type of headform that is used [20] and therefore the properties of the headform that is used should be understood in order to complete a full and detailed mechanical analysis.

The concept of measuring just linear acceleration stems from an early study that showed a correlation between linear and rotational acceleration [21]. Due to this, many researchers and standards agencies [22,23] have used linear acceleration as the sole indicator of the severity of an impact or as a measure of the impact attenuation performance of PPE. Although this correlation may hold true in some impact situations, in others this may not be the case – particularly in impacts of varied location and vector [21]. Due to limited research and a focus on linear acceleration, the importance of rotational acceleration during a projectile impact to the head, and in particular one in Cricket, remains unknown. As a result of this, PPE used in Cricket has been (and continues to be) developed with little regard for rotational acceleration, which, as previously mentioned, may play a pivotal role in the development of brain injury, and in particular concussion.

For a given impact force, the rotational acceleration observed during an impact with a head is dependent on both the distance from the center of gravity (CoG) of the headform to the impact site and the moment of inertia of the head about the axis of rotation. The distance between the CoG and impact site is clearly influenced by the impact location and although the moment of inertia of the head cannot be influenced by aspects of the impact characteristics, the impact location does determine the primary axis about which the head rotates. Impact location then, is also an important factor that influences the observed response of the head during a projectile impact, and as previously mentioned, also influences the relationship between linear and rotational acceleration.

This study is intended to be an initial step that will be proceeded by further research looking into Cricket specific impacts. As such, the aims of this study are to; 1) provide an initial explorative study into the presence of rotational acceleration in projectile impacts, and 2) identify the influence of impact location on linear and rotational acceleration. These aims will be investigated by impacting the headform through the frontal and lateral planes and taking measurements using high-speed video and accelerometers. Theoretically, linear acceleration should not be influenced by impact location since $F = ma$, and the mass of the headform remains constant. Impact location should however influence rotational acceleration, producing greater acceleration about the axis with the lowest moment of inertia, since $T = I\alpha$ (where T = torque, I = moment of inertia and α = rotational acceleration.

II. METHODS

Experimental Testing

A bespoke experimental setup allowed an instrumented BS EN 960:2006 headform [19], size 575 (mass 4.7 kg) to be suspended using bungee cords. This suspension arrangement was selected in an attempt to create a freely suspended set-up. Although this was not truly free, the stiffness of this arrangement was 1.8 N/mm, similar to the passive stiffness of the human neck [24] and therefore representative of a worst case scenario collision, where an impact would be unexpected and therefore have no recruited musculature to stiffen the neck and restrict head acceleration. The experimental arrangement allowed the orientation of the headform to be adjusted, so that the impact location could be varied.

The same type of BOLA ™ ball (solid polyurethane ball with a mass of 150g, and a diameter of 71 mm) was used throughout all of the tests. Although this ball is more compliant than a Cricket, Hockey or Baseball, it was chosen as it has a similar mass and diameter, and due to the increased sphericity, is more accurate when projected [25]. A pressurised air cannon was used to project the ball towards the headform at two impact speeds: 22 and 28 $m.s^{-1}$ ±2 (approx. 50 and 60 mph). Although these speeds are lower than would be seen in the professional forms of Cricket, Hockey and Baseball, they are representative of ball speeds in the recreational game, and are indeed used in the British Standard for head protectors for cricketers [22]. The headform orientation was adjusted (Fig. 1) so that impact would occur on the reference plane of the headform (136 mm from the base, which corresponds to around halfway up the human forehead), on the centre line of the frontal and lateral planes. These positions were primarily chosen for practical reasons, in that these locations are visibly marked on the headform, although future work should look to impact more varied locations and vectors. The impacts were completed on a bare headform, with no helmet present in order to investigate a baseline response, from which further investigations can be based.

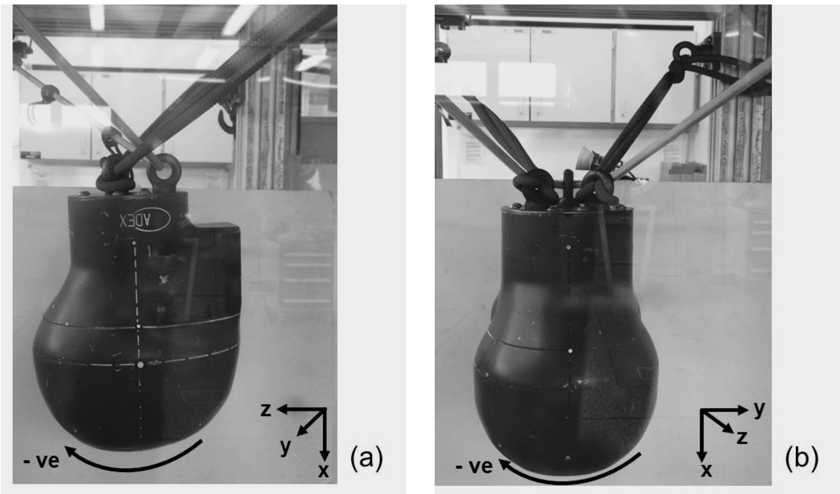

Fig. 1. (a) the headform orientation allowing frontal impacts (b) the headform orientation allowing lateral impacts. The ball was projected from right to left, parallel to the Z and Y axes respectively.

The headform manufacturer (Cadex, Canada) report the centre of gravity to be on the X axis of the headform (as shown in Fig. 1), 12.7 mm below the reference plane. The moments of inertia of the headform about the X, Y and Z axes are reported to be 193.2, 321.6 and 271.5 $kg.cm^2$ respectively.

Two PCB 356B21 accelerometers were fitted inside the headform using a mount. Accelerometer 1 was mounted on the X axis, 124 mm from the base of the headform. Accelerometer 2 was mounted 45 mm directly below this (Fig. 2). Following signal conditioning, the output from each of the accelerometers were recorded using two LeCroy WaveJet 324 digital oscilloscopes with a sample frequency of 1 MHz. The accelerometer sensitivities were determined in advance using a Bruel and Kjaer calibration unit to be 1.142, 1.16 and 1.153 $mV/m.s^{-2}$ for the X, Y and Z directions of accelerometer 1 and 1.181, 1.128 and 1.124 $mV/m.s^{-2}$ for the X, Y and Z orientations of accelerometer 2.

Two Arri pocket Par 400 lights were used to illuminate the test area. A Photron FastCam SA1 colour high-speed video camera operating at 50 kHz (448 x 224 spatial resolution) was positioned lateral and perpendicular to the plane of ball movement, 630 mm from the headform. This allowed the recording of a portion of the headform, the full impact, and around 140mm of ball movement before and after impact. In order to ensure that only impacts that fell within the required speed were recorded, a pair of timing light gates (200 mm apart) were used to calculate the ball speed directly out of the cannon. The signal from the light gate closest to the headform was used to trigger the high-speed video and both accelerometers simultaneously. The equipment set-up can be seen in Fig. 3.

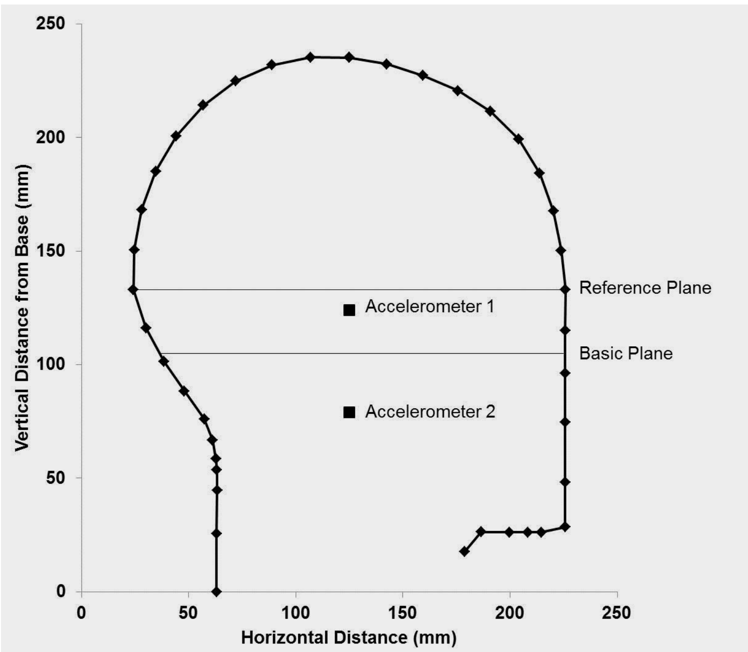

Fig. 2. Lateral cross section of the headform with Reference plane, Basic plane and accelerometer positions indicated.

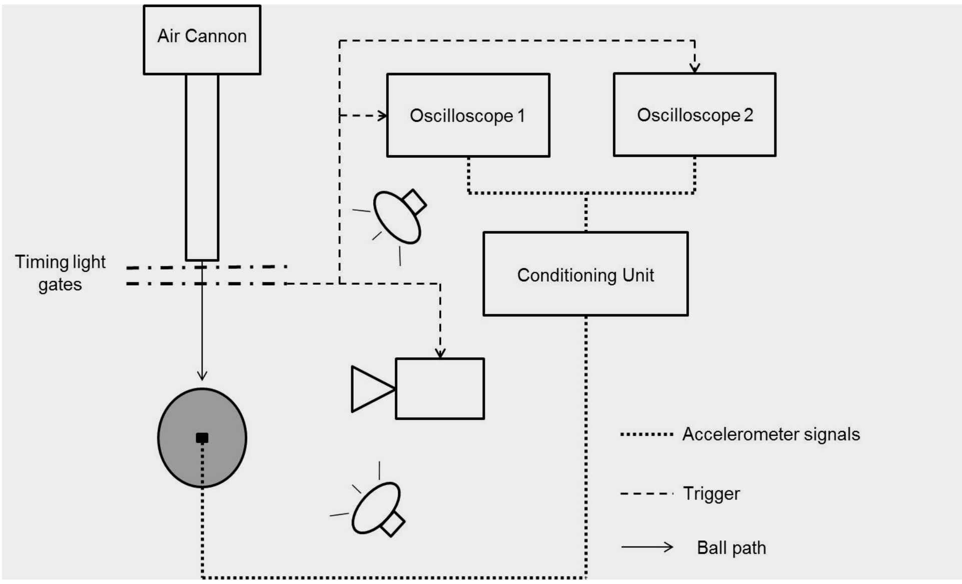

Fig. 3. Schematic of the equipment arrangement.

Data Processing

An image processing software application (ImagePro, MediaCybernetics Inc., MD) was used to process the high speed video data to identify the point of initial and final contact and to derive the magnitude and timing of the maximum ball deformation. Image calibration was completed using the initial, un-deformed ball diameter in each trial. This was measured to be 71 mm using a Vernier caliper and allowed the conversion from pixels to mm.

Accelerometer data were processed in Microsoft Excel. The outputs from both accelerometers were re-orientated so as to correspond with the global co-ordinate system shown in Fig. 1. Values were converted from V to SI units by applying the previously calculated sensitivities. As the high-speed video and accelerometer data were synchronized, the time stamps of the initial and final contact, as well as the instant of maximum ball deformation from the high-speed video were used to interrogate the accelerometer signals during the contact period. The instant of maximum ball deformation was used to divide the contact period onto the loading and unloading phases. Accelerations in the X, Y and Z directions from both accelerometers were time domain integrated to find velocity. Resultant linear accelerations and velocities were then calculated. Rotational acceleration about the primary axis of rotation were calculated by using the appropriate accelerometer outputs from accelerometers 1 and 2. For example, in the frontal impacts, the headform would primarily rotate about the Y axis and therefore the accelerations in the Z direction were used. Rotational acceleration was then time domain integrated to find the rotational velocity about the primary axis of rotation. For both linear and angular acceleration, the average acceleration over the loading phase was calculated (from initial contact to maximum ball deformation).

III. RESULTS

Contact Duration

Contact duration was determined through high-speed video recordings. Fig 4 shows the contact durations for the impacts that occurred in the frontal and lateral directions. It can be seen that the contact duration remains consistent regardless of impact location or speed, as all the data points are clustered closely together. This is confirmed in Table 1 which shows the average contact durations with standard deviations (SD). This shows consistent contact durations, with a variation of just 0.02 ms (1 frame).

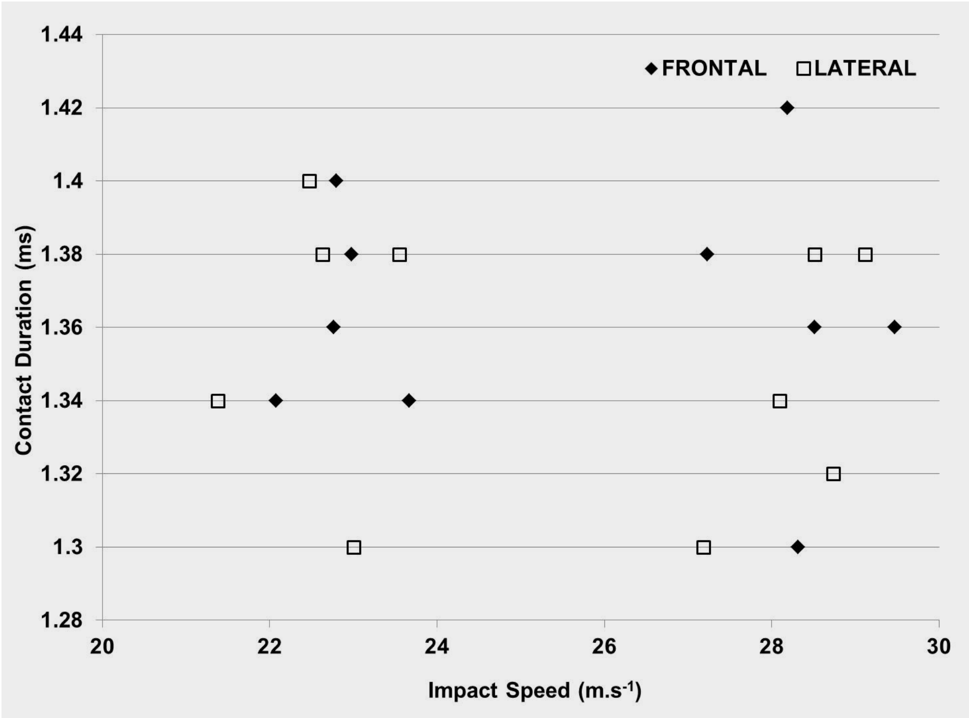

Fig. 4. Contact durations of the impacts occurring at the frontal and lateral locations.

TABLE I

	Impact Speed ($m.s^{-1}$)		Contact Duration (ms)	
	Mean	**SD**	**Mean**	**SD**
Frontal	22.86	0.51	1.36	0.02
	28.35	0.72	1.36	0.04
Lateral	22.61	0.72	1.36	0.04
	28.33	0.66	1.34	0.03

Deformation

The measured ball diameters at maximum deformation, as a percentage of the original ball diameter, for impacts at the frontal and lateral locations can be seen in Fig 5. It appears that, particularly for impacts occurring at 22 $m.s^{-1}$, that ball deformation is slightly greater at the frontal location than at the lateral location with the majority of the lateral data points sitting above the frontal equivalents. This trend also appears to occur with 28 $m.s^{-1}$ impacts, although to a lesser degree. Table II shows the average maximum ball deformations with SDs. This highlights the differences in maximum ball deformation at 22 m/s^{-1} with values of 85% (±0.8%) and 87.6% (±1.2%) for the frontal and lateral impact locations respectively. The average values also confirm that there is a slight difference at 28 $m.s^{-1}$ (82.7% (±0.7%) and 83.7% (±1.2%) for the frontal and lateral respectively), however, the SD values show some overlap.

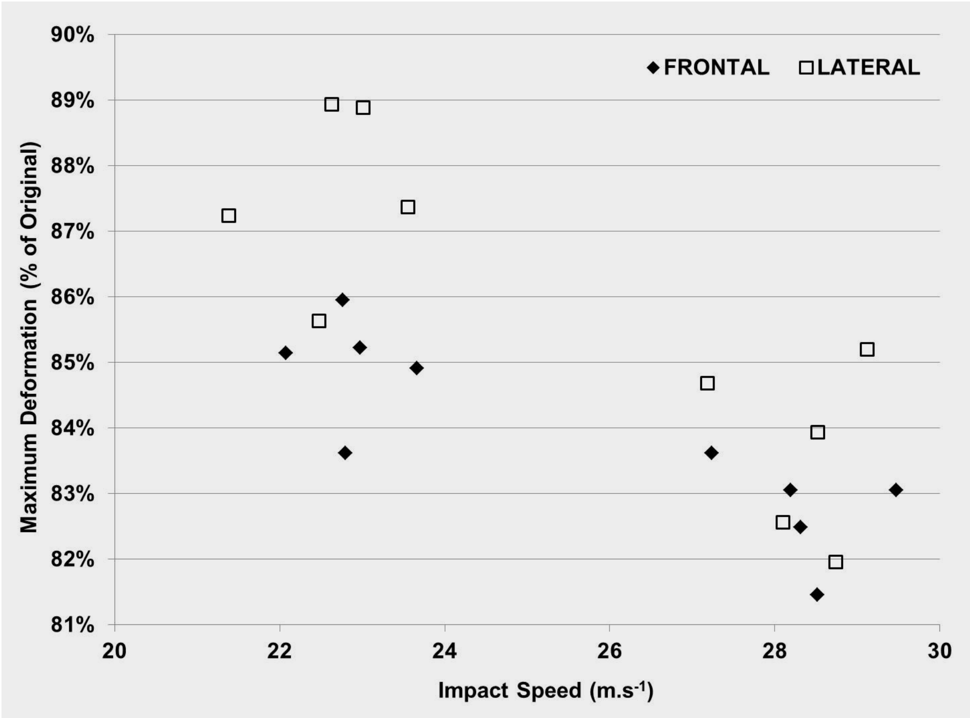

Fig. 5. Maximum ball deformation as a percentage of the original for the frontal and lateral impacts.

TABLE II

	Impact Speed ($m.s^{-1}$)		Maximum Deformation (% of original)	
	Mean	**SD**	**Mean**	**SD**
Frontal	22.86	0.51	85.0%	0.8%
	28.35	0.72	82.7%	0.7%
Lateral	22.61	0.72	87.6%	1.2%
	28.33	0.66	83.7%	1.2%

The timing of the maximum deformation was also determined through high-speed video recordings. It can be seen from Fig 6 that the duration between initial contact and maximum deformation is consistent between impact locations and impact speeds. This is confirmed in Table III which shows that the average values are similar, and when the SDs are considered there is substantial overlap between locations and speeds.

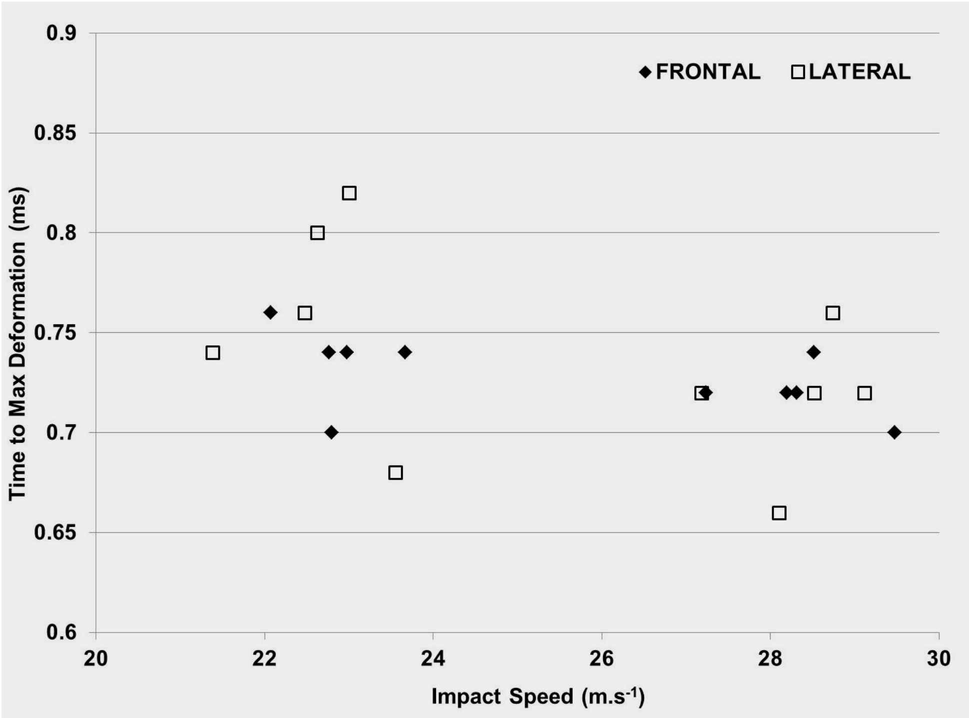

Fig. 6. Time from initial contact to maximum ball deformation for impacts at the frontal and lateral locations.

TABLE III

	Impact Speed ($m.s^{-1}$)		Time of Maximum Deformation (ms)	
	Mean	**SD**	**Mean**	**SD**
Frontal	22.86	0.51	0.74	0.02
	28.35	0.72	0.72	0.01
Lateral	22.61	0.72	0.76	0.05
	28.33	0.66	0.72	0.03

Linear Acceleration

The average resultant linear acceleration during the loading phase can be seen in Fig. 7. As expected, the average acceleration values increase with impact speed in both the frontal and lateral impact locations. It does however appear that there are differences in the average acceleration during loading between impact locations, with greater acceleration evident in frontal impacts than in lateral impacts. The average values for impacts at the frontal location where found to be 1898 (±47) $m.s^{-2}$ and 2117 (±142) $m.s^{-2}$ at nominal speeds of 22 and 28 $m.s^{-1}$ respectively. At the lateral impact location, lower values of 1412 (±63) $m.s^{-2}$ and 1747 (±82) $m.s^{-2}$ were observed at nominal impact speeds of 22 and 28 $m.s^{-1}$ respectively.

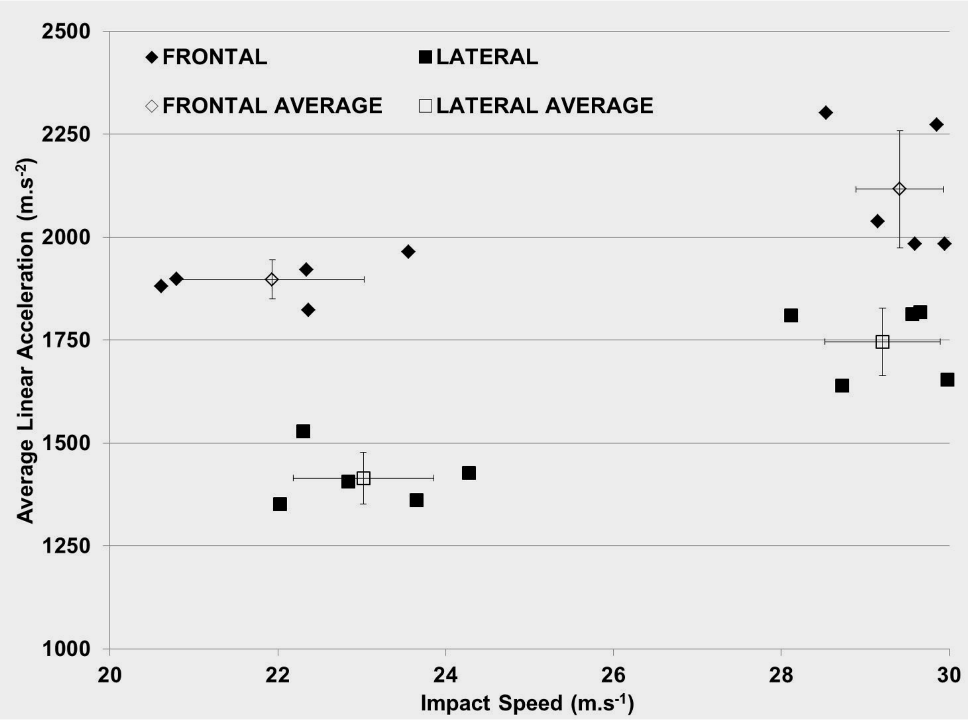

Fig. 7. Average linear acceleration during the loading phase for impacts at the frontal and lateral impact locations.

These differences may be explained by investigating the average velocity traces of each trial as shown in Fig. 8. Here it can be seen that whilst, at the end of contact, both impact locations appear to be moving at a similar velocity, the shape of the curves varies between impact locations. The frontal impacts appear to show a steeper initial increase to maximum velocity before plateauing, whereas the lateral impacts show a steadier increase up to the final velocity.

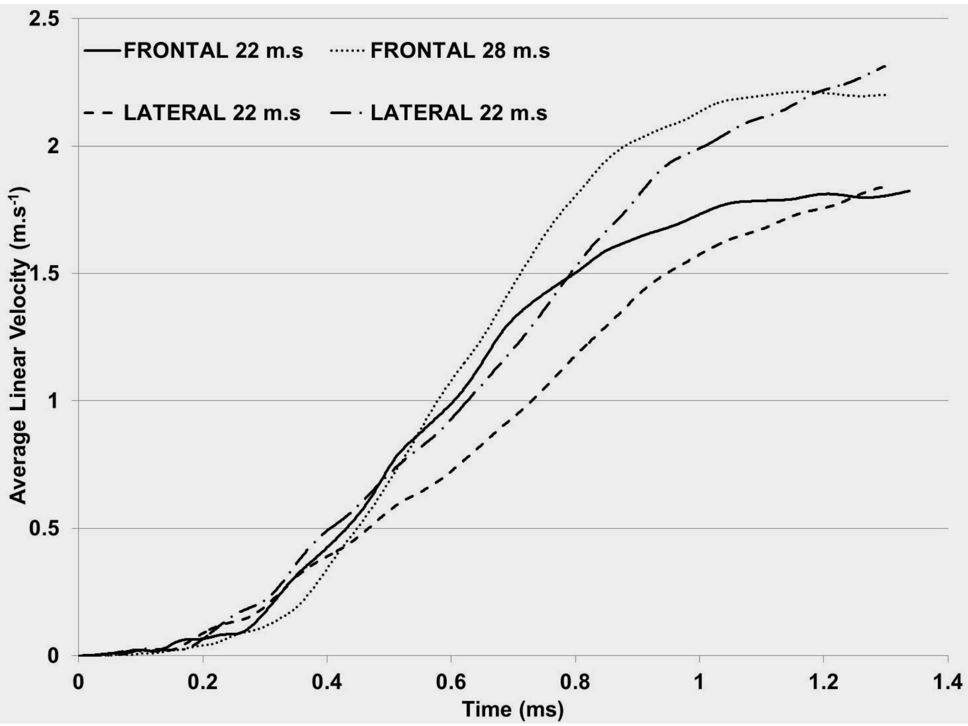

Fig. 8. Average linear velocity traces during the contact duration.

Rotational Acceleration

The average rotational acceleration observed during the loading phase can be seen in Fig. 9. It appears that for frontal impacts, impact speed has little effect on the observed rotational accelerations with average values of -6428 (± 558) rad.s^{-2} and -5771 (± 2282) rad.s^{-2} for the nominal impact speeds of 22 and 28 m.s^{-1} respectively.

Lateral impacts appear to be somewhat influenced by the impact speed and show greater acceleration during the loading phase with average values of -9196 (± 228) rad.s^{-2} and -12495 (± 547) rad.s^{-2}for the nominal impact speeds of 22 and 28 m.s^{-1} respectively.

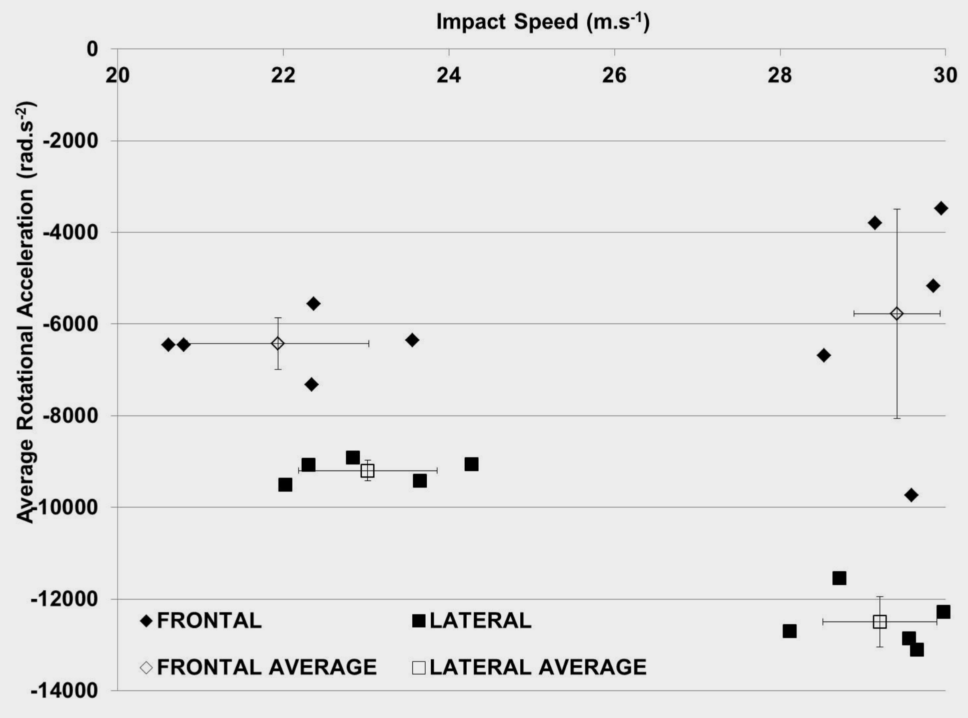

Fig 9. Average rotational acceleration during the loading phase for impacts at the frontal and lateral impact locations.

The average rotational velocity traces shown in Fig. 10 show that the lateral impacts produce a relatively stable change in velocity throughout the impact duration. The velocity observed during frontal impacts appears to be more variable, with substantial fluctuations. These traces suggest that increasing impact speed does indeed have an effect on the response of the headform during frontal impacts, however these differences may be disguised by the observed fluctuations.

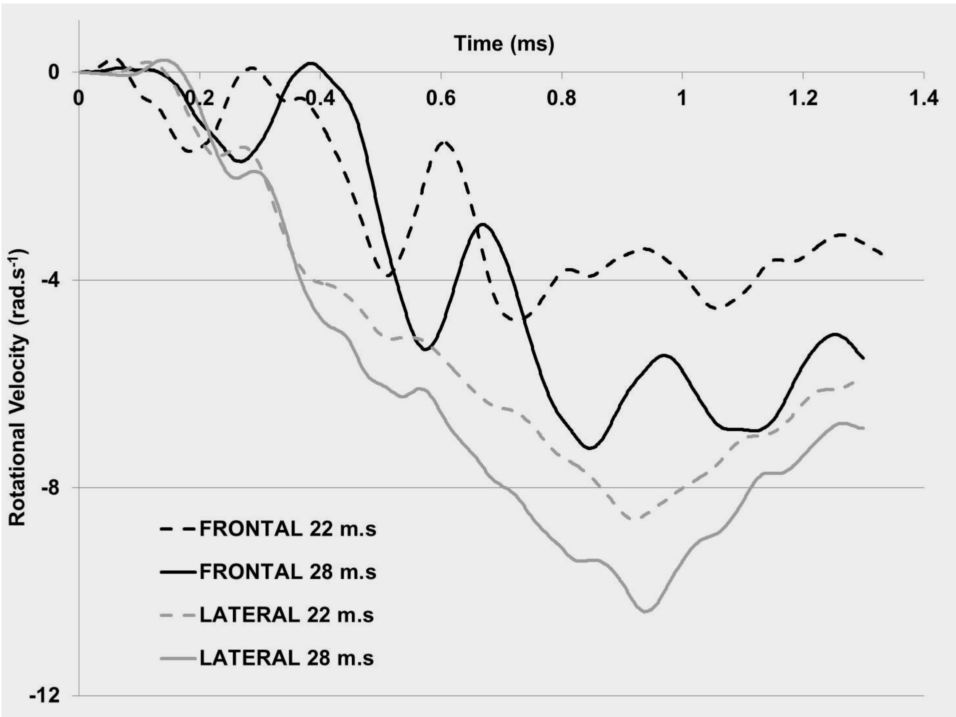

Fig. 10. Average rotational velocity traces during the contact duration.

IV. DISCUSSION

Since this study is intended to precede further investigations into cricket specific impacts, the BS EN 960:2006 headform used in this study based on its current use in the British Standard for head protectors for cricketers [21]. The mass and moments of inertia of this headform correspond reasonably well with values previously reported for the human head [24]. The stiffness of the suspension technique used here was found to be 1.8 N/mm, similar to the passive stiffness of the human neck [26] and therefore simulates a 'worst-case scenario' response. The adult BOLA ball was chosen due to its current us in the aforementioned cricket standard [22]. This type of ball is more compliant than a regular cricket ball, and therefore further research should look into the differences in the dynamic response of the headform between these ball types. It can be assumed however, that if angular acceleration is an important parameter in these impacts then this will certainly be the case when utilizing a cricket ball since increased stiffness has been shown to increase the response of a headform [27].

The contact durations of the impacts investigated in this study were found to be consistent regardless of impact location and impact speed. In a collision between two perfectly rigid bodies, the contact duration is dependent on the dimensions of the smaller body and the wave speed of the material [28], so these findings would be expected. In this practical case, some deformation is present in both the ball and, to a lesser extent, the headform. Therefore in this case the contact duration is dependent on a more complex set of parameters, it can be seen that contact duration remains constant at around 1.36 ms. This finding is in line with that of Daish [29] and Goldsmith [28] who both report that contact duration is greatest at low impact speeds, but as impact speed increases contact duration decreases until, at some sufficiently high impact speed, it becomes practically constant.

Measurement of ball deformation using high-speed video showed that slightly greater ball deformation was present in frontal impacts than in lateral impacts. This was potentially due to the differing local surface geometry of the headform at the point of impact as differences in the curvature of the headform at these sites were observed. It was also observed that there were little to no differences in time to maximum ball deformation between impact locations or speeds. The magnitude and timing of ball deformation should be considered as these influence the rate and magnitude dependencies of the ball stiffness. As greater ball deformation occurred during a similar time in the frontal impacts, it may be possible that the effective stiffness of the ball was greater during these impacts due to the aforementioned rate and magnitude dependencies. Although the determination of dynamic ball properties in realistic circumstances is challenging, further research into the dynamic visco-elastic properties of the impacting balls should be conducted in order to provide a more in-depth analysis of impacts. The effect of surface geometry on the ball deformation and the dynamic response of the headform should also be a focus of future research.

The analysis of linear acceleration focused primarily on the loading phase as it was reasoned that should any differences between impact locations be present, they would be most pronounced during this phase. The loading phase was defined as the period from initial contact to maximum ball deformation. As the linear acceleration of the headform is governed by Newton's second law, $F = ma$, it would be expected that, as the mass of the headform remains constant and the exerted force is dependent on the ball impact speed, the linear acceleration observed in the frontal and lateral impacts would be the same for a given impact speed. This was not the case in this study as the average acceleration during the loading phase was greater in the frontal impacts than in the lateral impacts at impact speeds of 22 and 28 $m.s^{-1}$. This however may be due to the previously mentioned differences in surface geometry leading to differences in ball deformation, and therefore slightly different ball stiffnesses due to the strain and magnitude dependencies, which has been shown to influence the dynamic response of a headform during an impact, with a stiffer ball producing a greater response [26]. The average velocity traces during the impacts also show interesting differences between the frontal and lateral impacts. Whilst the frontal impacts show a steeper increase in velocity before plateauing, the lateral impacts appear to show a steadier increase to a final velocity which is slightly greater than the frontal impacts. These differences are again potentially due to the differences in headform geometry at the point of impact. A simplified experimental protocol using a flat plate and increasing the level of contouring around the point of impact would provide a foundation of knowledge on which to build an increased understanding around this area.

The most important element of this study was concerned with rotational acceleration. Utilising multiple accelerometers allowed for the calculation of rotational acceleration about the principal axes of rotation. In the case of the frontal impacts, rotation would be expected about the Y axis. The principal axis of rotation for lateral impacts would be about the Z axis. Surprisingly, the average rotational acceleration during the loading phase of the frontal impacts appears to remain constant at both impact speeds of 22 and 28 $m.s^{-1}$. However, this is due to the fluctuating velocity traces presented in Fig. 10. These fluctuations are difficult to explain at the moment and require additional investigation. The velocity traces seen in the lateral impacts are much more consistent, with a relatively steady increase to maximum velocity, before showing some decrease. This steady increase led to average rotational acceleration values that were very consistent, and greater than those observed in frontal impacts. The differences in average acceleration during loading observed in lateral impacts compared to frontal impacts is due to the moments of inertia about the principal axes of rotation. For the frontal impacts, the headform would rotate principally about the Y axis, which has a reported moment of inertia of 321.6 $kg.cm^2$. As this is greater than the moment of inertia about the Z axis (271.5 $kg.cm^2$), which is the principal axis about which rotation would occur in lateral impacts, the lower average acceleration values are not surprising since $T = I\alpha$. In both impact locations, the levels of angular acceleration observed exceed previously published injury thresholds for concussion and are closer to the values associated with diffuse axonal injury [30]. Clearly rotational acceleration is an important factor that should be considered when investigating projectile impacts, particularly when considering the substantial previous research that has identified rotational acceleration as an injury mechanism in concussion and other more severe brain injuries [1-3,21]. This study provides an initial step in the mechanical analysis of projectile impacts, however as brain injuries, and concussions in particular, have such a complex aetiology more substantial research in collaboration with medical professionals is required in order to determine the exact result of angular acceleration. The influence of impact location is important in the determination of the observed rotational acceleration as it not only determines the principal axes of rotation, but also determines the distance between the centre of gravity of the headform and the impact location thereby influencing the torque generated. Additionally, impact vector should also be considered as in real-life impacts observed in sporting events are rarely direct and are often glancing blows. This research when combined with more in-depth mechanical analyses of projectile (and specifically cricket) impacts can have varied potential uses, including clinical diagnoses and treatments in addition to informing protective equipment design. Specifically in regard to the latter potential use, Cricket helmet manufacturers currently design products to pass the current standard [22] which assesses impact attenuation through linear acceleration alone. This research suggests that consideration should also be given to rotational acceleration, and perhaps further revision of the current standard should be considered in order to incorporate this.

V. CONCLUSIONS

Overall this study has utilized a realistic test method to provide an initial step in the determination of both linear and rotational acceleration during projectile impacts in sport. The dynamic response of the headform to projectile impacts was shown to vary with impact location. The differences in linear acceleration between impact locations observed in this study may be due to differences in surface geometry, resulting in different dynamic ball properties which requires further research. Differences in rotational acceleration were probably due to differences in the moment of inertia of the headform. Future research should look to investigate linear and rotational acceleration with more varied impact locations and vectors as the dynamic response of a headform has been shown to be sensitive to subtle changes in these impact characteristics [21].

VI. REFERENCES

[1] Meaney DF, Smith DH. Biomechanics of Concussion. Clinical Sport Medicine. 2011;30(1):19–31.

[2] Post A, Hoshizaki TB. Mechanisms of brain impact injuries and their prediction : A review. Trauma. 2012: 1–23.

[3] King A, Yang K, Zhang L. Is head injury caused by linear or angular acceleration. In: IRCOBI conference; 2003: 1–12.

[4] Gurdjian ES, Gurdjian ES. Re-evaluation of the biomechanics of blunt impact injury to the head. Gynecology and Obstetrics. 1975; 140(6): 845–50.

[5] Hoshizaki TB, Post A, Oeur RA, Brien SE. Current and future concepts in helmet and sports injury prevention. Neurosurgery. 2014; 75(4): 136–48.

[6] Post A, Blaine Hoshizaki T. Rotational Acceleration, Brain Tissue Strain, and the Relationship to

Concussion. Journal of Biomechanical Engineering. 2015 ;137(3)
[7] Zhang L, Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury. J Biomechanical Engineering. 2004; 126(2): 226–36.
[8] Zhang L, Yang KH, Dwarampudi R, Omori K, Li T, Chang K, et al. Recent advances in brain injury research: a new human head model development and validation. Stapp Car Crash J. 2001; 45(11): 369–94.
[9] Elkin BS, Morrison B. Region-specific tolerance criteria for the living brain. Stapp Car Crash J. 2007; 51: 127–38.
[10] Morrison B, Cater HL, Wang CC-B, Thomas FC, Hung CT, Ateshian G a, et al. A tissue level tolerance criterion for living brain developed with an in vitro model of traumatic mechanical loading. Stapp Car Crash Journal. 2003; 47: 93–105.
[11] Kleiven S. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. International Journal of Crashworthiness. 2006; 11(1): 65–79.
[12] Rowson B, Rowson S, Duma SM. Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets. Annals of Biomedical Engineering. 2015; 43(10): 14–8.
[13] Pellman E, Viano D, Tucker A, Casson I, Waeckerle J. Concussions in Professional Football: Reconstruction of Game Impacts and Injuries. Neurosurgery. 2003; 53(4): 799–814.
[14] Beyer J, Rowson S, Dma S. Concussions experienced by major league baseball catchers and umpires: field data and experimental baseball impacts. Annals of Biomedical Engineering. 2012; 40(1): 150-159.
[15] Murtaugh K. Injury Patterns among female field hockey players. Medicine and Science in Sports and Exercise. 2001; 33(2): 201-207.
[16] Ranson C, Peirce N, Young M. Batting head injury in professional cricket: a systematic video analysis of helmet safety characteristics. British journal of Sports Medicine. 2013; 1-5.
[17] McIntosh AS, McCrory P, Comerford J. The dynamics of concussive head impacts in rugby and Australian rules football. Medicine and Science in Sport Exercise. 2000; 32(12): 1980–4.
[18] McIntosh AS, Janda D. Evaluation of cricket helmet performance and comparison with baseball and ice hockey helmets. British Journal of Sports Medicine. 2003; 37(4): 325–30.
[19] Aare M, Kleiven S. Evaluation of head response to ballistic helmet impacts using the finite element method. International Journal of Impact Engineering. 2007; 34(3): 596–608.
[20] BS EN 960:2006 Headforms for use in the testing of protective helmets. British Standards Institution, London. 2006.
[21] Kendall M, Walsh ES, Hoshizaki TB. Comparison between Hybrid III and Hodgson – WSU headforms by linear and angular dynamic impact response. Proceedings of IMechE Part P Journal of Sports Engineering and Technology. 2012: 1–6.
[22] BS 7928:2013. Specification for head protectors for cricketers. 2013.
[23] Nocsae. Standard Performance Specification For Newly Manufactured Baseball / Softball Fielder 's Headgear Nocsae Doc (Nd) 029- 12m12. 2013
[24] Hodgson VR, Mason MW, Thomas LM. 1972. Head Model for Impact. In: Proceedings of the Stapp Car Crash Conference: 1-13.
[25] Halkon BJ, Velani N, Harland AR. The development of a test methodology for the determination of cricket batting helmet performance. IRCOBI Conference, Gothenburg, Sweden. 2013
[26] McGill SM, Jones K, Bennett G, Bishop PJ. Passive stiffness of the human neck in flexion, extension, and lateral bending. Clinical Biomechanics. 1994; 9(3): 193–8.
[27] Crisco JJ, Hendee SP, Greenwald RM. The influence of baseball modulus and mass on head and chest impacts: a theoreical study. Medicine and Science in Sports and Exercise. 1997; 29(1): 26–36.
[28] Goldsmith W. Impact. Richard Clay and Company; 1960.
[29] Daish CB. The Physics of Ball Games. The English Universities Press; 1972.
[30] Fernandes F, Sousa R. Hread injury predictors in sports trauma - A stae-of-the-art review. Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine. 2015; 229(8): 592-608.