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Evaluation of the protective capacity of baseball helmets for concussive impacts

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Abstract

The purpose of this research was to examine how four different types of baseball helmets perform for baseball impacts when performance was measured using variables associated with concussion. A helmeted Hybrid III headform was impacted by a baseball, and linear and rotational acceleration as well as maximum principal strain were measured for each impact condition. The method was successful in distinguishing differences in design characteristics between the baseball helmets. The results indicated that there is a high risk of concussive injury from being hit by a ball regardless of helmet worn.

Keywords: concussion; baseball; helmet

Introduction

In the U.S. approximately 19 million people engage in organized baseball every year, including over 5 million children under the age of 14 (Viano et al., 1993). At the more competitive level, there are 400,000 high school and 20,000 collegiate baseball athletes (Nicholls et al., 2004). Injuries in this sport occur as a result of a number of mechanisms, but the most prevalent and often severe result from baseball impacts. In particular, batters hit to the head by a pitch. Ball to player impacts account for 52-62% of all baseball related injuries, with the most severe resulting from ball hits to the player's head (Gessel et al., 2007). At the high school and collegiate level, concussive incidents have been reported to have occurred as 0.08 and 0.23 per 1000 athlete exposures during a game situation (at bats) (Gessel et al., 2007). At the professional level the rate of concussion remains unknown (Athiviraham et al., 2012).

Helmets are currently employed to reduce the incidence of head injury in baseball, and in large part have been successful in reducing the incidence of traumatic brain injury. While there is no literature reporting any reduction in concussion with the use of helmets in baseball, it is possible that the rate of concussion has not been affected, as has been found in other sports where head impact is common and helmets are used (Wennberg and Tator, 2003; Casson et al., 2010). Scientists investigating this phenomenon in other contact sports such as ice hockey and American football have found that concussion is largely influenced by rotational acceleration causing strain in the brain tissues (Forero Rueda et al., 2011; Post et al 2011; Post et al 2013b). The strain to the brain causes damage that is represented by the symptomology of concussion. The current helmet standards in all sports only measure linear acceleration, and thus do not account for this particular mechanism of concussion (Hoshizaki and Brien, 2004). This is not a novel concept, and researchers have developed methods of analysis of American football and ice

hockey helmet performance that focus on risks of concussion and metrics that are more closely associated with this type of injury (Post et al., 2011; Post et al., 2013b). An improved understanding of the mechanisms of injury within American football and ice hockey, and the situations that are likely to cause an increased risk of concussion has been studied (Zhang et al., 2004; Kleiven, 2007; Forero Rueda et al., 2011). As a result, improved standards are currently being developed, and rotational damping technologies have been developed in an effort to reduce the risk of injury. While this examination of mechanisms and helmet performance using unique methodologies linked to concussion has been developed in other sports (Aare and Halldin, 2003, Post et al., 2011; Post et al., 2013a), there has been no similar investigations in baseball. It is important to better understand the mechanism associated with concussive events in baseball to compare to other contact sports where concussion is also prevalent. In ice hockey the common mechanism of injury for concussion involves body to head (Hutchison et al., 2012; Rousseau, 2014) and in American football head to head impacts (Pellman et al., 2003) contact that create unique loading situations for brain tissue. Concussions in baseball involve high velocity (80 mph) low mass (baseball) impacts to the head (Athiviraham et al., 2012). The unique characteristics of this type of event demands a better description of the events resulting in concussion particular to baseball. The purpose of this research was to develop a methodology that could be used to evaluate the performance of helmets for impacts by baseballs a common cause of concussion in baseball.,

Methodology

Test apparatus

Since the primary cause of concussion for baseball involve impacts to the head from a thrown ball, this methodology focused on that particular mechanism. The impacting system used was a

pneumatic ball launcher, a Hybrid III 50th head and neckform outfitted with accelerometers to record three-dimensional motion, and a finite element model, specifically the University College Dublin Brian Trauma Model (UCDBTM) (Horgan and Gilchrist, 2003).

The impacting system was comprised of two main components: the pneumatic ball launcher, and the table that housed the Hybrid III 50th head and neckform to be impacted. The pneumatic ball launcher was attached to a steel frame to prevent the launcher from moving, and was attached to a compressed air cylinder (**Figure 1**). The amount of pressure in the cylinder was adjusted to achieve the appropriate impact velocity for a baseball. For this research the impact velocity was 35.7 m/s (80 mph), which was measured by laser timegate at the release point at the barrel of the pneumatic launcher. The table to which the Hybrid III 50th head and neckform was attached was constrained so that it could not move in translation, but had an adjustment system that allowed for positioning of the impact location of the headform in six degrees of freedom. Impact site of the headform was confirmed by high speed camera (High Speed Imaging PCI-512 Fastcam) sampling at 2,000 Hz that was recorded by Photron Motion Tools software.

Insert Figure 1 here

For this research a 50th percentile Hybrid III head (mass 4.54 +/- 0.01 kg) and neckform (mass 1.54 +/-0.01 kg) was used. The headform was instrumented with nine accelerometers set up in a 3-2-2-2 array for the measurement of the three-dimensional motion under impact (Padgaonkar et al., 1975). The accelerometers used were Endevco 7264C-2KTZ-2-300 (Irvine, CA, USA), which were sampled at 20 kHz and filtered using a CFC 1000 filter according to the SAE J211 head impact convention. The accelerometer signals were collected by a Diversified Technical Systems TDAS Pro Lab module (DTS, Seal Beach, CA, USA) before being processed

by TDAS software. The positive axes were aligned towards the front (x axis), right (y axis), and base (z axis) of the headform, rotations were defined with the left hand rule.

Helmets

Four models of baseball helmet were impacted according to the concussion testing protocol. All helmets were manufactured in 2014 and were fitted and secured to the Hybrid 50th headform according to manufacturer's instructions. The characteristics of the helmets can be found in **figure 2**.

Helmet	Circumference (cm)	Weight (g)	Shell type	Offset (mm)	Liner type	Image
A	73	720	ABS	8.6	Vinyl nitrile	
B	72.5	650	ABS and kevlar liner	8.0 (15.0 at front)	Vinyl nitrile	
C	70.5	548	ABS	9.5	Expanded polypropylene	
D	69	560	Composite	7.5	Vinyl nitrile	

Figure 2. Helmet specifications. (Offset does not include comfort foam).

Finite element model

To evaluate the performance of the helmets in terms of brain deformation a finite element model of the human head/brain was used. The model used was the UCDBTM and is one of few models used for this type of research in the world (Horgan and Gilchrist, 2003; Horgan and Gilchrist,

2004). The geometry of the model was developed from medical imaging (computed tomography and magnetic resonance imaging) of a head of a male cadaver. The UCDBTM was comprised of the following parts: scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum, and the brain stem (Horgan and Gilchrist, 2004). In total, the UCDBTM has approximately 26,000 hexahedral elements.

The material properties of the UCDBTM was based on anatomical testing on cadavers and tissue samples (Ruan, 1994; Zhou et al., 1995; Willinger et al., 1995; Zhang et al., 2001; Kleiven and von Holst, 2003) (**Tables 1 and 2**). The brain tissue was modelled using a linearly viscoelastic model combined with large deformation theory. The behaviour of the tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus (Horgan and Gilchrist, 2003). The compression of the brain tissue was considered elastic. The shear characteristics of the viscoelastic brain were expressed by:

$$(1) \quad G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t}$$

With G_∞ representing the long term shear modulus, G_0 the short term modulus and β is the decay factor. A Mooney-Rivlin hyperelastic material model was used for the brain to maintain these properties in conjunction with a viscoelastic material property in ABAQUS, giving the material a decay factor of $\beta = 145\ s^{-1}$ (Horgan and Gilchrist, 2003). The hyperelastic law was given by:

$$(2) \quad C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15}\ (\text{Pa})$$

where C_{10} is the mechanical energy absorbed by the material when the first strain invariant changes by a unit step input and C_{01} is the energy absorbed when the second strain invariant changes by a unit step (Mendis et al., 1995; Miller and Chinzei, 1997) and t is the time in seconds. To define the brain skull interaction, the UCDBTM uses a sliding boundary condition with no separation between the pia and the CSF. The modeling of the CSF was conducted using solid elements with the bulk modulus of water and a low shear modulus (Horgan and Gilchrist,

2003; Horgan and Gilchrist, 2004). The coefficient of friction between the sliding interface was 0.2 (Miller et al., 1998).

Table 1 - Material properties for UCDBTM

Material	Young's modulus (MPa)	Poisson's ratio	Density (kg/m^3)
Scalp	16.7	0.42	1000
Cortical Bone	15 000	0.22	2000
Trabecular Bone	1000	0.24	1300
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx and Tentorium	31.5	0.45	1130
Brain	Hyper Elastic	0.499981	1060
CSF	15 000	0.5	1000
Facial Bone	500	0.23	2100

Table 2 - Material properties of brain tissue used in the UCDBTM

	Shear modulus (kPa)		Decay constant (s^{-1})	Bulk modulus (GPa)
	G_0	G_∞		
Grey matter	10	2	80	2.19
White matter	12.5	2.5	80	2.19
Brain stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

The model was validated against the Nahum et al (1977) research into pressures resulting in the head from cadaveric impacts and Hardy et al.'s (2001) neutral density tracker research into cadaveric brain motion from impact. Further validations were carried out reconstructing real-life incidents that resulted in traumatic brain injury (Doorly and Gilchrist, 2006; Post et al., 2014a) and concussion (Zanetti et al., 2012; Rousseau, 2014) and were found to be in good agreement with magnitudes of stress and strain found in the literature.

Test procedure

Seven projectile impact sites were used for the development of the baseball helmet protocol (**Figures 3 and 4**). The impact sites were determined by conducting a review of game video of

Major League Baseball ball to head impacts that resulted in a concussion. From this analysis, four commonly impacted sites were decided upon, and a further three more were added to ensure evaluation of impacts similar to current standard methods. Each helmet was fitted to the Hybrid III headform and impacted at 35.7 m/s (80 mph) (Athiviraham et al., 2012). To eliminate the effect of repeated impacts to the same area of the helmet, a new helmet was used for each impact to the same site. Each helmet model was impacted three times per site (new helmet each time), for a total of 21 impacts per helmet model. The time between impacts was approximately 300 +/- 60 seconds. Impact site accuracy was determined through use of laser targeting system, and confirmed by high speed camera. The results of each impact were recorded in linear and rotational acceleration in the x, y, and z axis. These acceleration loading curves were then applied to the centre of gravity of the UCDBTM to determine the strains involved for each simulated impact. In this research maximum principal strain was determined as the most appropriate predictive variable due to its common usage for concussive threshold testing in anatomical and finite element research (Zhang et al., 2004; Kleiven, 2007; Morrison et al., 2006), thus providing comparative data points for the discussion of the appropriateness of this proposed methodology.

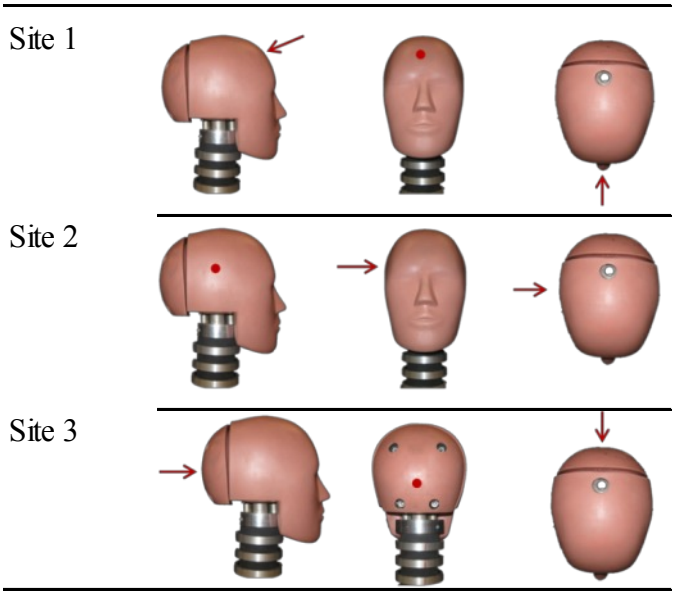


Figure 3. Impact sites on the unhelmeted Hybrid III headform similar to those from standard helmet methods.

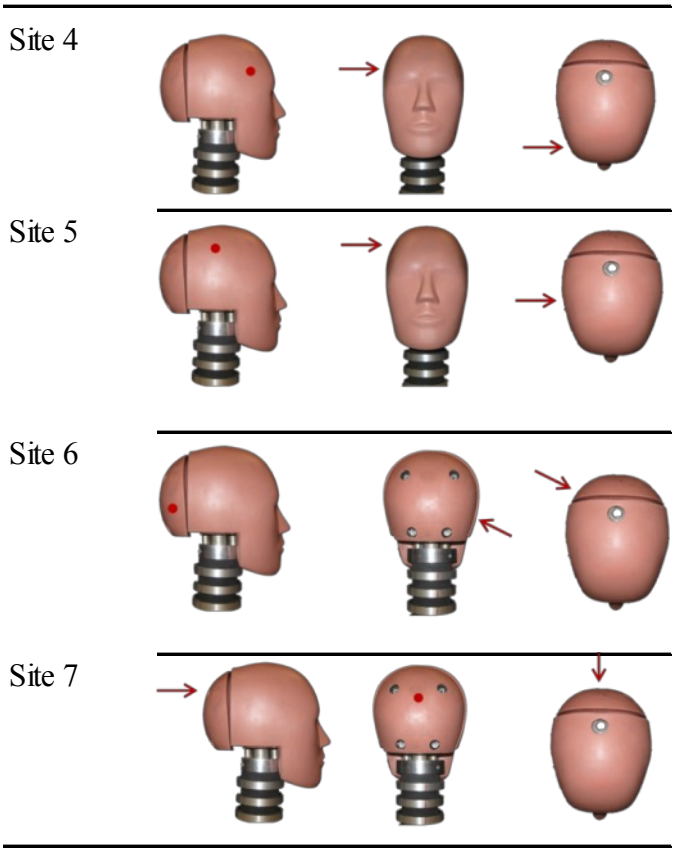


Figure 4. Impact sites on the unhelmeted headform as selected from video analysis of in game concussions.

Statistical analysis

Significant differences in peak resultant linear acceleration, peak resultant rotational acceleration, and maximum principal strain in the grey and white matter between impact sites were determined by using one-way analysis of variance. The same procedure was used to determine differences between helmet type. Tukey's post-hoc test was applied when significance was found to occur with the alpha values set to 0.05. Further analysis was conducted on the relationship between peak resultant linear acceleration, peak resultant rotational acceleration, and maximum principal strain by Pearson correlation to determine the influence of dynamic response on brain strain for this projectile impact methodology.

Results

The dynamic response and maximum principal strain results of the baseball helmet impacts are shown in **tables 3 and 4**. The Pearson correlation results are shown in **table 5**. The statistical analyses found significant main effects between both impact site, and helmet impacted.

Impact site

Significant main effects were found for the examination of differences between impact sites ($p<0.05$). When examining the effect of impact site there were many differences in magnitude of response found. Site 1 had similar magnitudes of linear acceleration (160.0 g) to the other sites, except for site 7 (121.0 g) ($p<0.01$). Site 2 was not significantly different from sites 1, 3, and 4, and larger in magnitude than sites 5, 6, and 7 ($p<0.05$). Site 3 was not significantly different from sites 1, 2, 3, and 7, but was larger than sites 5 and 6 ($p<0.05$). Impacts to site 4 produced magnitudes that were similar to sites 1, 2, and 3, and significantly larger than those at sites 5 through 7 ($p<0.05$). The site 5 impacts were similar to those impacts at sites 1, 6, and 7, and lower than those at sites 2 through 4 ($p<0.05$). Site 6 had impacts that produced results similar to

sites 5 and 7, and were lower in magnitude than the other impact sites ($p<0.05$). The impacts to site 7 produced magnitudes of linear acceleration that were similar to those at sites 1, 3, 5, and 6, and significantly lower than those at sites 2, and 4 ($p<0.05$). Overall, the lowest magnitude impacts were to site 6 (73.1 g) and the highest to site 2 (207.5 g).

When examining rotational acceleration, site 1 (10.6 krad/s^2) had similar results to sites 3 (14.3 krad/s^2), 5 (11.7 krad/s^2), 6 (7.0 krad/s^2) and 7 (7.3 krad/s^2), but lower than sites 2 (21.1 krad/s^2) and 4 (19.9 krad/s^2) ($p<0.05$). Impacts to site 2 produced magnitudes of rotational acceleration that were significantly larger than all other sites other than site 4 ($p<0.05$). The site 3 impacts produced magnitudes that were similar to sites 1 and 5, and lower than sites 6 and 7, but larger than sites 2 and 4 ($p<0.05$). Site 4 produced magnitudes of rotational acceleration that were larger than all other sites except site 2 ($p<0.05$). Impacts to site 5 produced results that were similar to sites 1, 3, 6, and 7, but lower than sites 2 and 4 ($p<0.05$). Sites 6 and 7 produced results of rotational acceleration that were similar to sites 1, 5 and 7, but lower than the remaining sites ($p<0.05$).

For maximum principal strain, the peak strains were found in the grey matter for this and all sites except site 6, where the peak was found in the white matter. The peak strains for site 1 (0.297) were similar to sites 2 (0.289), 4 (0.322), and 5 (0.260), and significantly larger than those found at sites 3 (0.223), 6 (0.195), and 7 (0.210) ($p<0.05$). Site 2 had results that were significantly larger than impacts to site 3, 6, and 7 ($p<0.05$), but similar to the remaining sites. The site 3 impacts were significantly lower in magnitude of MPS than sites 1, 2, and 4 ($p<0.05$), but similar to the impacts to sites 5 through 7. Site 4 impacts had the largest MPS magnitudes, that were significantly larger than impacts to site 3, 5, 6, and 7 ($p<0.05$), but no different from sites 1 and 2. Site 5 produced magnitudes of MPS that significantly different from sites 4, 6, and

7 ($p<0.05$), but similar to those at sites 1 through 3. The impacts to site 6 were significantly lower in magnitude than impacts to sites 1, 2, 4, and 5 ($p<0.05$), and no different from sites 3, and 7. Impacts to site 7 were of lower magnitude than hits to sites 1, 2, and 4 ($p<0.05$), and similar to impacts to sites 3, 5, and 6.

Helmet comparisons

Significant main effects were found for all helmets at all impact locations ($p<0.05$). Helmet D (178.0 g) produced the lowest magnitudes of linear acceleration in comparison to the other helmets at site 1 ($p<0.05$), with helmet C the largest (240.6 g) ($p<0.05$). When examining the rotational acceleration results, helmets A (9.9 krad/s^2), B (10.4 krad/s^2), and D (7.3 krad/s^2) produced the lowest results ($p<0.05$). Helmet C (14.9 krad/s^2) had the largest response but was no different from helmet B ($p>0.05$). Helmet A (0.276) and D (0.207) had the lowest magnitudes of MPS from impacts to site 1 ($p<0.05$). However, helmet A was no different from helmet B ($p>0.05$), and helmet C had the largest magnitudes (0.417) ($p<0.05$). For site 2, helmet D (130.6 g and 11.4 krad/s^2) produced the lowest magnitudes of linear and rotational acceleration, with helmet C (278.8 g and 30.1 krad/s^2) the largest ($p<0.05$). The results for MPS showed that helmets A, B, and D were very similar in responses, with only helmet C being significantly different ($p<0.05$). Site three only had significant main effects between linear and rotational acceleration metrics. Helmet D (113.5 g and 8.4 krad/s^2) produced the lowest magnitudes of linear and rotational acceleration ($p<0.05$), with the other helmets having very similar responses ($p>0.05$).

Impacts to site 4 had significant main effects for just the linear and rotational acceleration results. Helmets A (192.0 g), B (157.8 g) were not significantly different from helmets C (210.5 g) and D (151.8 g) for linear acceleration ($p>0.05$). Helmet D was lower in response in

comparison to helmet C however ($p<0.05$). For the rotational acceleration results, helmet D (13.8 krad/s^2) produced significantly lower magnitudes ($p<0.05$) than the other helmets, which were similar in results ($p>0.05$). The linear acceleration results for site 5 were similar for all the helmets ($p>0.05$). Helmet A (14.6 krad/s^2 and 0.285) and helmet B (8.3 krad/s^2 and 0.244) were significantly different ($p<0.05$) when rotational acceleration and MPS was used as the comparative metric, with all other helmets having similar results ($p>0.05$). Helmets A (50.3 g) had lower magnitudes of linear acceleration in comparison to helmets B (81.7 g) and D (92.5 g) ($p<0.05$), but similar to C (67.8 g) ($p>0.05$). Helmet C was also found to have lower magnitudes of linear acceleration in comparison to helmet D ($p<0.05$), but similar to the other two helmets ($p>0.05$). When comparing the helmets using rotational acceleration, helmets A (5.1 krad/s^2), B (6.4 krad/s^2), and C (6.3 krad/s^2) were similar in response ($p>0.05$) and lower in magnitude than helmet D (10.0 krad/s^2) ($p<0.05$). The MPS results at this impact location were the only one to have the peaks located in the white matter instead of the grey matter. The strain results demonstrated that helmet A (0.192) had lower magnitudes than helmets C (0.228) and D (0.233) ($p<0.05$), but similar to B (0.193) ($p>0.05$). At site 7 significant main effects were found only for the linear and rotational acceleration responses. At this site, helmet A (191.2 g) had the largest magnitude response of linear acceleration, with the other helmets being similar to each other ($p>0.05$). When the helmets were compared using rotational acceleration, helmet A (11.3 krad/s^2) had results that were larger in magnitude than helmet B (4.8 krad/s^2), but similar to helmets C (6.5 krad/s^2) and D (6.4 krad/s^2) ($p>0.05$). Helmet B had similar magnitudes of rotational acceleration to helmets C and D ($p>0.05$).

Table 3. Dynamic response and brain strain results for impact sites 1 – 3. Numbers in brackets denote standard deviation.

Impact site	Helmet model	Peak acceleration		Maximum principal strain	
		Linear (g)	Rotational (krad/s^2)	Grey matter	White matter
1	A	178 (22.2)	9.9 (1.7)	0.276 (0.03)	0.227 (0.01)
	B	143.4 (13.5)	10.4 (2.3)	0.287 (0.03)	0.374 (0.04)
	C	240.6 (24.3)	14.9 (2.3)	0.417 (0.04)	0.310 (0.04)
	D	77.6 (9.4)	7.3 (0.8)	0.207 (0.01)	0.187 (0.004)
2	A	227.1 (13.3)	22.2 (0.8)	0.282 (0.03)	0.224 (0.007)
	B	193.6 (12.4)	20.7 (0.4)	0.275 (0.002)	0.246 (0.003)
	C	278.8 (20.7)	30.1 (0.7)	0.355 (0.01)	0.286 (0.004)
	D	130.6 (18.5)	11.4 (0.5)	0.244 (0.01)	0.223 (0.02)
3	A	215.8 (34.4)	17.3 (3.2)	0.221 (0.02)	0.176 (0.008)
	B	177.0 (27.7)	15.6 (2.2)	0.238 (0.02)	0.211 (0.02)
	C	188.3 (2.7)	15.7 (0.4)	0.231 (0.003)	0.224 (0.02)
	D	113.5 (11.5)	8.4 (1.4)	0.204 (0.01)	0.210 (0.009)

Table 4. Dynamic response and brain strain results for impact sites 4 – 7. Numbers in brackets denote standard deviation.

Impact site	Helmet model	Peak acceleration		Maximum principal strain	
		Linear (g)	Rotational (krad/s^2)	Grey matter	White matter
4	A	192.0 (35.3)	22.9 (3.2)	0.336 (0.03)	0.277 (0.02)
	B	157.8 (5.8)	19.2 (0.7)	0.351 (0.01)	0.294 (0.01)
	C	210.5 (6.8)	23.8 (0.2)	0.328 (0.06)	0.298 (0.01)
	D	151.8 (18.3)	13.8 (1.7)	0.274 (0.05)	0.233 (0.03)

5	A	140.4 (23.4)	14.6 (3.8)	0.285 (0.008)	0.253 (0.04)
	B	99.7 (11.0)	8.3 (1.4)	0.244 (0.007)	0.168 (0.01)
	C	133.9 (15.8)	13.7 (0.9)	0.251 (0.03)	0.251 (0.03)
	D	94.8 (23.0)	10.4 (2.0)	0.261 (0.01)	0.220 (0.01)
6	A	50.3 (3.4)	5.1 (0.2)	0.168 (0.006)	0.192 (0.04)
	B	81.7 (13.9)	6.4 (1.0)	0.170 (0.007)	0.193 (0.02)
	C	67.8 (4.7)	6.3 (0.6)	0.223 (0.002)	0.228 (0.006)
	D	92.5 (11.4)	10.0 (1.3)	0.220 (0.04)	0.233 (0.01)
7	A	191.2 (24.4)	11.3 (3.7)	0.242 (0.05)	0.192 (0.04)
	B	96.5 (12.9)	4.8 (0.2)	0.186 (0.01)	0.172 (0.01)
	C	86.8 (5.0)	6.5 (1.3)	0.213 (0.04)	0.182 (0.02)
	D	109.4 (8.3)	6.4 (0.3)	0.197 (0.009)	0.178 (0.01)

Correlation results

The results of the Pearson correlation showed a significant high positive correlation between linear and rotational acceleration for this dataset (**Table 5**). In addition, correlations of linear and rotational acceleration to the grey matter MPS values were significant and high, with correlations to white matter strains being lower at moderate ($p<0.05$).

Table 5. Pearson correlation results for linear acceleration, rotational acceleration, and maximum principal strain for all impacts.

		Peak resultant acceleration		Maximum principal strain	
		Linear	Rotational	Grey matter	White matter
inear acceleration	Pearson Correlation	-	0.863	0.707	0.482
	Sig. (2-tailed)	-	0.001	0.001	0.001

	N	-	84	84	84
Rotational Acceleration	Pearson Correlation	0.863	-	0.713	0.541
	Sig. (2-tailed)	0.001	-	0.001	0.001
	N	84	-	84	84

Discussion

Impact site analysis

The impact sites for this research were chosen based on two criteria: impacts ensuring a full quantification of the helmet performance in all areas of the shell; and impacts that were matched in location to real concussive events from professional baseball games. As a result, 7 sites were impacted in this baseball helmet performance methodology, with sites 1 through 3 representing the more common standards-based impact sites through the centre of gravity, and sites 4 through 7 representing the impact sites that were developed from concussive impacts in baseball games from video analysis. In comparison to the literature the magnitudes of impact found for all sites would be in ranges that would suggest a high likelihood of concussion (Zhang et al., 2004; Kleiven et al., 2007; Rousseau, 2014), but below those where skull fracture might be expected as a result (Yoganandan and Pintar, 2004). There were very high rotational accelerations incurred for sites 2 and 4 that were in the region of risk for incurring a subdural hematoma for many of the helmets impacted (Doorly and Gilchrist, 2006; Post et al., 2014). When comparing the sites to each other, significant differences were present, but not between all sites, which is likely a result of the high standard deviations caused by collapsing all the helmet results into one analysis. Also of interest was that the upward angled impact (site 6) that represented a player ducking away from a ball impact had the largest strains in the white matter as opposed to grey matter. As the white matter is often suggested as a focal point for concussion

(Wright et al., 2012; Murugavel et al., 2014; Shin et al., 2014) this impact site in particular might be of interest for future helmet designs. However, this methodology does indicate a significant risk of concussion for all the impact sites, which demonstrates the lack of protective capacity for this type of injury for all the helmets examined against these impacts from baseballs at high velocity.

Helmet analysis

Four different models of helmets were evaluated using the seven impact site protocol. Overall, while the helmets undoubtedly reduced the magnitudes of impact from a severe brain injury range, there was still a significant risk of concussion regardless of helmet design. In many cases there was little to distinguish between the helmets when linear acceleration or MPS was used as the performance metric. Rotational acceleration however, showed more significant differences between the helmet models which suggests that this metric might be the most sensitive to changes in helmet designs. This is also pertinent as rotational acceleration has often been suggested as an important contributor to the occurrence of concussion (Gennarelli et al., 1987; Kleiven, 2013; Lamy et al., 2013). The fact that maximum principal strain does not distinguish in many cases between the helmets might be a result of the combination of linear and rotational acceleration for these impacts causing a very similar amount of strain in the brain, and thus masking helmet designs that would reduce one type of motion over the other. The helmets themselves all had a type of vinyl nitrile or expanded polypropylene liner and a hard shell to help deflect and spread out the impact forces. When looking at the effect of liner type, there seems to be little difference between helmet energy absorbing liner types. Interestingly helmet D, which had a very hard composite shell, was by far the best helmet for direct impacts. This suggests that for this type of projectile impact that having a very hard shell to help distribute the impact is a

desirable design consideration. This is evidenced in **figure 5**, where the typical helmet impact with a baseball is over before 1.2 ms, but with a harder shell this head acceleration is extended to 2 ms. Lengthening the duration of the impact helps engage more foam and reduce the magnitudes of the impact and reduce risk of injury. This benefit is reduced for impacts that are of a more glancing nature (sites 4 through 7) that represented actual concussive impacts within the sport, and was worst in rotational acceleration for site 6. It is likely that having a stiff shell that is beneficial for direct impacts must be complimented with improved energy absorbing liner technologies for when the impact is applied in a more tangential fashion, thus reducing the effect of the shell materials.

All of the helmet impacts were represented by very short duration (2 ms or less) linear and rotational acceleration loading curves. These curves resulted in MPS magnitudes in the UCDBTM that represented a considerable risk of injury (Zhang et al., 2003; Kleiven, 2007; Rousseau, 2014). This is a unique phenomenon specific to this mechanism of injury, as in other sports (ice hockey and American football), similar magnitudes of strain have been demonstrated, but for much longer duration acceleration loading curves (15 to 30 ms) (Kendall et al., 2012a; Zanetti et al., 2012; Rousseau 2014). It is likely that concussion in baseball is a result of these short magnitude events and thus to reduce the risk that helmet designs specific to managing this type of acceleration loading curve (lengthening and reducing magnitude) would be desirable. Because of the unique nature of the mechanism of concussion in baseball it is unlikely that applying technologies designed to reduce concussion from sports with differing mechanisms of injury without any adjustments would be appropriate.

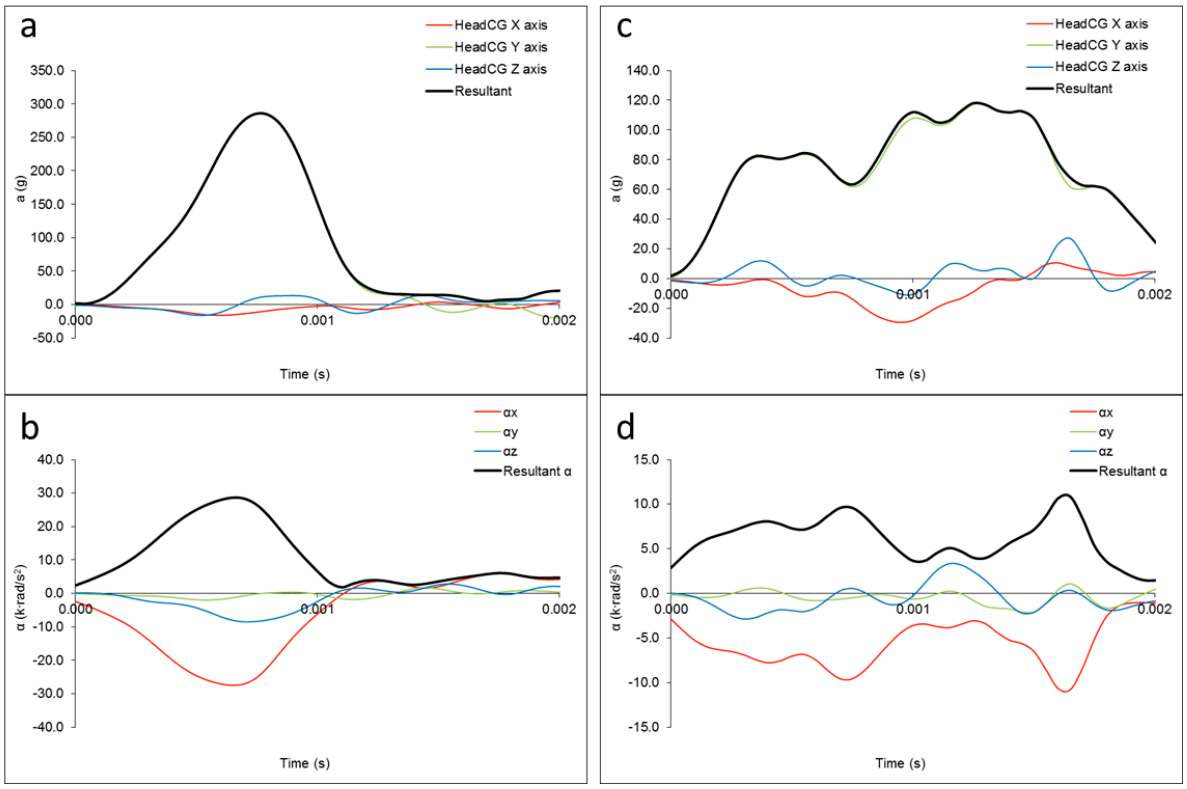


Figure 5. Comparison of the acceleration loading curves for helmet C (a and b) and D (c and d) for the site 2 impact

Correlations

A Pearson correlation analysis was conducted on the variables to determine which metrics might be most suitable for use as the performance criteria for this protocol. For these projectile impacts linear and rotational acceleration was highly correlated, which differedfrom literature discussing these types of correlations for ice hockey and American football impacts (Post et al., 2011; Post et al., 2013a; Post et al., 2014b). While linear and rotational acceleration was correlated in this particular research, rotational acceleration consistently distinguished between helmet designs and should likely be included as a variable for future analyses. When the relationships between linear and rotational acceleration and grey matter MPS were examined, both had high correlations to the strain in the brain tissue. This suggests that both linear and

rotational acceleration contributed to the presence of risk for this particular mechanism and that is similar to conclusions from the present and other authors in the literature (Bandak and Eppinger, 1994; Post and Hoshizaki, 2012). When looking specifically at the white matter correlations, the relationships are less definitive and moderate, which may be a representation of the material characteristics that represent the white matter in the brain. This indicates that the influence of these accelerations may vary depending on their interactions with the brain tissue, which may be relevant in future concussion analyses.

Summary and conclusion

To the authors knowledge is research represents the first examination of the performance of baseball helmets under loading conditions that commonly result in concussion. The results have shown that the methodology proposed used a series of impact sites that were successful in distinguishing differences in design characteristics between the baseball helmets in linear and rotational acceleration, and in many cases maximum principal strain. In addition, for impacts at 80 mph, these impact sites show a consistent significant level of risk regardless of the helmet worn, indicating limited protection by these helmets for concussive impacts. Notably, a stiffer shell does tend to reduce the magnitudes for direct impact but has less of an effect for impacts of a more tangential nature. Finally, the mechanism of concussion shown by this research are baseball impacts that result in very brief (2 ms or less) acceleration loading curves. These curves are distinct from those representing concussion in other sports, suggesting the understanding of concussion and therefore development of protective helmet technologies must be specific to baseball.

Limitations

The approach used in this research to establish and impact protocol and evaluate the performance of baseball helmets had some limitations. The use of the Hybrid III head and neckform for this type of analysis has become common for brain injury research, and evaluations of helmet performance in both linear and rotational acceleration. While being a reliable test device that provides accurate accounts of the motion of the head, and while it has been shown to produce results in the range of cadaveric human impacts (Kendall et al., 2012b), it is not a true representation of the human head motion from an impact. In addition, the use of the UCDBTM, while being one of few partially validated finite element models of the human brain, is limited in many of the material characteristics and assumptions are based upon cadaveric and other anatomical testing that may not be fully representative of living human responses. As such, these limitations should be taken into consideration when evaluating the results of this research.

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