THE FUTURE OF TRANSPORT

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Development of a New Cycle Helmet Assessment Programme (NCHAP)

Literature Review

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Contents

Introduction

This document is comprised of the electronic appendices to the executive summary report: “PPR920 Development of a New Cycle Helmet Assessment Programme (NCHAP): Summary Report”.

Appendix A provides a literature review of cyclist accidentology covering key areas such as cyclist demographics, helmet types/wearing rates, head/helmet impact locations and rates, collision impact partners and causes of collisions.

Appendix B summarises the key features of the major cycle helmet testing standards from around the world. An overview of the main legal requirements in force for each country a review of these standards is also provided within this appendix.

Finally, Appendix C is a review of the state-of-the-art in head injury criteria, which provides an overview of the brain injury and skull fracture continuums, before summarising the various injury mechanisms associated with these injuries.

Appendix A Cyclist Accidentology Review

A.1 Introduction

This Appendix summarises the key data of the conditions and circumstances surrounding bicycle accidents as described in key literature and reports. The review covers areas including:

- Cyclist demographics
- Helmet types and wearing rates
- Head/helmet impact locations and rates
- Collision impact partners
- Cause of collisions
- Speed of bicycle and vehicle (if applicable)
- Environmental conditions

A.2 Methodology

A series of accident analysis research questions were generated to provide a list of relevant search terms. A total of 80 pieces of literature were sourced and after initial assessment 71 were found to provide relevant accident data. Literature was rejected if the data was not strictly related to the research questions (e.g. anthropometric data of children's bicycles) or the accident data was a duplicate of a later study.

A.3 Cycling Participation

Cycling can provided a healthy and environmentally friendly mode of transportation. A survey of Attitudes of Europeans Towards Urban Mobility showed that on average half of Europeans[1] cycle, with 12% cycling at least least once a day, 17% cyling a few times a week and 20% cycling less than that (European Commission , 2013). The Netherlands have the highest proprtion of daily cyclists (43%) followed by Denmark (30%) and Finland (28%). The UK has one of the lowest rates of daily cycling at only 4%. In the UK 10% of respondants cycled a few times a week and 17% cycled a few times a month or less.

Similar rates of cycling amongst adults were recorded in The British Social Attitudes survey in 2016 with 31% of respondants reporting cycling activity (Table A.1) (Department for Transport, 2017). The Department for Transport statistics on Walking & Cycling in 2015/2016 showed that over 38% of respondants cycled at least once a month (Department for Transport, 2018).

[1] EU28 countries included Belgium, Lithuania, Bulgaria, Luxembourg, Czech Republic, Hungary, Denmark, Malta, Germany, The Netherlands, Estonia, Austria, Greece, Poland, Spain, Portugal, France, Romania, Croatia, Slovenia, Ireland, Slovakia, Italy, Finland, Republic of Cyprus, Sweden, Latvia, The United Kingdom

Table A.1: Frequency of cycling in Great Britain

Frequency of cycling	(Department for Transport, 2017) Percentage	Number of adults (18+) (million)	(Department for Transport, 2018) Percentage	Number of adults (16+) (million)
Every day	4	1.9	3.4	1.5
Multiple times a week	5	2.6	5.7	2.5
Once a week	5	2.7	12	5.4
At least once a month	6	3	17	7.6
Less often than once a month	10	5.1	-	-
Never	69	35	-	-

A.4 Cyclist Demographics

A.4.1 Age

An overview of the age of cyclists included in literature can be seen in Figure A.1. The cyclist ages were often grouped and excluding children the majority of cyclists were aged 25 – 65. The studies by Hansen *et al*. (2003), Olofsson *et al*. (2015) and Thomas *et al*., (1994) included children only. The studies by SNELL (1996), Ching et al. (1997), ATSB (2006) and Badea-Romero and Lenard (2012)also included a large proportion of child cyclists.

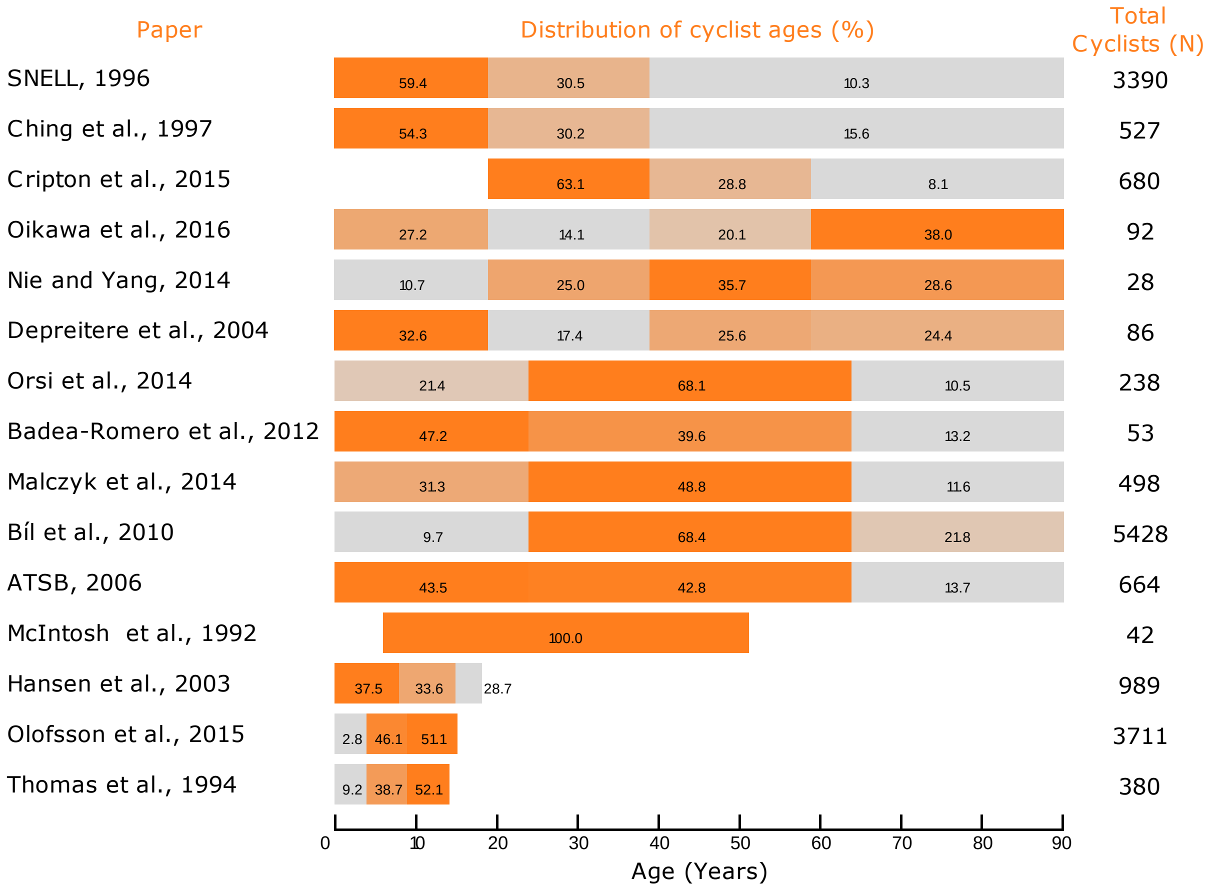

Figure A.1: Distribution of cyclist ages

A.4.2 Gender

In the majority of studies focussing on collisions there was a higher population of men than women (Figure A.2).

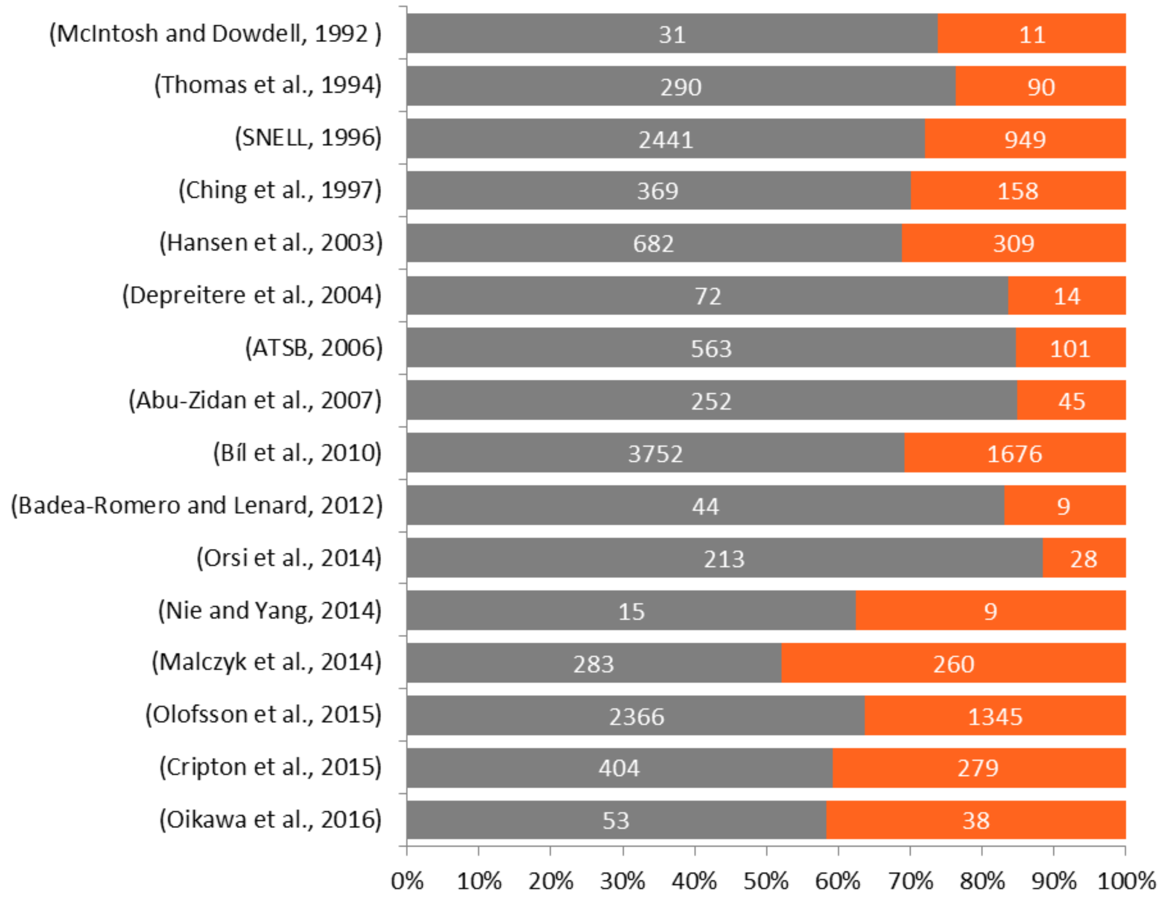

Figure A.2: Gender distribution of cyclists (males in grey, females in orange)

A.4.3 Height/weight

Nie and Yang (2014) documented individual cyclist height and weight (Table A.2); however, it must be noted that this information was extracted from the In-Depth Investigation of Vehicle Accidents in Changsha (IVAC) database so the heights and weights may not reflect the wider population of Europe or the UK. In the majority of cases (83.3%) the sitting height was less than the standing height of the cyclist; on average standing height was 11.1cm greater than sitting height.

Table A.2: Height, sitting height, weight and age of individual cyclists with averages shown in italics. (Nie and Yang, 2014)

Males			
Cyclist height (cm)	Cyclist sitting height (cm)	Cyclist weight (kg)	Age
152	152	54	56
153	151	54	14
158	155	49	65
165	155	54	14
165	139	55	55
168	152	50	47
168	160	67	63
170	172	60	59
170	159	80	58
176	149	82	29
177	155	82	31
178	156	78	67
179	156	74	44
179	158	74	44
185	176	72	25
169.5	*156.3*	*65.7*	*45*

Females			
Cyclist height (cm)	Cyclist sitting height (cm)	Cyclist weight (kg)	Age
156	161	53	59
158	152	49	42
159	173	88	57
160	151	45	14
161	144	45	23
164	146	57	28
168	148	72	77
173	167	110	36
177	165	72	37
164.0	*156.4*	*65.7*	*41*

A.5 Cycle Helmet Wearing Rates and Styles

A.5.1 General Helmet Wearing Rates

Cycle helmet rates are generally quite low however, three large surveys in 2004, 2006, and 2008 showed a significant increase in helmet wearing among British cyclists (Table A.3).

Table A.3: UK Bicycle Helmet Wearing Rates (McGarry and Sheldon, 2008)

Year	Helmet wearing rate Major built up roads	Helmet wearing rate Minor built up roads
2004	28.2%	9.6%
2006	30.7%	13.8%
2008	34.3%	16.7%

Data from a nationwide study into mobility in Germany conducted in 2008 showed that out of the 19,646 participants 12.4% wore helmets (Ritter and Vance, 2011). Men were more likely to wear helmets; 18.3% of males wore helmets compared to 10.6% of females. Those who cycled daily and those who only cycled monthly were more likely to wear a helmet than those who cycled weekly, this trend was apparent in both males and females (Figure A.3).

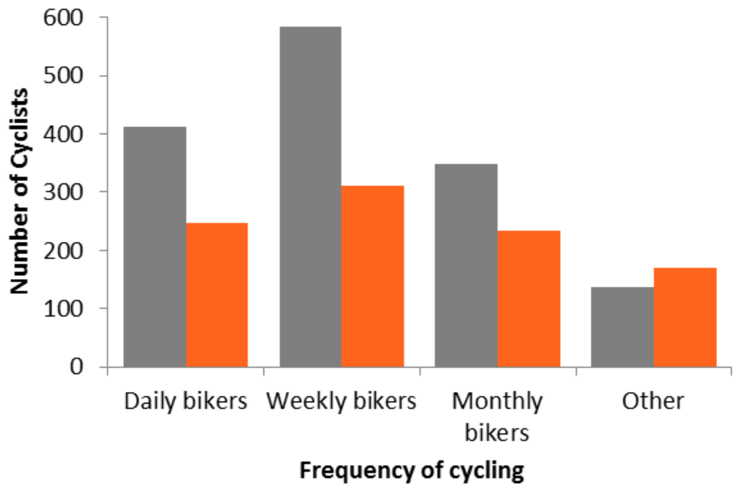

Figure A.3: Helmet wearing rate of cyclists based on frequency of cycling for males (grey) and females (orange) (Ritter and Vance, 2011)

A survey of 235 children and 106 parents in Norway showed that whilst children more commonly owned helmets (91.6%) than adults (83.0%), adults reported more frequent helmet use (Table A.4) (Lajunen, 2016).The helmet wearing rates in this study appear much higher than figures from other studies. This may be due to participants who wore helmets being more likely to reply to the survey.

Table A.4: Self-reported helmet wearing rates of children and parents in Norway (Lajunen, 2016)

How often do you wear a helmet?	Adults (%)	Children (%)
Never	11.3	16.6
Rarely	8.5	17.0
Sometimes	9.4	11.1
Often	15.1	10.2
Always	53.8	43.8

A.5.2 Helmet Wearing Rates during Collisions

The number of cyclists wearing helmets during collisions and the percentage of the sample population they accounted for are shown in Table A.5. It is important to note that Nie and Yang (2014) specified that accident data was only used for cyclists who were not using helmet protection at the time of the accident and conversely McIntosh and Dowdell (1992) and Ching et al. (1997) only included cyclists wearing helmets as their studies focused on helmet impact locations. Excluding these three studies there was an average helmet wearing rate of 33.2% of cyclists involved in collisions. In some papers helmet wearing rates were not included as there was insufficient data recorded at the time of the collision (Badea-Romero and Lenard, 2012). McIntosh and Dowdell (1992) reported that in two fatal collisions helmets were incorrectly worn or fastened.

Table A.5: Number of cyclists wearing helmets during collisions and the percentage of the sample population they accounted for

Paper	Worn		Not Worn		Unknown	
	N	%	N	%	N	%
(Abu-Zidan *et al.*, 2007)	116	*39.1*	63	*21.2*	118	*39.7*
(ATSB, 2006)	30	*26.5*	18	*15.9*	65	*57.5*
(Bourdet *et al.*, 2014)	2	*8.3*	22	*91.7*	0	*0.0*
(Ching *et al.*, 1997)	527	*100.0*	0	*0.0*	0	*0.0*
(Cripton *et al.*, 2015)	472	*69.1*	211	*30.9*	0	*0.0*
(Hansen *et al.*, 2003)*	282	*28.5*	670	*67.6*	39	*3.9*
(Haworth *et al.*, 2010)	10082	*73.1*	2101	*15.2*	1610	*11.7*
(Maimaris *et al.*, 1994)	114	*10.3*	928	*83.8*	65	*5.9*
(Malczyk *et al.*, 2014)	75	*13.8*	360	*66.4*	107	*19.7*
(McIntosh and Dowdell, 1992)	42	*100.0*	0	*0.0*	0	*0.0*
(Nie and Yang, 2014)	0	*0.0*	24	*100.0*	0	*0.0*
(Olofsson *et al.*, 2015)*	2146	*50.5*	1565	*36.9*	535	*12.6*
(Orsi *et al.*, 2014)	36	*14.9*	206	*85.1*	0	*0.0*
(Otte *et al.*, 2015)	433	*10.2*	3812	*89.8*	0	*0.0*
(SNELL, 1996)	1718	*50.7*	1672	*49.3*	0	*0.0*
(Thomas *et al.*, 1994)*	165	*37.1*	238	*53.5*	42	*9.4*

*Studies involving child cyclists only.

A.5.3 Helmet Styles Worn during Collisions

When bicycle helmets were first manufactured they consisted of a thick hard shell with a thin layer of foam liner. It was found that the thick hard outer shell was not necessary and so helmets consisting only of foam and no shell became more popular, especially for children. However, the no-shell helmets often broke up during collisions and so a thin hard shell was added to helmets to improve the integrity of the foam liners. It is thought that by moulding the thin shell with the foam liner the strength and impact management capabilities of the helmet could potentially be increased (BHSI, 2015). Helmet styles were only reported in a small number of papers, the majority of which were over 20 years old so this may not be reflective of the current helmet style wearing rates (Table A.6).

Table A.6: Helmet Styles Worn during Collisions

Paper	Hard Shell		Thin Shell		No Shell	
	N	%	N	%	N	%
(Ching *et al.*, 1997)	260	49.5	180	34.3	85	16.2
(*Hansen et al.*, 2003)*	182	64.5	-	-	100	35.5
(McIntosh and Dowdell, 1992)	21	50.0	6	14.0	15	36.0
(SNELL, 1996)	842	49.0	498	29.0	326	19.0
(Thomas *et al.*, 1994)*	115	87.8	4	3.1	12	9.2

*Studies involving child cyclists only.

A.5.4 Helmet Efficacy

In an attempt to estimate the efficacy of helmets many research groups have calculated the odds ratio (OR) to try to determine the effects of helmet wearing on risk head injury (Table A.7). These studies however are unlikely to be able to calculate a robust efficacy, because it is very difficult to control adequately for exposure and difficult to control for other confounding factors so the following results should be viewed with this in mind.

Head injuries were, however, found to be significantly reduced in helmeted versus un-helmeted cyclists. Thompson *et al*. estimated that helmet use is associated with head injury reductions of 63-88% (Thompson *et al.*, 2006). Dorsch *et al.* (1987) also estimated high reductions of 76% for AIS 1+ head injuries; however they estimated much more conservative values for AIS 2+ head injury (45%) and AIS 3+ head injuries (42%). Cook and Sheik (2003) estimated that helmets prevent 60% of serious head injuries. Wearing a helmet was found to halve the odds of a fatality compared to a hospitalisation of a cyclist involved in a collision (OR 0.49) (Haworth *et al.*, 2010).

Table A.7: Odds ratio for the effects of helmet wearing on head injury

Paper	Odds Ratio	95% confidence interval
(Abu-Zidan *et al.*, 2007)	0.48	0.23 – 1.03
(Amoros *et al.*, 2012)	0.69	0.59 – 0.81
(Attewell *et al.*, 2000)	0.40	0.29 – 0.55
(Bambach *et al.*, 2013)	0.41	0.35 – 0.49
(Dinh *et al.*, 2013)	0.22	0.09 – 0.57
(Dinh *et al.*, 2015)	0.34	0.15 – 0.76
(Dorsch *et al.*, 1987)	0.24	0.11 – 0.49
(Hansen *et al.*, 2003)	0.50	0.33 – 0.77
(Haworth *et al.*, 2010)	0.40	–
(Jacobson *et al.*, 1998)	0.37	0.19 – 0.73
(Maimaris *et al.*, 1994)	0.30	0.08 – 0.82
(McDermott *et al.*, 1993)	0.50	0.36 – 0.69
(Thomas *et al.*, 1994)	0.49	0.10 – 0.78
(Thompson *et al.*, 1989)	0.25	0.15 – 0.43
(Thompson *et al.*, 1996)	0.30	0.24 – 0.36

A.6 Head Injuries

Head injuries are often the most common injury type sustained in cyclist collisions. Nie and Yang (2014) reported that the head accounted for 32% of injuries sustained.

Head injuries were sustained by both helmeted and un-helmeted cyclists and generally there were higher proportions of cyclists with head injuries who were not wearing a helmet at the time of collision (Table A.8).

Table A.8: Number of head injuries sustained during cyclist collisions

Paper	Head Trauma		Head Trauma with helmet		Head Trauma without helmet	
	N	%	N	%	N	%
(Abu-Zidan *et al.*, 2007)	75	*25.3*	-	-	-	-
(Badea-Romero and Lenard, 2012)	55	*33.3*	-	-	-	-
(Haworth *et al.*, 2010)	1157	*100.0*	727	*62.8*	347	*30.0*
(Ching *et al.*, 1997)	311	*59.0*	-	-	-	-
(Depreitere *et al.*, 2004)	86	*100.0*	3	*3.5*	56	*65.1*
(Hansen *et al.*, 2003)*	279	*28.1*	59	*21.9*	220	*81.5*
(Maimaris *et al.*, 1994)	104	*13.5*	4	*3.8*	100	*96.2*
(Malczyk *et al.*, 2014)	239	*54.8*	34	*7.8*	155	*35.6*
(McIntosh and Dowdell, 1992)	10	*23.8*	10	*100.0*	0	*0.0*
(Nie and Yang, 2014)	15	*62.5*	-	-	15	*62.5*
(Oikawa *et al.*, 2016)	47	*51.0*	-	-	-	-
(Olofsson *et al.*, 2015)*	1020	*27.5*	437	*42.8*	583	*57.2*
(Orsi *et al.*, 2014)	129	*56.6*	11	*10.0*	118	*90.0*
(Otte *et al.*, 2012)	1409	*35.6*	-	-	-	-
(Otte *et al.*, 2015)	1581	*41.5*	129	*8.2*	1452	*91.8*
(SNELL, 1996)	1374	*40.5*	-	-	-	-
(Thomas *et al.*, 1994)*	102	*26.2*	31	*7.9*	67	*17.2*

*Studies involving child cyclists only.

A.6.1 Head Injury Severity

The Abbreviated Injury Scale (AIS) was used in numerous papers to describe the severity of head injuries. Head injuries ranged between superficial lesions (AIS 1), to skull fractures and moderate brain injury (AIS 2), to the more serious brain injuries (AIS 3-5) all the way to catastrophic crush type injuries (AIS 6).

Data from the On the Spot (OTS) research study showed that of 165 cyclists involved in a road traffic accident 55 sustained head injuries (Badea-Romero and Lenard, 2012). The head injury data was divided into two groups; head impacts with the road/ground and head impacts with a vehicle (Table A.9). On the whole head impacts with the road were found to be more numerous but less severe. Out of 55 cyclists 10 sustained head injuries with an AIS score of ≥2, 7 of which were sustained with an impact to the road and 3 with an impact with a vehicle. There was no information included on helmet use.

Table A.9: Maximum AIS scores of head impacts with the road or a vehicle (Badea-Romero and Lenard, 2012)

Max head AIS	Head Impact partner	
	Road or ground	Vehicle
Nil	3	3
AIS 1	28	10
AIS 2	4	0
AIS3+	3	3
Unknown	0	1
Total	38	17

In the study by Hansen et al. (2003) 281 out of 991 cyclists (56.6%) received head injuries with AIS score of ≥2. Only 59 of the 281 cyclists (21.0%) were wearing a helmet at the time of the collision. Of the head injuries sustained 155 were due to a fall, 90 were due to a collision with an obstacle and 28 were due to a collision with a car. 106 of the 281 cyclists that obtained a head injury also sustained facial injuries.

Talbot et al. (2014) described the maximum AIS score for the head and neck of fatally injured cyclists in London. Of the 46 cyclists 28 sustained head and neck injuries with an AIS score of ≥2. 61% of the cyclists sustained head and neck injuries with an AIS score of ≥3 and 49% with an AIS score of ≥4. Only 14 cyclists were wearing helmets at the time of the collision, with one helmet worn incorrectly. Talbot et al. (2014) identified 7 cases where wearing a correctly fitted helmet may have reduced or mitigated the head injuries sustained, but not prevented them entirely. A database linking STATS19 data with Hospital Episode Statistics data highlighted that the head was the most common body region injured in cyclists involved in collisions. Of 16,011 injured cyclists 12528 injuries to the head were reported, almost twice the number of upper extremity (6569) and lower extremity (6675) injuries.

Olofsson et al. (2015) studied helmet use and injuries sustained by children involved in bicycle collisions. Of a total of 3711 children, 363 sustained head injuries with as AIS score of ≥2. 157 of these children were wearing a helmet, approximately 43%.

In the study conducted by Malczyk et al. (2014) 239 cyclists sustained head injuries, 184 (77.0%) sustained AIS 1 head injuries including soft tissue injuries and concussion. Head injuries with AIS level 2 were sustained by 32 (13.4%) of cyclists, AIS level 3 by 23 (9.6%) of cyclists and AIS level 4 by 2 (0.8%) cyclists. Only 34 (18.5%) of cyclists with head injuries were wearing a helmet at the time of the collision, and none of the cyclists with AIS 3 injuries were helmeted. Whilst AIS 1 concussion rates were found to be higher in helmeted cyclists, only 1 out of 34 helmeted cyclists sustained cerebral injuries or skull fractures.

A.6.2 Head Impact Locations

Malczyk et al. (2014) deduced the points of impact on the head during a collision from the location of soft tissue injuries, documenting the frequency and distribution of impacts of both helmeted and un-helmeted cyclists. Generally fewer areas in the temporal and parietal region presented injuries in helmeted cyclists; however helmeted cyclists sustained a higher frequency of injuries in the frontal region. Malczyk et al. (2014) assumed this higher frequency of frontal injuries was caused by the impacted forces on the helmets being transferred to the skull by the inner plastic head band. The regions around the eyebrows and eyes were affected less frequently in helmeted cyclists. Soft tissue injuries were not detected at the immediate top region of the skull in either helmeted or un-helmeted cyclists. Finally, Depreitere et al. (2004) deduced the location of head impacts through CT scans, x-rays and scalp injuries and found that 57% of impacts occurred at the side (parietal, temporal and frontotemporal regions) and 27% of impacts occurred at the front of the head.

A.6.3 Helmet Impact Locations

Several studies have focussed on the impact locations on helmets Table A.10. The frontal and lateral regions appear to sustain the highest frequencies of impact damage. Accident analyses performed by Oikawa *et al.* (2016) also showed that the most frequent helmet impact locations were the side and front. When the lateral region is split further into left and right lateral regions the frequency of impacts are closer to those found in the vertex and occipital regions.

Table A.10: Helmet impact locations

Paper	Frontal		Lateral		Left Lateral		Right Lateral		Vertex		Occipital	
	N	%	N	%	N	%	N	%	N	%	N	%
(Ching *et al.*, 1997)	285	*40.9*	212	*30.5*	108	*15.5*	104	*14.9*	95	*13.6*	133	*19.1*
(Malczyk *et al.*, 2014)	16	*20.8*	19	*24.7*	13	*16.9*	6	*7.8*	10	*13.0*	13	*16.9*
(Otte *et al.*, 2015)		*24.4*		*63.4*		*35.4*		*28.0*		*7.3*		*4.9*
(McIntosh and Dowdell, 1992)		*27.0*		*56.0*						*13.0*		*4.0*

Malczyk et al. (2014) established helmet impacts from scratches, dents or fractures present on the helmets. In total 58 impacts were determined from 26 helmets (Figure A.4). The frontal region of the lower rim of the helmet was found to be the most frequently damaged area, closely followed by the left lateral lower rim region, these two regions accounted for almost a third of the total impact frequency.

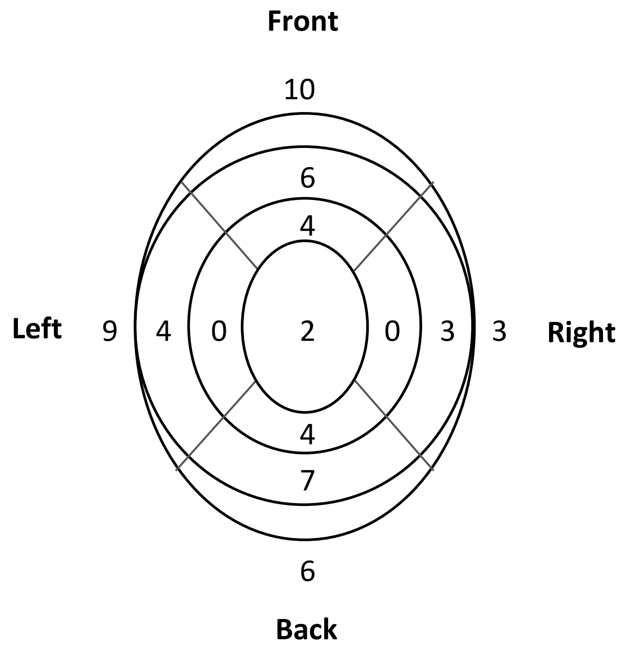

Figure A.4: Impact distribution of 26 bicycle helmets (adapted from Malczyk et al. (2014))

McIntosh and Dowdell (1992) studied the primary impact sites of 42 helmets. They found that primary impacts were most frequent over the frontal/temporal region (67%) (Figure A.5). Impacts resulting in head injuries of AIS ≥2 occurred most frequently in region 4 (70%), region 5 (20%) and region 1 (10%).

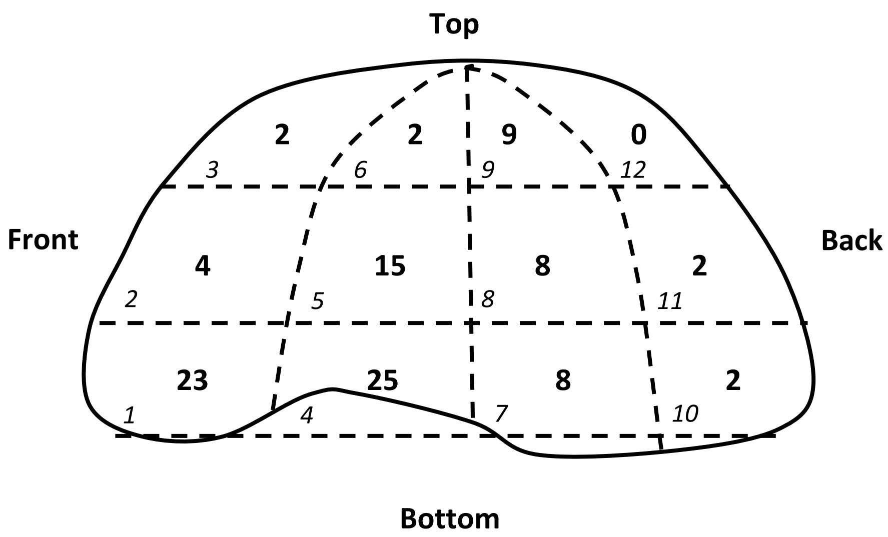

Figure A.5: Location of helmet impacts split across 12 regions (in brackets). All impacts transposed to one side and presented as percentages. Adapted from McIntosh and Dowdell (1992)

Otte *et al.* (2015) documented the damage to 134 helmets by splitting the severity of the damage into three categories: cracks/breakages, deformations/indentations and smears/abrasions. Otte *et al.* (2015) reported similar locations of helmet damage to Malczyk and colleagues; both indicate that the top of the helmet is very rarely subjected to an impact. The lateral edges suffered the most frequent amount of damage. Cracks and breaks were most predominant on the lateral edges and the upper lateral regions. Inexplicably the left side of the helmet suffered more frequent damage than the right side. There was no significant difference found between the type of collision and the helmet damage observed. The helmet damage was also recorded against the severity of head injury sustained by the cyclist. The left and right lateral impacts were combined to focus on lateral damage as a whole. For cases where no head injury was sustained there were slightly more impacts to the top of the helmet (59%) than to the sides (41%).For cases which cyclists sustained head injuries of AIS 1 or 2 there was an approximately equal frequency of impacts on the top and sides of the helmet. However for cases where head injuries of AIS 3+ were sustained all the impacts were to the sides of the helmet. This suggests a correlation between side impacts to the helmet leading more severe head injury; however this is based only on 6 instances of AIS3 + injuries.

Ching *et al.* (1997) recorded the damage of 527 helmets on a 2D diagram split into 3 rings and 6 segments Figure A.6). The outermost ring covered damage within 1" from the edge of the liner, the middle ring for damage between 1" from the edge and half the distance to the vertex and the innermost ring covered the rest of the remaining area. Multiple areas of damage were scored separately.

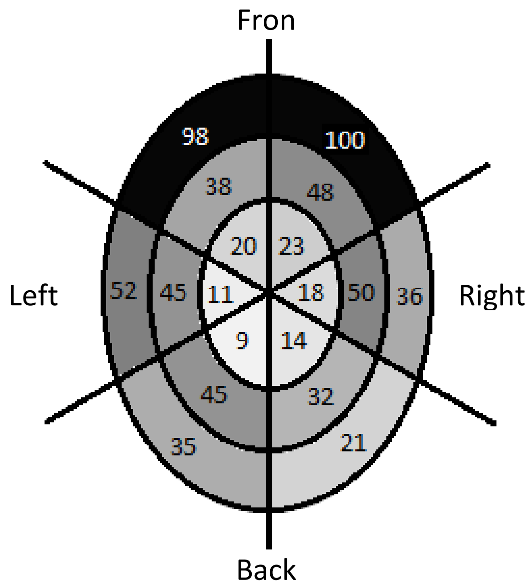

Figure A.6: Number and location of 527 helmet impacts. The Scale shows the number of impacts to the helmet at that particular location. White indicates no impacts and black indicates 100 impacts. Adapted from Ching et al. (1997)

Ching et al. (1997) found an association between the extent of damage on the helmet and the risk of head injury. A damage score was assigned to each helmet and plotted against the risk of head injury (Figure A.7). Once the helmet was subjected to major damage where the liner was compressed or cracked the risk of head injury more than doubled.

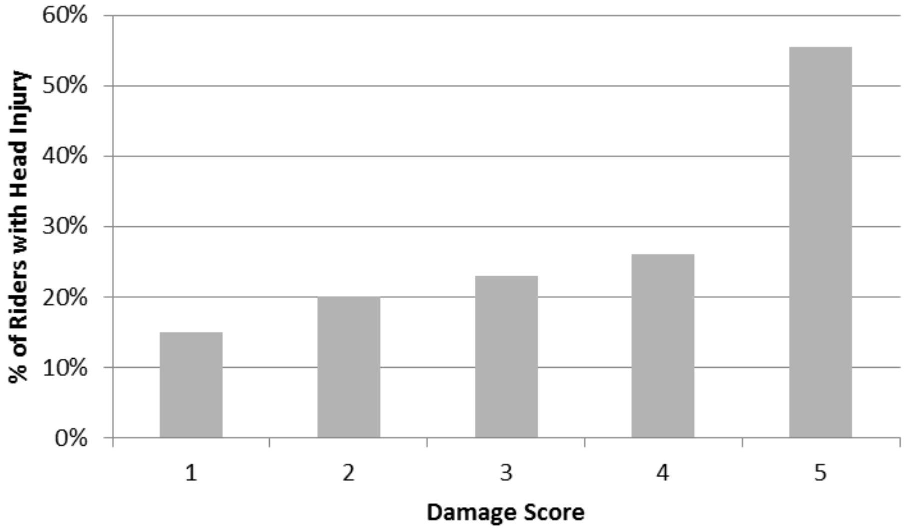

Figure A.7: Risk of Head Injury in helmeted cyclists according to damage score of the helmet (adapted from Ching et al., (1997))

A.6.4 Head Impact Angle

The cyclist head impact angles in the study by Nie and Yang (2014) ranged from 10.9° to 83.5°, with a mean value of 45.7°, which were comparable with the head impact angles described by Peng *et al*. (2012) 11° to 73°, with a mean value of 38.4°. The head impact angle distributions calculated by Bourdet et al. (2014) are shown in Figure A.8 and centred around 33±20°.

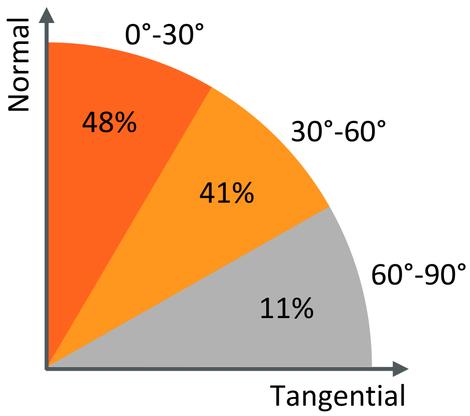

Figure A.8: Head impact angle distribution from 26 accident reconstructions (adapted from Bourdet et al., (2014))

A.7 Collision Circumstances

A.7.1 Cause of Collision

Single-vehicle collisions or falls were one of the most common causes of collision along with collisions involving a motor vehicle Table A.11. Malczyk *et al.* (2014) suggest that the because cyclist injury analyses is often based on police reported data which tends to focus on collisions involving motor vehicles, the recorded proportion of collisions involving vehicles may be elevated. Data based on hospital documentation coupled with information on the crash circumstances may provide a more realistic reflection of the causes of collisions involving cyclists. Nie and Yang (2014) specified that accident data was only used for cyclists whose head impacted the bonnet, windscreen or roof of a vehicle and so no other cause of collision was included.

Table A.11: Collision Causes

Paper	Motorised vehicle		Fall		Stationary Object		Cyclist/ pedestrian / animal		Non - Motorised Vehicle		Tram/ tracks	
	N	%	N	%	N	%	N	%	N	%	N	%
(Abu-Zidan *et al.*, 2007)	107	*36.0*	163	*54.9*	10	*3.4*	17	*5.7*				
(ATSB, 2006)		*86.0*		*5.0*		*4.0*		*1.0*		*0.0*		*1.0*
(Badea-Romero and Lenard, 2012)	52	*94.5*	3	*5.5*								
(Haworth *et al.*, 2010)	12252	*91.8*	684	*5.1*	114	*0.9*	197	*1.5*	104	*0.78*		
(Ching *et al.*, 1997)	77	*14.6*	246	*46.7*	176	*33.4*	28	*5.3*				
(Cripton *et al.*, 2015)	231	*34.0*	177	*26.0*	138	*20.0*	40	*5.9*			97	*14.0*
(Depreitere *et al.*, 2004)	44	*51.2*	42	*48.8*								
(Hansen *et al.*, 2003)	73	*7.4*	557	*56.2*	291	*29.4*						
(Maimaris *et al.*, 1994)	288	*27.6*	662	*63.5*			20	*1.9*	72	*6.9*		
(Malczyk *et al.*, 2014)	102	*18.8*	275	*50.6*			80	*14.7*				
(McIntosh and Dowdell, 1992)	26				16							
(Nie and Yang, 2014)	15	*100.0*										
(Olofsson *et al.*, 2015)	172	*4.6*	3184	*85.8*	81	*2.2*	266	*7.2*				
(Orsi *et al.*, 2014)	137	*56.6*	59	*24.4*					46	*19.0*		
(Otte *et al.*, 2012)	2742	*69.2*			1222	*30.8*						
(SNELL, 1996)	518	*15.3*	1695	*50.0*	980	*29.0*	197	*5.8*				
(Thomas *et al.*, 1994)	31	*15.4*			170	*84.6*						

When collision cause was linked to head injury severity Malczyk *et al.* (2014) found that falls or single vehicle crashes accounted for approximately 56% of collisions where cyclists sustained AIS 1 and AIS 2 head injuries but only 40 % of AIS 3+ head injuries. Instead, the proportion of collisions involving motorised vehicles increased with the severity of head injuries. Collisions involving motorised vehicles accounted for 47 % of AIS 3+ head injuries.

Isaksson-Hellman and Werneke (2016) calculated the risk of sustaining injuries of AIS 3-5 during collisions involving cyclists and cars. They proposed that there is a 1% risk of AIS 3-5 head injury in collisions where a bicycle and car cross paths and almost a 5% risk when a collision occurs due to the bicycle and car moving in the same/opposite directions.

A.7.2 *Impact Partners*

Single vehicle collisions or falls were often found to be the most common collision type; McIntosh *et al.* (1992) reported that in the 16 instances of single vehicle collisions head impacts were against flat rigid objects such as the road or a stationary object on the roadside. Head injuries (AIS 2+) were sustained by 5 (31.3%) cyclists.

Vescheuren *et al.* (2009) included data for 11 cyclists who fell off their bicycle and in all cases the cyclists head or helmet impacted the ground.

The study by Shepers et al. (2012) into single vehicle bicycle crashes found that infrastructure related crashed accounted for just over half the crashes within the sample (n=350 ; 52.3%). These included collisions with bollards, opening car doors, parked cars, temporary fencing (n=77) as well as the front wheel or pedal hitting the kerb and obstacles on the road shoulder such as trees, lamp posts and fences (n=142).

Whilst single vehicle collisions or falls account for a significant number of serious and sometimes fatal injuries, it is indicated in several studies that collisions involving a motor vehicle lead to more severe injuries and higher mortality rates.

Badea-Romero and Lenard (2012) found that passenger cars were more commonly involved in cyclist collisions than other vehicles. Out of the 43 head impacts sustained during bicycle-car collisions, the majority of head injuries were due to contact with the ground whereas only 14 were sustained due to contact with the car (Table A.12). The windscreen glazing was the most common impact region accounting for 5 out 14 head injuries. All together the edges of the windscreen, the A-pillars and scuttle panel accounted for fewer head injuries than the windscreen glazing but more than the bonnet. Similar results were reported by Nie and Yang (2014) and Peng *et al.* (2012); the majority of head injuries were sustained due to an impact with the windscreen glazing and these injuries were found to be less severe (AIS 0-2) (Figure A.9). Head injuries sustained from contact with the windscreen frame or A-pillars were less common but more severe (AIS 3+). Mizuno and Kajzer (2000) proposed that there is a relationship between the injury patterns of the head and the stiffness of the contacted parts of the vehicle, with stiffer areas such as the A-pillars and areas near to the windscreen frame causing more severe head injuries.

Table A.12: Distribution of 43 head impacts sustained during cyclist collisions with a passenger car. (Badea-Romero and Lenard, 2012)

Impact location	n	%
Road or ground	29	67.5
Outer edges of bonnet	1	2.3
Base of windscreen	3	7.0
Middle of windscreen	2	4.6
Header rail	3	7.0
Base of A-pillar	2	4.6
A-pillar above base	3	7.0

Otte et al., (2015) found that helmet damage resulted from head impacts on the windscreen, roof edge or road surface. They reported that 88.3% of all head impacts happened on a flat surface while 11.7% happened on an edgy surface. Previously Otte *et al.* (2012) reported that although the majority of head impacts with a car occur on flat surfaces (93%), the edge impacts are much more severe. They recommended that helmet standards should consider using flat and edgy anvils for testing.

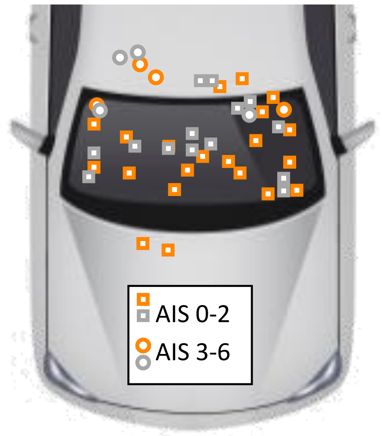

Figure A.9: Distribution and severity of head impacts sustained during cyclist-car collisions by un-helmeted cyclists. Orange - Nie and Yang (2014), grey - Peng et al (2012)

A.7.3 Impact Velocity

The average bicycle moving velocity was found to be 12.0 km/h and the average vehicle impact velocity was found to be 41.0 km/h (Table A.13). The range of velocities stated in Bourdet *et al*. (2014), Nie and Yang (2014) and Peng *et al.* (2012) can be seen in Figure A.10. Nie and Yang (2014) specified that accident data was only used if the impact speed of the vehicle was greater than 20km/h.

Table A.13: Bicycle and vehicle velocities during collisions

Paper	Bicycle moving speed (km/h)				Vehicle impact speed (km/h)			
	Min	Max	Mean	STD.	Min	Max	Mean	STD.
(Bourdet *et al.*, 2014)	5.0	20.0	14.0	5.49	7.5	70.0	31.5	18.20
(Nie and Yang, 2014)	1.8	24.8	10.2	5.48	23.8	77.4	43.5	13.68
(Peng *et al.* 2012)	5.0	25.0	11.8	5.31	32.4	77.4	48.2	12.38
Average	*3.9*	*23.3*	*12.0*	*5.4*	*21.2*	*74.9*	*41.0*	*14.8*

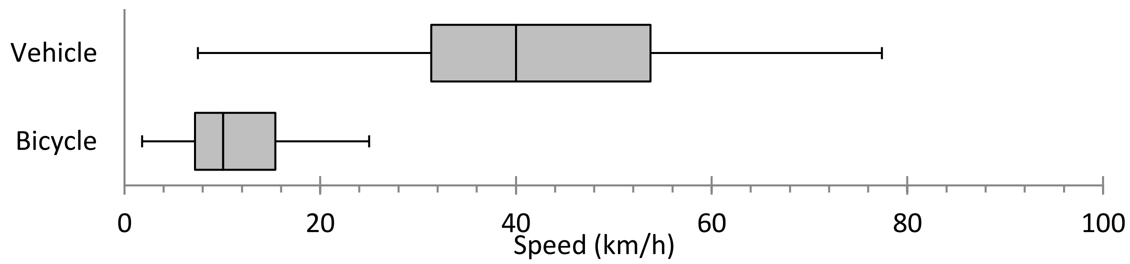

Figure A.10: Vehicle and bicycle speeds during collisions amalgamated from Bourdet et al. (2014), Nie and Yang (2014) and Peng et al. (2012)

Peng *et al.* (2012) found that during collisions involving vehicles the head impact velocity is lower than the vehicle impact velocity. The average relative head impact velocity was 32.5km/h which was consistent with the value of 33.9 km/h stated by Nie and Yang (2014).

Verschueren (2009) analysed two single bicycle collisions, one high speed (55km/h) and one low speed (25km/h). The results showed that the head impact location rather than the impact speed was the decisive parameter in the injury severity. Although both cyclists wore helmets the low speed collision resulted in a fatality. As a result Veschueren recommended a 2-3cm extension of bicycle helmet around the temple area.

Shepers *et al.* (2012) performed a study into single vehicle bicycle crashes and found that cyclists were more likely to lose control due to an uneven road surface or due to abrupt steering manoeuvres when travelling at higher speeds (Table A.14).

Table A.14: Association of single-bicycle crash types with causes and speed prior to crash Adapted from (Shepers et al. (2012)

Speed prior to the collision (km/h)	Crashes		Loss of control due to uneven road surface		loss of control due to abrupt steering manoeuvres	
	N	%	OR	95% CI	OR	95% CI
>25	49	7.6%	3.54	(1.32-9.53)	2.59	(1.13-5.93)
15-25	139	21.5%	2.59	(1.20-5.59)	1.5	(0.80-2.81)
5-15	243	37.6%	1.87	(0.90-3.91)	1.21	(0.72-2.06)
<5	216	33.4%	1		1	

Van Schijndel et al. (2012) performed crash testing using a Polar III ATD mounted on a bicycle and a Volvo V70. Four impact scenarios were tested with the car travelling at a speed of 40km/h and the bicycle travelling at 15km/h (stationary for the rear impact). The head impact speed varied from 9.7 to 16.3 m/s and no head contact was made with the vehicle in the rear impact scenario (Table A.15).

Table A.15: Testing set up and head impact location and speeds for bicycle vs car impacts Adapted from (van Schijndel *et al.*, 2012)

Test	VRU impacted	VRU speed (km/h)	Vehicle speed (km/h)	Head impact location	Head impact speed (m/s)
1	Right side	15	40	Nearside A-Pillar	n/a
2	Right side	15	40	Centre	14.8
3	Right side	15	40	Farside A-Pillar	9.7
4	Rear	0	40	Right A-pillar	16.3

A.7.4 Impact Direction

Peng *et al.* (2012) found that the majority of cyclist collisions involve a vehicle impacting from either 9 o'clock or 3 o'clock directions (Figure A.11).

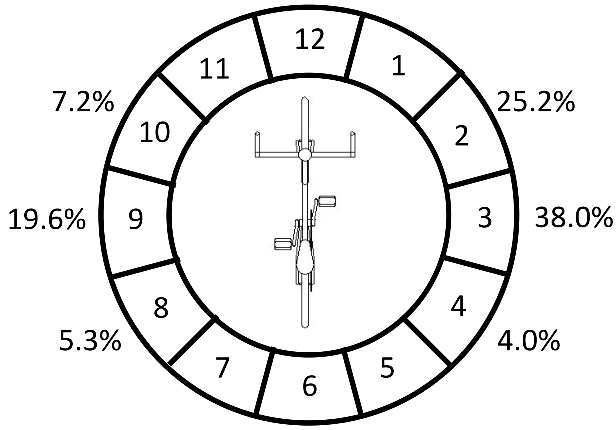

Figure A.11: Impact direction and frequency of cyclists and cars recorded in clock system (Peng *et al.*, 2012).

So although the majority of vehicle car collisions involve impacting the bike from a lateral direction Table A.16) Malczyk *et al.* (2014) found that the distribution of head impacts suggest that the majority of impacts to the head came front, either straight or oblique. Furthermore Bil et al. (2010) reported that the most fatal collision direction is head-on (OR = 1.91; 95% CI 1.51–2.43).

Table A.16: Collision Impact Directions

Paper	Impact Direction							
	Lateral		Head on		From Side		From Behind	
	N	%	N	%	N	%	N	%
(Bíl *et al.*, 2010)	1326	24.4	1104	20.3	1958	36.1	1040	19.2
(ATSB, 2006)	24	19.5	18	14.6	35	28.5	46	37.4
(Bourdet *et al.*, 2014)	0	0.0	5	22.7	19	86.4	0	0.0

A.8 References

Abu-Zidan F, Nagelkerke N and Sudhakar R (2007). Factors affecting severity of bicycle-related injuries: The role of helmets in preventing head injuries. *Emergency Medicine Australasia, 19*, 366–371. doi:10.1111/j.1742-6723.2007.00967.x.

Amoros E, Chiron M, Martin J, Thelot B and Laumon B (2012). Bicycle helmet wearing and the risk of head, face, and neck injury: a French case control study based on a road trauma registry. *Injury Prevention, 18*, 27-32. doi:10.1136/ip.2011.031815.

APROSYS (2008). *Biomechanics of real world injuries and their associated injury criteria*. EC.

ATSB (2006). *Road Safety Report: Deaths of cyclists due to road crashes*. Australian Transport Safety Bureau.

Attewell R, Glase K and McFadden M (2000). *Bicycle helmets and injury prevention: A formal review*. Australian Transport Safety Bureau.

Badea-Romero A and Lenard J (2012). Source of head injury for pedestrians and pedal cyclists: Striking vehicle or road? *Accident Analysis and Prevention*.

Bambach M, Mitchell R, Grzebieta R and Oliver J (2013). The effectiveness of helmets in bicycle collisions with motor vehicles: A case–control study. *Accident Analysis and Prevention, 53*, 78– 88. doi:10.1016/j.aap.2013.01.005.

Benson BW, Hamilton GM, H MW, P M and J D (2009). Is protective equipment useful in preventing concussion? A systematic review of the literature. *Br J Sports Med, 43*, i56–i67. doi:10.1136/bjsm.2009.058271.

BHSI (2015). *Hard Shell Helmets*. Bicycle Helmet Safety Institute.

Bíl M, Bílova M and Müller I (2010). Critical factors in fatal collisions of adult cyclists with automobiles. *Accident Analysis and Prevention, 42*, 1632–1636.

Bogerd C, Shinar D, Chliaoutakis J, Cavallo V, Crundall D, Dias J, Haworth N, Holt N, Houtenbos M, Kuklane K, et al. (2015). *Final report of Working Group 2: Traffic psychology*. COST Action TU1101 / HOPE, Brussels, Belgium.

Bourdet N, Deck C, Serre T, Perrin C, Llari M and Willinger R (2014). In-depth real-world bicycle accident reconstructions. *International Journal of Crashworthiness, 19*(3), 222-232. doi:10.1080/13588265.2013.805293.

Ching R, Thompson D, Thompson R, Thomas D, Chilcott W and Rivara F (1997). Damage to Bicycle Helmets Involved with Crashes. *Accident Analysis and Prevention, 29*(5), 555-562.

Cook A and Sheikh A (2003). Trends in serious head injuries among English cyclists and pedestrians. *Injury Prevention, 9*, 266-267.

Cripton P, Dressler D, Stuart C and Dennison CRD (2014). Bicycle helmets are highly effective at preventing head injury during head impact: Head-form accelerations and injury criteria for helmeted and unhelmeted impacts. *Accident Analysis and Prevention, 70*, 1-7. doi:http://dx.doi.org/10.1016/j.aap.2014.02.016.

Cripton P, Shen H, Brubacher J, Chipman M, Friedman S and Harris M (2015). Severity of urban cycling injuries and the relationship with personal, trip, route and crash characteristics: analyses using four severity metrics. *BMJ Open, 5*. doi:10.1136/bmjopen-2014-006654.

Cummings P, Rivara F, Thompson D and Thompson R (2006). Misconceptions regarding case-control studies of bicycle helmets and head injury. *Accident Analysis and Prevention, 38*, 636–643.

Curnow W (2005). The Cochrane Collaboration and bicycle helmets. *Accident Analysis and Prevention, 37*, 569–573. doi:10.1016/j.aap.2005.01.009.

DeMarco A, Chimich D, Gardiner J and Siegmund G (2016). The impact response of traditional and BMX-style bicycle helmets at different impact severities. *Accident Analysis and Prevention, 92*, 175–183. doi:http://dx.doi.org/10.1016/j.aap.2016.03.027.

Department for Transport (2017). *British social attitudes survey: 2016*. British social attitudes survey: 2016, Table ATT 0305.

Department for Transport (2018). *Walking and cycling statistics, England: 2016*. Walking and cycling statistics, England: 2016, Table WCS CW0302.

Depreitere B, Van Lierde C, Maene S, Plets C, Vander Sloten J, Van Auderkercke R, Van der Perre G and Goffin J (2004). Bicycle-related head injury: a study of 86 cases. *Accident Analysis and Prevention, 36*, 561–567.

Dinh M, Curtis K and Ivers R (2013). The effectiveness of helmets in reducing head injuries and hospital treatment costs: a multicentre study.. *Med J Aust , 198* , 415–17.

Dinh M, Kastelein C, Hopkins R, Royle TBKCD and Ivers R (2015). Mechanisms, injuries and helmet use in cyclists presenting to an inner city emergency department. *Emergency Medicine Australasia, 27*, 323–327. doi:10.1111/1742-6723.12407.

Dorsch M, Woodward A and Somers R (1987). Do bicycle safety helmets reduce severity of head injury in real crashes? *Accid Anal & Prev, 19*(3), 183-190.

Elvik R (2016). Safety-in-numbers: Estimates based on a sample of pedestrian crossings in Norway. *Accident Analysis and Prevention, 91*, 175-182.

European Commission (2013). *Attitudes of Europeans Towards Urban Mobility*. Directorate-General for Mobility and Transport, European Commision.

Fahlstedt M, Halldin P and Kleiven S (2016). The protective effect of a helmet in three bicycle accidents - A finite element study. *Accident Analysis and Prevention, 91*, 135-143. doi:http://dx.doi.org/10.1016/j.aap.2016.02.025.

Fanta O, Boucek J, Hadraba D and Jelen K (2013). Influence of the front part of the vehicle and cyclist's sitting position on the severity of head injury in side collision. *Acta of Bioengineering and Biomechanics, 15*(1). doi:10.5277/abb130113.

Fishman E, Washington S and Haworth N (2013). Bike Share: A Synthesis of the Literature. *Transport Reviews: A Transnational Transdisciplinary Journal, 33*(2), 148-165. doi:10.1080/01441647.2013.775612.

Forman JL, Lopez-Valdes F, Duprey S, Bose D, del Pozo de Dios E, Subit D, Gillispie T, Crandall J and Segui-Gomez M (2015). The tolerance of the human body to automobile collision impact – a systematic review of injury biomechanics research, 1990–2009. *Accident Analysis and Prevention, 80*, 7-17.

Franklin J and Chapman G (2005). *Quantifying the Risk of Head Injury to Child Cyclists in England: An Analysis of Hospital Admissions Data*. Bicycle Helmet Research Foundation.

Gopinath B, Jagnoor J, Craig A, Kifley A, Dinh M, Ivers R, Boufous S and Cameron I (2016). Describing and comparing the characteristics of injured bicyclists and other injured road users: a prospective cohort study. *BMC Public Health, 16*. doi:10.1186/s12889-016-2988-y.

Hagel B and Pless B (2006). A critical examination of arguments against bicycle helmet use and legislation. *Accident Analysis and Prevention, 38*, 277–278.

Hansen K, Engesæter L and Viste A (2003). Protective Effect of Different Types of Bicycle Helmets. *Traffic Injury Prevention, 4*(4), 285-290.

Haworth N, Schramm A, King M and Steinhardt D (2010). *Bicycle Helmet Research*. CARRSQ Monograph 5, CARRSQ: Brisbane, Australia.

Haworth N, Shinar D and Oron-Gilad T (2014). Developing an international survey of bicycle and helmet usage. *International Cycling Safety Conference - 18-19 November 2014, Göteborg, Sweden.*

Hendrie D, Legge M, Rosman D and Kirov C (2008). *An economic evaluation of the mandatory bicycle helmet legislation in Western Australia*. Road Accident Prevention Research Unit, Department of Public Health.

Hewson P (2005). Cycle Helmets and Road Casualties in the UK. *Traffic Injury Prevention, 6*(2), 127 — 134.

Hummel T (2015). *Typical Cyclist-Car Accidents: Compact accident research*. GDV - German Insurers Accident Research, German Insurance Association.

Isaksson-Hellman I and Werneke J (2016). Special Issue Article - Cycling Safety: Detailed description of bicycle and passenger car collisions based on insurance claims. *Safety Science*. doi:http://dx.doi.org/10.1016/j.ssci.2016.02.008.

Jacobson G, Blizzard L and Dwyer T (1998). Bicycle injuries: Road trauma is not the only concern. *Austr J Public Health*, 451-455.

Kopjar B (2000). Population preventable fraction of bicycle related head injuries. *Injury Prevention, 6*, 235–238.

Laios L and Giannatsis J (2010). Ergonomic evaluation and redesign of children bicycles based on anthropometric data. *Applied Ergonomics, 41*, 428–435.

Lajunen T (2016). Barriers and facilitators of bicycle helmet use among children and their parents. *Transportation research part F: traffic psychology and behaviour, 41* , 294-301.

Lee A, Mann N and Takriti R (2000). A hospital led promotion campaign aimed to increase bicycle helmet wearing among children aged 11–15 living in West Berkshire 1992–98. *Injury Prevention, 6*, 151–153.

Lee R, Hagel B, Karkhaneh M and Rowe B (2009). A systematic review of correct bicycle helmet use: how varying definitions and study quality influence the results. *Injury Prevention, 15*, 125-131. doi:10.1136/ip.2008.019695.

Macpherson A and Spinks A (2008). Bicycle helmet legislation for the uptake of helmet use and prevention of head injuries (Review). *Cochrane Database of Systematic Reviews*(3). doi:10.1002/14651858.CD005401.pub3..

Maimaris C, Summer C, Browning C and Palmer C (1994). Injury patterns in cyclists attending an accident and emergency department: a comparison of helmet wearers and non wearers. *BMJ, 308*(1537). doi:http://dx.doi.org/10.1136/bmj.308.6943.1537.

Malczyk A, Bauer K, Juhra C and Schick S (2014). *Head Injuries in Bicyclists and Associated Crash Characteristics*. GDV - German Insurers Accident Research.

McDermott F, Lane J, Brazenor G and Debney E (1993). The effectiveness of bicycle helmets: a study of 1710 casualties. *Journal of Trauma, 34*, 834-844.

McGarry T and Sheldon R (2008). *Road Safety Research Report No. 84: Cycle Helmet Wearing in 2006*. Department for Transport: London.

McGuire L and Smith N (2000). Cycling safety: injury prevention in Oxford cyclists. *Injury Prevention, 6*, 285–287.

McIntosh A and Dowdell B (1992). A Field and Laboratory Study of the Performance of Pedal Cycle Helmets in Real Accidents. *1992 International IRCOBI Conference on the Biomechanics of Impacts*.

McIntosh A, Doedell B and Svensson N (1998). Pedal Cycle Helmet Effectiveness: A Field Study of Pedal Cycle Accidents. *Accident Prevention and Analysis, 30*(2), 161-168.

McIntosh A, Lai A and Schilter E (2013). Bicycle helmets: head impact dynamics in helmeted and unhelmeted oblique impact tests. *Traffic Inj Prev, 14*, 501–08.

Mizuno K and Kazjer J (2000). Head Injuries in Vehicle-Pedestrian Impact. *Biomechanics Research*.

Nie J and Yang J (2014). A study of bicyclist kinematics and injuries based on reconstruction of passenger car–bicycle accident in China. *Accident Analysis and Prevention, 71*, 50-59.

Oikawa S, Matsui Y, Wakabayashi A, Gomei S, Nakadate H and Aomura S (2016). Severity of cyclist head injuries caused by impacts with vehicle structure and road surface. *Journal of Biomechanical Science and Engineering*. doi:10.1299/jbse.15-00613.

Olofsson E, Bunketorp O and Andersson A (2015). Special Issue Article - Cycling safety: Helmet use and injuries in children's bicycle crashes in the Gothenburg Region. *Safety Science*. doi:http://dx.doi.org/10.1016/j.ssci.2015.11.024.

Orsi C, Ferraro O, Montomoli C, Otte D and Morandi A (2014). Alcohol consumption, helmet use and head trauma in cycling collisions in Germany. *Accident Analysis and Prevention, 65*, 97-104. doi:http://dx.doi.org/10.1016/j.aap.2013.12.019.

Otte D, Jänsch M and Haasper C (2012). Injury protection and accident causation parameters for vulnerable road users based on German In-Depth Accident Study GIDAS. *Accident Analysis and Prevention, 44*, 149-153.

Otte D, Jänsch M, Morandi A, Orsi C, Stendardo A, Bogerd C, Tzamalouca G, Papadakaki M, Chliaoutakis J, Parkkari K, Dias J and Weber T (2015). Final report of Working Group 1: In-depth accident observations and injury statistics. *COST Action TU1101 / HOPE*, Brussels, Belgium.

Peng Y, Chen Y, Yang J, Otte D and Willinger R (2012). A study of pedestrian and bicyclist exposure to head injury in passenger car collisions based on accident data and simulations. *Safety Science, 50*, 1749-1759. doi:http://dx.doi.org/10.1016/j.ssci.2012.03.005.

Post A (2013). *The influence of dynamic response characteristics on traumatic brain injury*. University of Ottawa, Ottawa, Canada.

Ritter N and Vance C (2011). The determinants of bicycle helmet use: Evidence from Germany. *Accident Analysis and Prevention, 43*(1), 95-100.

Schepers P and Wolt K (2012). Single-bicycle crash types and characteristics. *Cycling Research International, 2*.

Scholten A, Polinder S, Panneman M, van Beeck E and Haagsma J (2015). Incidence and costs of bicycle-related traumatic brain injuries in the Netherlands. *Accident Analysis and Prevention, 81*, 51-60. doi:http://dx.doi.org/10.1016/j.aap.2015.04.022.

SNELL (1996). *Circumstances and Severity of Bicycle Injuries: Summary Report of Harborview Helmet Studies*. Snell Memorial Foundation.

Stier R, Otte D, Müller C, Petri M, Gaulke R, Krettek C and Brand S (2016). Effectiveness of Bicycle Safety Helmets in Preventing Facial Injuries in Road Accidents. *Arch Trauma Res*. doi:10.5812/atr.30011..

Talbot R, Reed S, Barnes J, Thomas P and Christie N (2014). *Pedal cyclist fatalities in London: analysis of police collision files (2007-2011).* University College London and Loughborough University on behalf of Transport for London.

Thomas S, Acton C, Nixon JBD, Pitt W and Clark R (1994). Effectiveness of bicycle helmets in preventing head injury in children: case-control study. *BMJ, 308*, 173-176.

Thompson R, Rivara F and Thompson D (1989). A case-control study of the effectiveness of bicycle safety helmets. *N Engl J Med , 320*, 1361–67.

Thompson R, Rivara F and Thompson D (1996). Effectiveness of bicycle safety helmets in preventing head injuries. A case-control study. *JAMA, 276*(24), 1968-73.

Thompson D, Rivara F and Thompson R (2006). Helmets for preventing head and facial injuries in bicyclists (Review). *Cochrane Database of Systematic Reviews*(4). doi:10.1002/14651858.CD001855.

Towner E, Dowswell T, Burkes M, Dickinson H, Towner J and Hayes M (2002). *Bicycle Helmets – A Review Of Their Effectiveness: A Critical Review Of The Literature*. Department for Transport.

van Schijndel M, de Hair S, Rodarius C and Fredriksson R (2012). Cyclist kinematics in car impacts reconstructed in simulations and full scale testing with Polar dummy. *IRCOBI Conference*.

Verschueren P (2009). *Biomechanical Analysis of Head Injuries Related to Bicycle Accidents and a New Bicycle Helmet Concept*. Katholieke Universiteit Leuven, Belgium.

Appendix B Comparison of Helmet Testing Standards

B.1 Introduction

This Appendix summarises the key points of the major cycle helmet testing standards from around the world. An overview of the main legal requirements in force for each country is documented. Typically, these reference one or more existing standards as demonstrating suitable performance for cycle helmets and a review of these standards is performed within this section.

The review of these standards provides context for establishing the key features of the NCHAP protocols, whilst also identifying the key differences between current international cycle helmet testing standards.

B.2 Cycle Helmet Standard Summaries

Most cycle helmet regulations, including those in force in the UK, require helmets to meet the requirements of an existing standard; that is, the regulation does not itself define the requirements that the helmet must meet. As a result, the main standards referred to in regulations were reviewed.

The following sections give an overview of seven common cycle helmet standards currently in use around the world. An additional six standards were identified: four of which have been superseded by CEN EN1078 (BS 6863 of Britain, DIN 33954 of Germany, KOV 1985:6 of Sweden and BFU R 8602 of Switzerland); the ANSI Z90.4 standard of the United States, which was superseded by ASTM F1447; and the Snell B-90A/B-90C standards, which were superseded by Snell B-95 (1998 Addendum).

While each standard has its own specific requirements and methods, the majority of test programmes generally involve the following three tests: impact tests, a retention system strength test and a retention system stability test. The following sections summarise the requirements, similarities and differences between each of the reviewed standards.

This section gives an overview of the following standards and regulatory requirements:

- AS/NZS 2063:2008;
- ASTM F1447:2006;
- CPSC 16 CFR 1203:1998;
- EN 1078:2012;
- EN 1080:2013;
- JIS T 8134:2007; and
- Snell B-95 (including 1998 “B-95C” addendum).

B.2.1 AS/NZS 2063:2008

The mandatory cycle helmet testing standard of the joint Standards Australia/Standards New Zealand Committee (AS/NZS 2063) was first introduced in 1996 and was last updated in

2008. This standard is further supported by the AS/NZS 2512 series of standards that provide further detail on the methods for testing protective helmets.

Impact Tests: The AS/NZS 2063 standard uses a flat anvil helmet drop test from a height of 1.5 m (approximately equivalent to an impact velocity of 5.4 m/s). The peak headform acceleration in this test is not allowed to exceed 250 *g.* Furthermore, the AS/NZS standard also requires the cumulative duration of the headform acceleration not to exceed 200 *g* over a period of 3 ms and 150 *g* over a period of 6 ms. Finally, the standard also includes a unique load distribution test in which the helmet is dropped from 1.0 m and where the helmet must not create a force greater than 500 N over a 100 mm^2 circular area. Although the impact energy of this test is relatively low, these additional requirements mean that the standard is well regarded.

Retention System Strength Tests: The retention system strength test is performed with an inertial hammer suspended from the straps of the helmet. The mass of the hammer is 10 kg and the fall length is 250 mm, which results in an energy of approximately 24 J. The straps must not elongate more than 30 mm.

Retention System Stability Tests: The retention system stability test applies a quasi-static load of 50 N to the helmet to roll the helmet off a headform. The quasi-static force is applied tangentially to the helmet for a period lasting between 15 and 30 seconds. This test will be repeated for a rearward force applied to the front of the helmet and a forward force applied to the rear of the helmet. On completion of testing, the helmet must not expose a test band bounded by the basic plane, a parallel line located above the basic plane (74 mm for size A headforms and 85 mm for size J headforms) and two perpendicular lines located 25 mm either side of the mid-sagittal plane.

B.2.2 ASTM F1447: 2012

This voluntary bicycle helmet standard was produced by the American Society for Testing and Materials (ASTM). While it is technically still in use, it has since been superseded by the mandatory (in the US) CPSC standard. ASTM F1447 was last updated in 2012 and is further supported by ASTM F1446 which describes in further detail standardised equipment and procedures used for evaluating the performance characteristics of protective headgear.

Impact Tests: The ASTM F1447 standard involves a drop test onto three different anvils: flat, hemispherical and kerbstone. These drops are performed at velocities of 6.2 m/s on the flat anvil and 4.8 m/s on the hemispherical and kerbstone anvils (approximately equivalent to helmet drop heights of 2.0 m and 1.2 m, respectively). The peak headform accelerations during testing must not register more than 300 g.

Retention System Strength Tests: The retention system strength test is performed with an inertial hammer suspended from the straps of the helmet. The mass of the hammer is 4 kg and the fall length is 600 mm, which results in an energy of approximately 24 J. The straps must not elongate more than 30 mm.

Retention System Stability Tests: The retention system stability test is performed by using a 4 kg inertial hammer to deliver an impact load to the helmet in a manner to roll the helmet off a headform. The inertial hammer is attached to the opposite edge of the helmet on a headform inclined at 45° to the vertical and the testing is performed for both upward and downward facing headforms. The helmet is permitted to move on the headform; however, the helmet should not come off or move excessively (defined as exposure of those parts of the coronal plane previously covered by the helmet).

B.2.3 CAN/CSA-D113.2-M89

The cycle helmet standard of the Canadian Standards Association (CSA) was originally introduced in 1989, last updated technically updated in 1996, and reaffirmed in 2004.

Impact Tests: The CSA-D113.2-M86 drop test is performed onto two types of anvil: flat and cylindrical. Two flat anvil drop tests, for adult and older child helmets, are performed at two separate helmet impact locations and using impact velocities of 5.7 m/s and 4.7 m/s (comparable to impact energies of 80 J and 55 J, respectively). Two cylindrical anvil helmet drop tests are then both performed at 4.7 m/s impact velocities at two separate helmet impact locations. The same velocities are used with a smaller headform for helmets for children five years old and under, giving equivalent impact energies of up to 67 J and 45 J.

For child and adult helmets the maximum headform accelerations are 250 *g* for the 80 J flat anvil test, 200 *g* for the 55 J flat anvil test, and 250 *g* for the 55 J cylindrical anvil test. For younger child helmets the limits are 200 *g* for the flat anvil tests and 150 *g* for the cylindrical anvil. The standard also recommends that manufacturers ensure that the Gadd Severity Index is less than 1500 for all tests.

Retention System Strength Tests: The retention system strength test is performed by dropping a 2 kg weight attached to the helmet from a height such that 20 J of energy is imparted to the retention system. The retention system must not release, dynamic elongation must not exceed 25 mm and post-test static elongation must not exceed 12 mm.

Retention System Stability Tests: The retention system stability test is performed by subjecting the helmet to a 250 N tangential force for 5 seconds, if the helmet moves more than 10 mm during this time then the force is continued for another 5 seconds. Helmet rotation must not exceed 45°.

B.2.4 CPSC 16 CFR 1203:1998

This bicycle helmet standard was produced by the Consumer Product Safety Commission (CPSC). The standard was developed in conjunction with ASTM and the test procedures are largely similar to the ATSM F1447 standard. It was introduced in 1998 and was made compulsory in 1999. The standard is part of the US Code of Federal Regulations and as such is a legal requirement in all US States.

Impact Tests: The drop test uses three different anvils: flat, hemispherical and kerbstone. These drops are performed at velocities of 6.2 m/s on the flat anvil and 4.8 m/s on the hemispherical and kerbstone anvils. These velocities are approximately equivalent to drop heights of 2.0 m and 1.2 m respectively. The instrumented headform must not register more than 300 *g* acceleration throughout each test.

Retention System Strength Tests: The retention system strength test is performed with an inertial hammer suspended from the straps. The hammer mass is 4 kg and the fall length is

600 mm, corresponding to a fall energy of about 24 J. The straps must not elongate more than 30 mm.

Retention System Stability Tests: The retention system stability test is performed by attaching an inertial hammer of 4 kg mass to the opposite edge of the helmeted headform inclined at 45° and dropping the mass. The helmet is permitted to move on the headform, but it must not come off, whilst the testing is performed for both upward and downward facing headforms.

The CPSC standard requires helmets for children under the age of five to cover a larger proportion of the head than helmets for older children and adults.

The CPSC standard also requires manufacturers to implement a 'reasonable testing programme' to ensure that products meet the certification requirements. This testing may be conducted by a third party, but the manufacturers and importers are responsible for ensuring that samples from each production lot are compliant with the standard or a reasonable testing programme. CPSC will test for compliance to the standard. There are also specific requirements for record-keeping for helmet tests.

B.2.5 EN 1078:2012 and EN 1080:2013

EN 1078 and 1080 are the cycle helmet standards produced by the European Committee for Standardisation (CEN). EN 1078 applies to cycle helmets for both children and adults, whilst EN 1080 applies specifically to helmets for young children (young children indirectly defined only by head size). These were first introduced by all CEN member states[2] in 1997 and were last updated in 2012 and 2013 for EN 1078 and EN 1080, respectively.

Impact Tests: The impact test requirements are identical for both standards and involve two anvils: flat and kerbstone. These tests are performed at velocities of 5.42 m/s and 4.57 m/s respectively, which correspond to drop heights of 1.5 m and 1.06 m. The instrumented headform must not register more than 250 *g* acceleration throughout each test.

Retention System Strength Tests: EN 1078 tests the strength of the retention system with an inertial hammer suspended from the straps. The hammer mass is 4 kg and the fall length is 600 mm, which results in a fall energy of about 24 J. The straps must not elongate more than 35 mm dynamically and the residual extension must not exceed 25 mm. It must be possible to operate the fastening system with one hand while under load.

In contrast, EN 1080 requires that the fastening system should self-release when a force of greater than 90 N but less than 160 N is applied quasi-statically (100 mm/min). This is designed to prevent strangulation by ensuring that the strap will release if the helmet becomes trapped, for instance in playground equipment.

[2] CEN members are the national standards bodies of Austria, Belgium, Bulgaria, Croatia, Cyprus, Czech Republic, Denmark, Estonia, Finland, France, Germany, Greece, Hungary, Iceland, Ireland, Italy, Latvia, Lithuania, Luxembourg, Malta, Netherlands, Norway, Poland, Portugal, Romania, Slovakia, Slovenia, Spain, Sweden, Switzerland, Turkey and United Kingdom.

Retention System Stability Tests: EN 1078 tests the stability of the helmet and retention system by attaching an inertial hammer of 10 kg mass and 175 mm drop height to the opposite edge of the helmet. The helmet is permitted to move on the headform, but it should not come off the headform. EN 1080 does not define a stability test.

EN 1078 and EN 1080 contain no conformity of production requirements.

B.2.6 JIS T 8134:2007

This bicycle helmet safety standard was produced by the Japanese Industrial Standards (JIS) Committee Standards Board. It was first introduced in 1982, updated in 1995 and was last technically revised in 2007.

Impact Tests: The drop test uses two different anvils: flat and hemispherical. These drops are performed at velocities of 5.42 m/s on the flat anvil and 4.57 m/s on the hemispherical. These impact velocities are approximately equivalent to drop heights of 1.50 m and 1.06 m respectively. The instrumented headform must not register more than 300 *g* acceleration throughout each test. Furthermore, this standard also requires the cumulative duration of the headform acceleration not to exceed 150 *g* over a period of 4 ms.

Retention System Strength Tests: The retention system strength test is performed with an inertial hammer suspended from the straps. The hammer mass is 4 kg and the fall length is 600 mm, corresponding to a fall energy of about 24 J. The straps must not elongate more than 35 mm and the fastening device should easily detach after the test.

Retention System Stability Tests: The retention system stability test is performed by attaching an inertial hammer of 10 kg mass and 175 mm drop height to the centre of the back of the helmet. The helmet is permitted to move on the headform, but must not come off during testing.

The CPSC standard requires helmets for children under the age of five to cover a larger proportion of the head than helmets for older children and adults.

B.2.7 Snell B-95 (1998 addendum)

Snell B-95 cycle helmet standard is produced by the Snell Memorial Foundation and replaced the Snell B-90 cycle helmet standard. The B-95 standard (introduced in 1995, and updated in 1998) is generally considered to be the most stringent performance standard in the cycle helmet testing industry.

Impact Tests: Snell B-95 perform drop tests involving three different anvils; flat, kerbstone and hemispherical. These drops are performed at impact energies of 110 J, 72 J and 72 J respectively, which are approximately equivalent to 6.6 m/s and 5.3 m/s impact velocities. The instrumented headform must not register more than 300 *g* acceleration across all tests.

Retention System Strength Tests: The retention system strength test is performed with an inertial hammer suspended from the straps. The hammer mass is 4 kg and the fall length is 600 mm, which results in a fall energy of about 24 J. The straps must not elongate more than 30 mm during testing and must support all mechanical loading.

Retention System Stability Tests: The retention system stability test is performed by attaching an inertial hammer of 4 kg mass to the opposite edge of the helmeted headform inclined at 45° and dropping the mass. The helmet is permitted to move on the headform, but it must not come off, whilst the testing is performed for both upward and downward facing headforms.

Snell periodically tests helmets intended for the consumer (e.g. bought from a retailer) to ensure on-going compliance with the standard.

Snell B-95 contains an addendum (Snell B-95C) that updates the standard to incorporate requirements for helmets intended for use by children aged from one up to five years old. Primarily, this updates the requirements for coverage and field of view to match CPSC requirements.

B.3 Comparison of Cycle Helmet Impact Test Characteristics

The following sections provide a comparison of the various components that form the cycle helmet impact tests performed for each standard. These key issues are, at first, compared between standards, before highlighting any potential issues and solutions that may exist to progress the development of the NCHAP protocols.

B.3.1 Headform Characteristics

The current headforms used by all cycle helmet testing standards are characterised by a range of non-deformable “head-shaped” masses. The geometry and masses of this range of headforms are taken from specific test headform standards (AS/NZS 2063: AS/NZS 2512.1; EN 1078/EN 1080: EN 960; ASTM F1447: ASTM F2220; CAN/CSA-D113.2-M89, CPSC 16 CFR Part 1203, JIS T 8134 & Snell B-95: ISO/DIS 6220). All four test headform standards used by the helmet testing standards specify test headforms that represent head geometries and masses that range from younger children to adults with larger heads. The characteristics of the test headforms adopted by the standards reviewed by this Appendix are compared in Table B.1 overleaf.

The headform masses specified by these standards include, in all instances, the combined mass of the entire assembly that is subjected to the impact, minus the mass of the helmet. Furthermore, the designs of all headform assemblies are controlled to account for the mass effects of the neck. From the above, however, it is clear that there are two key philosophies associated with specifying the mass of the head. The most common approach is adopted by AS/NZS 2063, ASTM F1447, CAN/CSA-D113.2M89, EN 1078, EN 1080 and JIS T 8134, where the mass of the headform is varied with the head circumference to represent the relevant combined head and neck mass at the simulated age. CPSC 16 CFR Part 1203 and Snell B-95/B-95C, however, specify a single headform mass (5 kg) regardless of the circumference of the headform. The rationale behind this decision was because they found insufficient evidence to understand the effects that changing the headform mass or peak acceleration limits would have on injury outcomes.

Table B.1: Headform geometric and mass requirements

Headform Size	Child	A	C	E	G	J	K	M	O
Headform Circumference /mm									
AS/NZS 2063		500		540		570		600	620
ASTM F1447		500	520	540		570		600	620
CAN/CSA-D113.2-M89		500		540		570		600	620
CPSC 16 CFR Part 1203		500		540		570		600	620
EN 1078/EN 1080	455	495		535		575		605	625
JIS T 8134		500		540		570		600	620
Snell B-95/ Snell B-95C		500		540		570		600	620
Headform Mass /kg									
AS/NZS 2063		3.1		4.1		4.7		5.6	6.1
ASTM F1447		3.1	3.6	4.1		4.7		5.6	6.1
CAN/CSA-D113.2-M89		3.1		4.1		5		5	
CPSC 16 CFR Part 1203		5		5		5		5	5
EN 1078/EN 1080	1.97	3.1		4.1		4.7		5.6	6.1
JIS T 8134		3.1		4.1		4.7		5.6	6.1
Snell B-95/ Snell B-95C		5		5		5		5	5

This approach is, however, biomechanically flawed. If optimising helmet designs for passing current test standards based upon a 5 kg headform, the stiffness of the energy attenuating inner liner of the helmet will be optimised for the headform mass and the impact scenario simulated. For children, where the actual mass of the head is significantly lower than the 5 kg proposed by CPSC 16 CFR Part 1203 and Snell B-95/B-95C (i.e. ~3-4 kg), this means that the stiffness of the inner liner of the helmet will be much stiffer than required for impact scenarios representing that simulated by these standards. This could predispose children in the US to an increased severity of injuries at lower impact energies (i.e. falling over).

It is important to note that the circumferences of the test headforms specified in EN 960 differ slightly from those specified by the other standards.

Defining the inertial characteristics of the headform is also essential to ensuring that the kinematics of the testing represents that experienced in real-world accidents. Despite this, there are critical differences between the standards for defining the centre of gravity (CoG) of the headform. When comparing the European standards to the rest of the World, it is important to note that the specified headforms differ appreciably (this is discussed further in Appendix B.3.2). Whilst ASTM F1447, CAN/CSA-D113.2-M89, CPSC 16 CFR Part 1203 and AS/NZS 2063 specify the use of half headforms for impact tests and full headforms for the retention system tests, EN 1078 and EN 1080 specify the use of full headforms for all tests. JIS T 8134, however, specifies both methods. Thus EN 1078, EN 1080 and JIS T 8134 specify the headform CoG within a defined spherical boundary, whilst the other standards specify the CoG within a cone formed by a surface angled at 10° to vertical and its vertex located at the external surface of the headform at the point of impact. No standard attempted to specify the requirements for the moment of inertia of the headform.

Finally, all standards were found to use non-deformable metallic headforms to simulate the biomechanics of the head. Whilst appropriate for providing a repeatable headform for the comparison of inter-helmet performance, it is widely recognised that human heads deform during linear impacts (through deformation of both the skull and scalp) and that the scalp also provides a frictional surface that resists the sliding motion of the helmet on the head. The headforms used by these standards therefore results in a non-biofidelic response to head impacts.

Issues: Headform geometry, mass, inertial and mechanical properties should be made more biofidelic

Potential Solutions: Hybrid III headform replicate these essential properties better and also represents friction between scalp and helmet. Specify moment of inertia requirements for EN 960 headform to be within biofidelic range.

Potential Protocol Change: Use Hybrid III headform range as test headform or specify EN 960 headform moment of inertia.

B.3.2 Drop Test Assembly and Neck Anchorage Characteristics

There are significant differences between the drop test assemblies specified by European standards and those specified by other standards across the rest of the World (Table B.2). European standards specify that helmets are strapped to full headforms that are supported in the correct orientation by a carriage, before being dropped in a guided free fall onto an anvil in an unconstrained impact. All other standards require the helmet to be strapped to a half headform rigidly supported by an arm. This both holds the head securely in position during the guided free fall and constrains the motion of the test headform during impact. All standards do, however, allow drop tests to be guided via either wires or rails.

Table B.2: Cycle helmet drop assembly specifications

Standard	Drop Assembly Type		Guidance Type	
	Support Arm	Free-Fall Carriage	Guide Wire	Guide Rail
AS/NZS 2063	✓		✓	✓
ASTM F1447	✓		✓	✓
CAN/CSA-D113.2-M89	✓		✓	✓
CPSC 16 CFR Part 1203	✓		✓	✓
EN 1078/EN 1080		✓	✓	✓
JIS T 8134	✓	✓	✓	
Snell B-95/ Snell B-95C	✓		✓	✓

These differences in drop test assembly configurations introduce fundamental differences between the kinematics underpinning European testing standards and those underpinning all other standards. Furthermore, no single standard specified any requirements to anchor the headform to a torso via a representative and validated neck. It is therefore difficult to comment upon which method is most appropriate method for adopting for these protocols, as there has been no previous research performed to establish which approach is the most representative.

Issues: Headform and neck drop test assembly configuration should be more biofidelic

Potential Solutions: Hybrid III headform and neck attached to a representative torso could replicate these essential properties better. Adopt AS/NZS 2063, ASTM F1447, CAN/CSA-D113.2-M89, CPSC 16 CFR Part 1203, JIS T 8134 and Snell B-95/B-95C approach and attach Hybrid III headform and neck to support arm. Develop pendulum approach with pendulum body representing inertial properties of body. Develop moving anvil approach with headform angled to replicate angle of impact with floor moving at speed of impact.

Potential Protocol Change: Maintain the status quo in the short term and use new drop test assembly with Hybrid III headform and modified neck attached in the longer term.

B.3.3 Anvils

When comparing helmet test standards, although there are small differences in the design of the anvils, the most important differences are found in the specification of which anvils are used for testing helmet impact performance (Table B.3). Whilst all eight standards specify the use of flat anvils, only five use kerbstone anvils, four use hemispherical anvils and one uses a cylindrical anvil. No standard was found to specify the use of an angled anvil. Further anvil design variations include differences in the specifications for the diameters of the flat anvils, the lengths of the kerbstone anvils and between the diameters of the cylindrical and hemispherical anvils.

Table B.3: Impact anvil specifications

Standard	Flat	Kerbstone	Hemispherical/ Cylindrical
AS/NZS 2063	Flat circular anvil with diameter of ≥125 mm		
ASTM F1447	Flat circular anvil with diameter of ≥125 mm and ≥24 mm thick	Two flat faces angled at 105±5°, a striking edge of radius 15±0.5 mm, height of ≥50 mm and length of ≥200 mm	Hemispherical anvil with radius of 48±1 mm
CAN/CSA-D113.2-M89	Flat circular anvil with diameter of ≥150 mm		Cylindrical anvil with radius of 40±1 mm and length of 200±1 mm
CPSC 16 CFR Part 1203	Flat circular anvil with diameter of ≥125 mm and ≥24 mm thick	Two flat faces angled at 105°, a striking edge of radius 15±0.5 mm, height of ≥50 mm and length of ≥200 mm	Hemispherical anvil with radius of 48±1 mm
EN 1078	Flat circular anvil with diameter of 130±3 mm	Two flat faces angled 52.5±2.5° to vertical, a striking edge of radius 15±0.5 mm, height of ≥50 mm and length of ≥125 mm	

Standard	Flat	Kerbstone	Hemispherical/ Cylindrical
EN 1080	Flat circular anvil with diameter of 130±3 mm	Two flat faces angled 52.5±2.5° to vertical, a striking edge of radius 15±0.5 mm, height of ≥50 mm and length of 130 (-0/+3) mm	
JIS T 8134	Flat circular anvil with diameter of 130±3 mm		Hemispherical anvil with radius of 50±2 mm
Snell B-95/ Snell B-95C	Flat anvil with surface area of ≥0.0127 mm (i.e. diameter of ≥127 mm)	Two flat faces at a dihedral angle of 105° (each angled 52.5±2.5° to vertical), a striking edge of radius 15±0.5 mm, height of ≥50 mm and length of ≥200 mm	Hemispherical anvil with radius of 48±0.5 mm

Whilst testing with flat anvils headforms to represent linear impact situations is common, these only really represent instances where the cyclist falls over whilst either at a standstill or when cycling slowly. The purpose of using kerbstone, hemispherical and cylindrical anvils is to represent collisions with impact partners that, through a smaller radius of curvature, load the helmet at much greater pressures. The benefit of these impact tests are, however, unknown, although it is likely that these mean that a much stiffer shell is used in the design of cycle helmets. There would be a benefit to performing an accident analysis to evaluate the proportion of collisions that cyclists have with small radius of curvature impact partners and determining whether there is any benefit to continuing with this form of impact testing.

Finally, no standard was found to use angled anvils to impart oblique impact forces on the helmeted headform. As it is widely recognised that cyclist collisions often occur at speed, the lack of a tangential component to the head impact force has been of concern for many years. It is clear that further research has to be performed to establish the anvil design that best represents the kinematics of the head during cyclist collisions.

Issues: Anvils used for testing should represent real-world impact partners

Potential Solutions: Harmonised flat anvil design, introduction of an angled anvil, accident analysis to establish impact partners and design of representative anvils for all accidents where incidence of impact partner is >5 %.

Potential Protocol Change: Harmonised flat anvil design and introduction of angled anvil procedure

B.3.4 Cycle Helmet Coverage and Impact Locations

Current standards determine minimum cycle helmet coverage by providing a definition of the test area for impacting. All standards define a test line, above which the impact tests are performed. If the cycle helmet fails to cover the area bounded by this line, all helmets immediately fail the standard. All standards, however, provide a slightly different definition of this test area (Table B.4), with the majority of standards describing a stepped test line that is slightly higher at the front, whilst EN 1078 and EN 1080 specify a test line inclined at 10° to the horizontal. CAN/CSA-D113.2-M89, CPSC 16 CFR Part 1203, JIS T 8134 and Snell B-

95C also define more stringent coverage criteria for children by requiring proportionally greater protection for the lateral aspects of the head, when compared to adult helmets.

Table B.4: Minimum cycle helmet coverage requirements (*bold italics* represent most stringent coverage requirements)

Standard	Reference Line Type	Front* /mm	Lateral* /mm	Rear* /mm
Adult helmets †				
AS/NZS 2063	2 step	66.0	41.0	36.0
ASTM F1447	1 step	68.5	68.5	52.5
CAN/CSA-D113.2-M89	1 step	***52.5***	***32.5***	32.5
CPSC 16 CFR Part 1203	1 step	68.5	68.5	54.5
EN 1078	10° slope	55.0¶	41.25¶	***27.5***
JIS T 8134	1 step	68.5	68.5	54.5
Snell B-95Δ	1 step	53.0	33.0	33.0
Young Child Helmets ‡				
AS/NZS 2063	2 step	***41.6***	34.5	26.5
ASTM F1447	1 step	62.0	62.0	47.0
CAN/CSA-D113.2-M89	2 step	49.0	***29.0***	***4.0***
CPSC 16 CFR Part 1203	2 step	54.0	36.7	11.0
EN 1080	10° slope	50.0¶	36.75¶	23.5
JIS T 8134	2 step	62	36.7	26
Snell B-95CΔ	2 step	39.0	21.0	-6.0

* Positive helmet coverage dimensions are taken vertically upward from the basic plane

† Size J adult helmet

‡ Size A child helmet

¶ Estimated from 10° slope and known rear helmet coverage requirements

Δ Snell B-95/B-95C test line is 15 mm inside of helmet coverage requirements

Due to the wide range of different headform sizes and shapes and the various coverage strategies used by these standards, it would be a complex task to calculate the exact areas defining the extent of protection offered by each standard. From the information given for adult (size J) helmets, however, CAN/CSA-D113.2-M89 provides the greatest coverage for the frontal and lateral aspects of the adult head, whilst EN 1078 provides the best coverage for the rear of the adult head (Table B.4). The most stringent coverage requirements for child helmets are provided by the Snell B-95C standards across all aspects of the helmet (Table B.4). When taking into consideration the 15 mm impact test line offset, however, it could be argued that the most stringent standard for child helmets is AS/NZS 2063 for the front of the head and CAN/CSA-D113.2-M89 for the lateral and rear aspects of the head. It is important to note that the greater the impact test area defined by a standard, the more likely a compliant helmet is to provide a greater extent of protection to the head.

Issues: Cycle helmet coverage, test area and impact points must be more representative of real-world incidents

Potential Solutions: Link in to in-depth collision analysis data to determine where helmet should provide protection

Potential Protocol Change: Determine any change to test area and also measure, score and weight total helmet coverage relative to other key inputs

B.3.5 Cycle Helmet Conditioning Environment

Current cycle helmet test standards specify a range of impact tests to be performed within specific environmental conditions. Across all testing standards there are five approaches to helmet pre-conditioning, with each standard specifying the control of the environmental conditions for testing with a combination of ambient temperature, high temperature, low temperature, water immersion and artificial ageing pre-conditioning techniques (Table B.5). Whilst all standards specified the performance of helmet impact tests at both high and low temperatures, EN 1708, EN 1080 and Snell B-95 do not specify ambient test conditions and EN 1708 and EN 1080 do not require testing after immersion of the helmet in water. EN 1708 and EN 1080 do, however, specify more stringent requirements for the artificial aging of the helmet, requiring the helmet to be conditioned by UV irradiation for 48 hours and sprayed with water at a rate of 1 L/min for 4-6 hours.

Table B.5: Cycle helmet conditioning environment requirements

Standard	Ambient Temperature		High Temperature		Low Temperature		Water Immersion		Artificial Ageing
	°C	hours	°C	hours	°C	hours	°C	hours	
AS/NZS 2063	21.5±3.5	4-30	50±2	4-30	-10±2	4-30	21.5±3.5	4-30	
ASTM F1447	20±3	4-24*†	50±3	4-24*	-15±2	4-24*	19±4	4-24*^	
CAN/CSA-D113.2-M89	20±5	≥4‡	50±2	≥4	-10±2	≥4	22.5±4.5	≥4	
CPSC 16 CFR Part 1203	22±5	≥4*Δ	50±3	4-24*	-15±2	4-24*	22±5	4-24*^	
EN 1078/EN 1080			50±2	4-6	-20±2	4-6			✓¶
JIS T 8134	23±5	≥4	50±2	4-24	-10±2	4-24	25±5	4-24	
Snell B-95/ Snell B-95C			50±2	4-24*Δ	-20±2	4-24*	22±5	4-24*^	

* Barometric pressure of conditioning environment between 75-110 kPa

† Humidity of conditioning environment between 25-75 %

‡ Humidity of conditioning environment at 60±5 %

Δ Humidity of conditioning environment between 20-80 %

^ Fully immersed "crown down" in potable water to a crown depth of 305±25 mm

¶ Helmet exposed successively to UV irradiation by a 150 W xenon-filled quartz lamp for 48 hours at a range of 250 mm and spray for 4-6 hours with water at ambient temperature and a rate of 1 L/min

Due to the wide range of approaches taken for pre-conditioning the cycle helmets prior to testing, the best approach would be to develop specifications for helmet preconditioning based on an approach that either allows the harmonisation of these approaches or, when harmonisation is not possible, specifies the most stringent conditions specified. Thus, it is suggested that helmets be tested after ≥4 hours pre-conditioning at ambient (20±3 °C), hot (50±2 °C) and cold temperatures (-20±2 °C) and after artificial aging (UV irradiation by a 150

W xenon-filled quartz lamp for 48 hours at a 250 mm range, followed by full "crown down" immersion in potable water at ambient temperature to a crown depth of 305±25 mm for ≥4 hours). Although there is perhaps some question as to the relevance of some of the more stringent environmental conditions (e.g. will anyone cycle in temperatures of -20 °C?), testing to the most stringent conditions will allow the relative comparison of helmet performance globally against the toughest testing conditions.

Issues: Range of conditioning environments used across standards, conditions generally harsher than expected environmental conditions for cycling

Potential Solutions: Harmonisation or adhering to the most stringent conditions specified

Potential Protocol Change: Specify conditions based on standards harmonisation or most stringent conditions specified across standards as described above

B.3.6 Cycle Helmet Impact Energy

Current standards specify a range of cycle helmet impact energies. Two philosophies have been adopted for this approach, with the majority of standards specifying the impact speed for each tested anvil (often including an associated equivalent drop height) and with Snell B-95 specifically regulating the desired impact energy (Table B.6). Furthermore, CAN/CSA-D113.2-M89 also adopts a different philosophy by requiring a second impact test for adult helmets, which is performed at a lower velocity than the first impact.

Table B.6: Cycle helmet impact energy requirements

Standard	Flat Anvil			Kerbstone Anvil			Hemispherical/ Cylindrical Anvils		
	H /m	V /m/s	E /J	H /m	V /m/s	E /J	H /m	V /m/s	E /J
AS/NZS 2063	***1.5***	***5.42***							
ASTM F1447	***2.0***	***6.2***		***1.2***	***4.8***		***1.2***	***4.8***	
CAN/CSA-D113.2-M89	(1) 1.7 (2) 1.1	***(1) 5.7*** ***(2) 4.7***‡					1.1	***4.7****	
CPSC 16 CFR Part 1203	***2.0***	***6.2***	96.1†	***1.2***	***4.8***	57.6†	***1.2***	***4.8***	57.6†
EN 1078	***1.497***	***5.42***		***1.064***	***4.57***				
EN 1080	***1.497***	***5.42***		***1.064***	***4.57***				
JIS T 8134	***1.50***	***5.42***					***1.06***	***4.57***	
Snell B-95/Snell B-95C	2.24†	6.63†	***110***	1.47†	5.37†	***72***	1.47†	5.37†	***72***

Bold italics text denotes measure/s specified by standard, plain text denotes those that are calculable across the entire range of headforms and are included for comparative purposes

* Cylindrical anvil

† Calculated for all headforms due to constant headform mass requirements of 5 kg

‡ Second impact is specified for adult helmets only

When comparing the differences in impact velocities (which can be either directly specified or calculated), it is clear that a wide range of helmet impact velocities are specified. When impacting against a flat anvil, AS/NZS 2063, EN 1078, EN 1080 and JIS T 8134 have the least stringent helmet impact velocity requirements (5.42 m/s), whilst Snell B-95/B-95C has the toughest requirements (6.63 m/s). This is similar for kerbstone anvils, where EN 1078 and EN 1080 have the least stringent requirements (4.57 m/s) and Snell B-95/B-95C has the toughest requirements (5.37 m/s). For the hemispherical/cylindrical anvils it is clear that JIS T 8134 has the least stringent requirements (4.57 m/s) and Snell B-95/B-95C, again, has the toughest requirements (5.37 m/s). This variation in the impact velocities used between standards makes the harmonisation of standards difficult, with these differences further exasperated by the differences in pass/fail impact test criteria also used by each standard (discussed further in Appendix B.3.7).

One of the key weaknesses of the approaches currently adopted by the standards is the use of single velocity impact tests. This allows cycle helmet manufacturers to optimise the mechanical performance of the helmet to attenuate impacts at this particular impact energy. This approach may mean that there is a disproportionately greater risk of a head injury at lower impact energies due to the stiffness of the helmet liner being optimised to pass the requirements of the standards, without proper consideration given to the transfer of energy to the head at lower impact energies. To mitigate this issue it is important to test helmets across a range of impact energies to ensure that good helmet impact performance at higher impact energies does not compromise performance at lower impact energies.

This is currently implemented by CAN/CSA-D113.2-M89, which impact tests cycle helmets at impact velocities of 5.7 m/s and 4.7 m/s on the flat anvil. No rationale is, however, given for the selection of these impact velocities. To determine this, however, it is important to establish the minimum head impact energy that would cause injury to unhelmeted cyclists and to also assess the maximum impact energy a helmet should reasonably be expected to sustain before significantly increasing the risk of severe injury to the wearer. This range of impact velocities therefore requires determining for developing these protocols.

Issues: Range of impact energies used across standards, only one impact requirement per anvil type tested (aside from Canadian standards) and should instead specify a range, use of impact velocities/drop heights vs. impact energy

Potential Solutions: Develop effective impact performance range that tests the lower and higher energy attenuating performance of helmet, link into real-world impact speed range

Potential Protocol Change: Test at two impact drop test speeds for the same anvil

B.3.7 Cycle Helmet Impact Test Criteria

Current standards determine the protective performance of the cycle helmet by measuring the acceleration of the headform during impact and applying pass/fail criteria for the peak accelerations experienced during testing. When comparing these criteria across the current standards, however, it is apparent that there are again clear variations between standards (Table B.7). Whilst the US and Japanese standards permit headform accelerations of up to 300 *g* (ASTM F1447, CPSC 16 CFR Part 1203, JIS T 8134 and Snell B-95/B-95C), all remaining standards require accelerations of less than 250 *g*. AS/NZS 2063, CAN/CSA-D113.2-M89 and JIS T 8134 all specify further conditions to the pass/fail criteria, with AS/NZS 2063 and JIS T 8134c including requirements for the impact duration and CAN/CSA-D113.2-M89 including a 200 *g* limit for the second impact and a 1500 Gadd Severity Index limit that also considers impact duration. Finally, only CAN/CSA-D113.2-M89 provided any differentiation between the safety performance of child and adult helmets. This standard provides more stringent requirements for child helmets by reducing the peak acceleration criteria, recognising that paediatric injury biomechanics is very different to adult biomechanics.

Table B.7: Cycle helmet impact test headform acceleration pass/fail criteria requirements

Standard	Flat Anvil		Kerbstone Anvil		Hemispherical/ Cylindrical Anvils	
	Peak *g*	Other Criteria	Peak *g*	Other Criteria	Peak *g*	Other Criteria
Adult Helmets						
AS/NZS 2063	250 *g*	200 *g* for ≤3 ms 150 *g* for ≤6 ms				
ASTM F1447	300 *g*		300 *g*		300 *g*	
CAN/CSA-D113.2-M89	(1) 250 *g* (2) 200 *g*	GSI > 1500			250 *g**	GSI > 1500*
CPSC 16 CFR Part 1203	300 *g*		300 *g*		300 *g*	
EN 1078	250 *g*		250 *g*			
JIS T 8134	300 *g*	150 *g* for ≤4 ms			300 *g*	150 *g* for ≤4 ms
Snell B-95	300 *g*		300 *g*		300 *g*	
Young Child Helmets						
AS/NZS 2063	250 *g*	200 *g* for ≤3 ms 150 *g* for ≤6 ms				
ASTM F1447	300 *g*		300 *g*		300 *g*	
CAN/CSA-D113.2-M89	200 *g*	GSI > 1500			150 *g*	GSI > 1500
CPSC 16 CFR Part 1203	300 *g*		300 *g*		300 *g*	
EN 1080	250 *g*		250 *g*			
JIS T 8134	300 *g*	150 *g* for ≤4 ms			300 *g*	150 *g* for ≤4 ms
Snell B-95C	300 *g*		300 *g*		300 *g*	

GSI: Gadd severity index
* Cylindrical anvil

Comparing the above pass/fail criteria requirements is made more complex by the range of impact energies used by these standards (as discussed in Appendix B.3.6). Although it may seem like the US standards may be less stringent than other standards, it should be noted that these standards also test at considerably greater impact energies (Table B.6). It is clear that, for a given pass/fail criteria, CAN/CSA-D113.2-M89 and Snell B-95/B-95C provide more stringent test requirements, as both standards test at greater impact energies than the other standards also requiring similar pass/fail criteria (250 *g* and 300 *g*, respectively).

When considering the range of impact energy approach discussed in in Appendix B.3.6, it is clear that a range of impact injury criteria, specific to the impact energies that the tests are trying to protect against, need to be developed. This should target injury criteria related to the minimum head impact energy that would cause injury to unhelmeted cyclists and the maximum impact energy a cycle helmet should reasonably be expected to sustain before significantly increasing the risk of severe injury to the wearer. Furthermore, the practice of using linear accelerations to define head injury criteria is widely recognised as an outdated approach, with current state-of-the-art methods taking into consideration the six degree-of-freedom motion of the head and the strain properties of the brain. A discussion around which injury criteria should be used for these protocols is discussed further in Appendix C.

Issues: Range of impact criteria used across standards, only one requirement per anvil type (aside from Canadian standards) and should instead specify a range, only two standards consider impact duration, potentially outdated approach to assessing injury risks

Potential Solutions: Link injury criteria to impact performance testing range that tests the lower and higher energy attenuating performance of helmet, link into current state-of-the-art injury criteria.

Potential Protocol Change: Specify two head impact criteria for the same anvil based on state-of-the-art injury criteria

B.3.8 Number of Cycle Helmet Impact Tests

Due to the large number of pre-conditioning environments, impact anvils and requirements to impact the cycle helmet multiple times, current standards vary significantly in the total number of helmets required for testing (Table B.8). When observing the total number of cycle helmet samples required for testing, ASTM F1447, CAN/CSA-D113.2-M89 and CPSC 16 CFR Part 1203 specify the greatest number of helmets (8), whilst EN 1078, EN 1080 and JIS T 8134 only require three helmets for completion of the impact testing. CAN/CSA-D113.2-M89 further specifies the greatest number of impacts per helmet (6, when taking all anvil strikes into consideration), whilst AS/NZS 2063 requires just the single impact per helmet. ASTM F1447, CAN/CSA-D113.2-M89, CPSC 16 CFR Part 1203, EN 1078, EN 1080 and JIS T 8134 all also require impact tests with either the kerbstone or hemispherical anvil directly after impacting against the flat anvil.

Table B.8: Cycle helmet impact test numbers

Standard	Total No. of Test Samples	Flat Anvil		Kerbstone Anvil		Hemispherical/ Cylindrical Anvils	
		No. of Samples	No. of Tests per Sample	No. of Samples	No. of Tests per Sample	No. of Samples	No. of Tests per Sample
AS/NZS 2063	4†	4	1				
ASTM F1447	8	4	1	4	1	4‡	1‡
CAN/CSA-D113.2-M89	8	4	4			4	2
CPSC 16 CFR Part 1203	8	4	2	4	1	4‡	2‡
EN 1078/EN 1080	3Δ	3	1	3‡	1‡		
JIS T 8134	3	3	2			3‡	2‡
Snell B-95/B-95C	5^	3¶	4	3¶	4	3¶	4

* Total number of samples tested per helmet shell/liner combination

† Not including a further four load distribution tests

‡ Testing performed using samples used in flat anvil test

Δ Fourth sample kept in reserve as reference sample

^ Sixth sample kept in reserve as reference sample

¶ Each anvil shall be used at least once

To reduce manufacturer and test house costs it is clear that the testing performed by these protocols must minimise both the total number of helmets tested and the total number of impacts per helmet. Whilst there is great variation between the approaches taken by the different standards, it is likely that, due to the greater testing demands of these protocols, there will be a greater requirement to optimise the total number of test helmets and total number of impacts per helmet. Multiple impacts per test helmet sample may therefore be required; with the above standards all requiring the tests sites to be a minimum distance away from each other. EN 1078 provided the most stringent requirements by specifying a minimum separation distance of 150 mm along the chord of the helmet.

Issues: Range of impact schedules used across standards, differing number of cycle helmets and impact tests between standards

Potential Solutions: Minimise costs for manufacturer and test house by reducing number of helmets to be tested and the number of impact tests

Potential Protocol Change: Implement minimised testing schedule

B.4 Cycle Helmet Retention Test Characteristics

All standards examined by this review involved a cycle helmet retention system strength test, with the exception of EN 1080. EN 1080 differs from the other standards by testing the retention system for its ability to self-release between a specified load range (90-160N) to avoid strangulations in playgrounds. The retention test is usually set up by placing the helmet on a fixed headform, with an inertial hammer suspended from the retention straps. The weight of the inertial hammer is then dropped through either a specified distance or a distance by which the specified impact energy is achieved. The test pass criteria are usually a limit on the amount of elongation sustained by the straps. When comparing the retention system strength test standards, however, it is clear that there are slight variations between the test specifications and pass/fail criteria used (Table B.9).

Table B.9: Cycle helmet retention test requirements

Standard	Pre-Conditioning					Inertial Hammer		Impact Energy		Criteria		
	Ambient	Hot	Cold	Immersion	Ageing	Assembly Mass† /kg	Hammer Mass /kg	Height /mm	Energy /J	Dynamic Ext. /mm	Residual Ext. /mm	Release
AS/NZS 2063	✓	✓	✓	✓		7	10	***250***	24.5	30		
ASTM F1447		✓	✓	✓		7	4	***600***	23.5	30		
CAN/CSA-D113.2-M89	✓	✓	✓	✓		7	2	1020	***20***	25	12	
CPSC 16 CFR Part 1203	✓	✓	✓	✓		7	4	***600***	23.5	30		
EN 1078*			✓		✓	5	4	***600***	23.5	35	25	✓
JIS T 8134	✓					7	4	***600***	23.5	35		✓
Snell B-95/B-95C	✓	✓	✓	✓		7	4	***600***	23.5	30		✓

Bold italics text denotes measure/s specified by standard, plain text denotes those that are calculable across the entire range of headforms and are included for comparative purposes

* EN 1080 excluded from analysis due to method requiring different testing principles

† Mass of retention system strength test apparatus excludes hammer mass

From this comparison of standards it is difficult to establish what the most stringent testing standards are, as a range of hammer masses and impact velocities are used. Despite similar impact energies, the use of different impact velocities and hammer masses to achieve this means that the retention systems, which have viscoelastic mechanical properties, are likely to perform differently for the testing philosophies adopted by each standard. This is further compounded through the use of a range of pass/fail criteria and conditioning environments.

Two approaches could, however, be adopted for the testing of retention system strength for these protocols. The first could take the approach adopted by EN 1080, which quasi-statically loaded the retention system whilst monitoring the force/deflection characteristics of the retention system. The second could dynamically load the helmet retention system, as performed by the majority of testing standards, before reporting the peak and residual retention system extensions.

Issues: Range of hammer masses, drop heights and criteria between standards, EN 1080 takes a quasi-static approach to testing retention system strength

Potential Solutions: Perform dynamic retention strength tests, perform quasi-static retention system strength tests

Potential Protocol Change: Dynamic and quasi-static retention system strength tests

B.5 Cycle Helmet Stability Test Characteristics

Cycle helmet stability tests are used to assess the way in which the retention system of a helmet keeps it on the head of a rider. These tests are performed by the standards through either dynamic or quasi-static testing; which fit the helmet to a fixed headform and apply a tangential load to the front and rear of the helmet by the appropriate method. All but two standards were found to use dynamic inertial hammer drop tests to test retention system stability, with AS/NZS 2063 and CAN/CSA-D113.2-M the only two standards using a quasi-static testing approach (Table B.10 & Table B.11). It must also be noted that EN 1080 does not include a cycle helmet stability test, presumably as cycle helmet retention systems are designed to release between loads of 90-160 N when conforming with this standard and so there is no way to repeatably perform this test.

Table B.10: Cycle helmet dynamic stability test requirements

Standard	Test Set Up			Inertial Hammer		Impact Energy		Criteria	
	Headform Angle*	Loading Angle†	Front /Rear	Assembly Mass‡ /kg	Hammer Mass /kg	Height /mm	Energy /J	Helmet Ejection	Coronal Plane$^{\Delta}$
ASTM F1447	135°	45°	Both	7	4	***600***	23.5	✓	✓
CPSC 16 CFR Part 1203	135°	45°	Both	1	4	***600***	23.5	✓	
EN 1078	0°	45°	Rear	3	10	***175***	17.2	✓	
JIS T 8134	0°	45°	Rear	3	10	***175***	17.2	✓	
Snell B-95/B-95C	135°	45°	Both	1	4	***600***	23.5	✓	

Bold italics text denotes measure/s specified by standard, plain text denotes those that are calculable across the entire range of headforms and are included for comparative purposes

* Angle calculated between the vertical plane and the coronal plane of the headform

† Angle calculated between the cable and the transverse plane of the headform

‡ Mass of retention system strength test apparatus excludes hammer mass

$^{\Delta}$ Parts of the coronal plane of the headform that were previously covered by the helmet become exposed after testing

Table B.11: Cycle helmet quasi-static stability test requirements

Standard	Test Set Up			Quasi-Static Loads		Criteria	
	Headform Angle*	Loading Angle†	Front /Rear	Load /N	Time Period /s	Test Band‡	Helmet Angle$^{\Delta}$
AS/NZS 2063	0°	0°	Both	50	15-30	✓	
CAN/CSA-D113.2-M89	N/S	N/S	Both	250	≥5		45°

N/S: Not specified

Bold italics text denotes measure/s specified by standard, plain text denotes those that are calculable across the entire range of headforms and are included for comparative purposes

* Angle calculated between the vertical plane and the coronal plane of the headform

† Angle calculated between the cable and the transverse plane of the headform

‡ Test band obscuration/exposure calculated for region bounded by the basic plane, a line a set distance above and parallel to the basic plane and two vertical lines 25 mm either side of the mid-sagittal axis

$^{\Delta}$ Angle calculated between the basic plane and a line between the point where the force is being applied and a point at the intersection of the basic and coronal planes

From this comparison of standards it is clear that, for dynamic helmet stability, the US test standards (ASTM F1447, CPSC 16 CFR Part 1203 and Snell B-95/B-95C) are more stringent than the European and Japanese testing standards. When comparing these to AS/NZS 2063 and CAN/CSA-D113.2-M89, however, the philosophies adopted by these quasi-static test standards do not allow for direct comparison with the dynamic testing standards adopted by the US, Europe and Japan. It is difficult to establish which of these approaches is the most appropriate; however, it may be postulated that the dynamic testing is more likely to reflect the stability response of the helmet during impact, whilst the quasi-static testing is more likely to reflect helmet stability during normal usage. Therefore, for the purposes of these protocols, it is clear that the dynamic testing approach should be adopted. However, it is important to point out that the quasi-static testing approach could be adopted for a cycle helmet comfort rating scheme.

Issues: Significant differences between testing philosophies and requirements, with US standards being the most stringent safety testing standards, pass/fail criteria used only

Potential Solutions: Adopt US standards, measure change in angle, quasi-static testing for comfort rating

Potential Protocol Change: Dynamic cycle helmet stability tests performed to US standard specifications

B.6 Cycle Helmet Field of Vision Test Characteristics

Despite a number of standards not regulating the vertical field of vision requirements and CAN/CSA-D113.2-M89 not regulating the field of vision at all, field of vision requirements, when specified, vary only slightly between standards (Table B.12). Snell B-95 provides the most stringent horizontal field of vision clearance requirements, whilst both EN 1078 and EN 1080 provide the most stringent vertical field of vision clearance requirements.

Table B.12: Minimum cycle helmet field of vision clearance requirements

Standard	Horizontal		Vertical	
	Left	Right	Upward	Downward
AS/NZS 2063	105°*	105°*	25 mm*	
ASTM F1447	105°*	105°*		
CAN/CSA-D113.2-M89				
CPSC 16 CFR Part 1203	105°*	105°*		
EN 1078/EN 1080	105°†	105°†	25°†	45°*
JIS T 8134	105°‡	105°‡		
Snell B-95/ Snell B-95C	110°Δ	110°Δ	25°†	30°*

* Measurements taken in/from/above the basic plane

† Measurement taken in/from the reference plane

‡ Measurement taken for the area bounded by the reference and basic plane

Δ Measurement taken for the area bounded by the reference and S_4 (S_2 for children) planes

It is important to consider the trade-off between the field of vision and extent of protection when defining the field of vision requirements. Any increase in field of vision requirements causes a decrease in the protective design area and so there is a clear trade-off between primary safety (vision) and secondary safety (impact protection) requirements. However, as it is well established that the maximum possible peripheral vision is 110° to the left and right;

all standards will currently require helmets to, within reason, permit the maximum achievable horizontal field of vision. When considering the vertical field of vision, it is also clear that restricting helmet coverage by increasing the field of vision offers limited safety benefit when compared to the benefit gained by allowing greater helmet coverage.

Issues: Pass/fail criteria used only, maximum field of vision not reported

Potential Solutions: Remove vertical field of vision requirements, not appropriate to rate

Potential Protocol Change: Remove vertical field of vision requirements, keep minimum horizontal field of view requirement check at 105°

B.7 Cycle Helmet Certification Approach

In addition to prescribing the test methods and performance criteria, these standards also include requirements as to how and where the testing is to be performed, who certifies the helmet and if there is to be any ongoing testing to retain certification.

The standards may be grouped by the type of certifying body specified within the standard, with this split into certification by the manufacturer (AS/NZS 2063, ASTM F1447 and CPSC 16 CFR Part 1203) and by an independent testing body approved by the relevant standards organisation (CAN/CSA-D113.2-M89, EN 1078, EN 1080, JIS T 8134 and Snell B-95/B-95C). Despite this top level grouping, there remain considerable differences in the approaches taken within each group.

AS/NZS 2063 requires that testing is performed by the manufacturer on a sample from all batches of helmets manufactured, with verification testing performed by an independent testing body. ASTM F1447 and CPSC 16 CFR Part 1203 both require self-certification by the manufacturer, with follow-up testing only performed in litigation cases. CAN/CSA-D113.2-M89 specifies that testing must be carried out at an approved test laboratory and the only follow-up is an inspection of the manufacturer’s factory. European (EN 1078 and EN 1080) and Japanese standards both require testing at approved national test laboratories, which then award the appropriate product certification. Finally, the Snell B-95/B-95C standards involve an initial certification testing new helmet models with the Snell Foundation testing helmets bought from stores at randomised intervals after certification.

Issues: Wide range of certification processes, not all valid for protocols

Potential Solutions: Approved body provides certification testing. Process could allow manufacturers to self-certify and approved body could audit cycle helmet models and manufacturers

Potential Protocol Change: Approved body could allow manufacturers to self-certify and audit cycle helmet models and manufacturers

Appendix C Head Injury Criteria Review

C.1 Background on Head Injuries

C.1.1 Introduction

This Appendix provides a summary of the state-of-the-art surrounding the characteristics of blunt trauma to the head during impacts. It provides an overview of the brain injury and skull fracture continuums, before summarising the various injury mechanisms associated with these injuries.

This state-of-the-art review is based primarily upon information and discussion abstracted from the two key review articles (Hoshizaki *et al.*, 2013a; Yoganandan *et al.*, 2015). The information from these articles has been abstracted and then slightly altered to make it appropriate to this particular project.

C.1.2 Overview

Blunt head injuries are inherently complex in nature and describing the relationship between the trauma and resulting injury can be extremely difficult. A large volume of research has investigated the mechanisms of head injury through experimental, reconstructive and finite element modelling methods with the aim of reducing the incidence and severity of such injuries.

Experimental research has been performed on different types of brain tissue in order to assess the failure thresholds in both mechanical and functional terms at a mesoscopic level. Human responses were then linked to the output levels of stress and strain (King *et al.*, 2003). As the name suggests, reconstructive research involves the reconstruction of how an individual was impacted in order to ascertain how the resulting injury is connected to accelerations, stresses and strains encountered during the impact (Zhang *et al.*, 2004; Kleiven, 2007b; Post *et al.*, 2012a). Finite element modelling provides a further tool to simulate the responses of the head and brain tissues resulting from impacts and can be used to predict the level of injury that may be experiences from certain impact conditions.

There are several different types of head injury categories with a variety corresponding mechanisms and predictive variables (Post, 2013). These multiple categories of injury also result in differing levels of injury severity, resulting in a head injury continuum effect where the category of injury is inherently linked to impact severity. Falls, collisions, projectiles and punches make up the most common mechanisms of brain injury. The contribution of these mechanisms to injury outcomes within the head injury continuum will be examined in this review.

C.1.3 The Head Injury Continuum

Head injuries can be subdivided up into four broad categories: skeletal, focal, diffuse and external soft tissue injuries. Skeletal injuries include linear, depressed, diastatic and basilar fractures to the skull. Focal brain injuries are caused by mechanical deformations due to tension, shear and compression and include Epidural haematoma (EDH), subdual haematoma (SDH), subarachnoidal haemorrhage (SAH), and contusions (Oeur *et al.*, 2013; Post, 2013). Diffuse injuries can vary in symptoms, recovery times and consequences and can include mild concussion, cerebral concussion associated with loss of consciousness, diffuse injury and diffuse axonal injuries (McKee *et al.*, 2010; Meaney and Smith, 2011).

C.1.4 Skull and Facial Fractures

The skull and facial bones are biomechanically very complex with varying bone geometry and surfaces. This makes the fracture response after an impact highly dependent on the location of impact and shape of impactor.

Mandible fractures are the second most common facial fracture after nasal bone fractures due to blunt trauma (King *et al.*, 2004; Hwang and You, 2010). In an Indian study gunshot wounds and assaults attributed to half of mandible fractures and motor vehicle crashes accounted for a third of mandible fractures (Pappachan and Alexander, 2006). Motor vehicle collisions (including bicyclists and pedestrians), assaults and falls have been found to be the most common cause of skull fracture (Liu-Shindo and Hawkins, 1989; Chee and Ali, 1991; Jager *et al.*, 2000).

There are four major types of skull fracture categories including linear, depressed, diastatic and basilar fractures. Linear skull fractures traverse the full thickness of the skull from the outer to inner table. They include straight or curved fracture lines, with no displacement of the bone, and tend to occur at unsupported regions of the skull (e.g. across the supra-orbital ridges). The most common cause of injury is blunt force trauma where the impact energy transferred over a wide area of the skull. Depressed skull fractures are caused by a force applied in a focussed area. The outer and inner aspects of the skull are driven inwards, often causing damage to the brain or its meninges. This often results in a comminuted fracture with fissures radiating outwards from the depressed area of the skull. Diastatic fractures are linear fractures that occur along the suture lines of the skull and are predominantly experienced by younger children and infants. Finally, basilar fractures are linear or ring fractures that occur in the floor of the cranial vault (skull base). This fracture mechanism characteristically requires greater forces to cause than other areas of the skull and is primarily seen in high-energy incidents involving loads being transferred up the spine.

C.1.5 Traumatic Brain Injury

Traumatic brain injury (TBI) is a major cause of mortality, hospitalisation and disabilities worldwide. Each year in England and Wales around 1.4 million people attend accident and emergency departments with head injuries and 200,000 of these are admitted into hospital (NICE, 2014). Skull fractures or evidence of brain damage were present in 20% of admissions and although death only occurred in 0.2% of admissions, head injury is the most common cause of death and disability in the under 40s in the UK. TBIs are a leading cause of long-term disabilities among survivors and those who sustain TBI often have decreased life expectancy compared with the general population (Langlois *et al.*, 2006; Brooks *et al.*, 2013; Ma *et al.*, 2014; Brooks *et al.*, 2015). Due to the serious consequences of brain injuries, a great deal of research investigating the mechanisms of TBI has been undertaken in an effort to reduce its incidence.

Brain injuries can be separated into two broad categories; focal injuries and diffuse injuries. Focal injuries are defined as those in which a lesion is large enough to be visualised by the naked eye. They include injuries such as:

	Injury Description	Cause
Contusions	Cerebral contusions on the same side as the impact site (coup) or on the opposite side (contrecoup)	Direct impact trauma to the head
Epidural (EDH) haematoma	An accumulation of blood between the inner skull surface and dura mater	Trauma to the skull – often associated with skull fractures
Subdural (SDH) haematoma	Blood located between the arachnoid membrane and dura mater	Lacerations of cortical veins or arteries, tearing of veins bridging the subdural space, bleeding of a large contusion. Caused by rapid acceleration - direct impact not required
Intracerebral (ICH) haematoma	Rupture of small blood vessels within the brain	Trauma to the head
Subarachnoid (SAH) haemorrhage	Bleeding into the subarachnoid space	Non-traumatic SAH caused by aneurysms, traumatic SAH caused by bleeding of contusions, laceration of arteries, increase in intravascular pressure

Diffuse brain injuries on the other hand are not often associated with visible brain lesions; instead more widespread disruption of neurological function or structures is apparent.

	Injury Description	Cause
Mild concussion	Temporary disturbance of neurological function	No loss of consciousness
Classic cerebral concussion	Temporary reversible deficiency of neurological function	Temporary loss of consciousness (less than 24 hours)
Diffuse Injury	Residual neurological defects such as memory loss or reduced motor function	Prolonged loss of consciousness (more than 24 hours)
Diffuse axonal injury (DAI)	Mechanical disruption of many axons resulting in severe memory and motor deficits	Immediate loss of consciousness lasting for days to weeks

The term traumatic brain injury is often used to describe a variety of injuries. Research has been undertaken to examine the individual injuries that make up this head injury continuum in terms of impact magnitudes and occurrence of injury.

Gennarelli *et al.* (1981b) performed a study on 424 hospital patients presenting head injuries from falls, assaults and vehicle collisions. Cerebral and cortical concussion were most commonly sustained but were associated with low mortality rates whereas SDH and shearing injuries combined caused 57% of all of the deaths. A further study across seven head injury centres showed that these types of injuries accounted for 64% of deaths.

Post (2013) found that magnitude and occurrence of injuries followed the same pattern, with SDH being produced with the lowest magnitude impact and occurring the most often and EDHs produced with the highest magnitude impacts and occurring least often (Figure C.1) (Post, 2013; Post *et al.*, 2014).

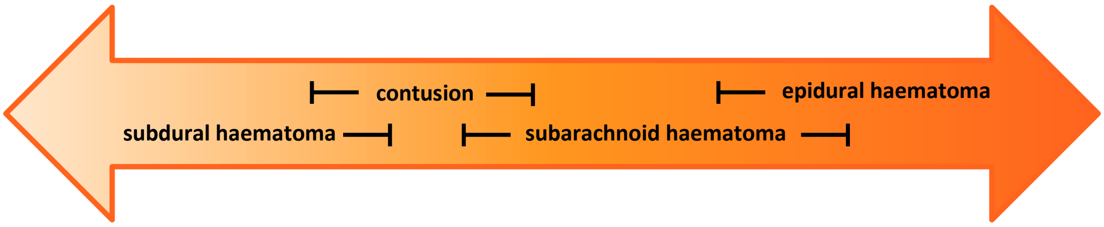

Figure C.1: Illustration of the focal brain injury continuum (adapted from Hoshizaki *et al.* (2013b))

A similar continuum has been proposed for concussive injuries (Figure C.2). Persistent concussive syndrome (PCS) occurs when symptoms of concussion last in excess of 1 month (Alves *et al.*, 1993; Marshall *et al.*, 2012). Transient concussion usually lasts for a maximum of 14 days before systems resolve and sub-concussive injuries result in no, or very brief symptoms of concussion (DeWitt *et al.*, 2013).

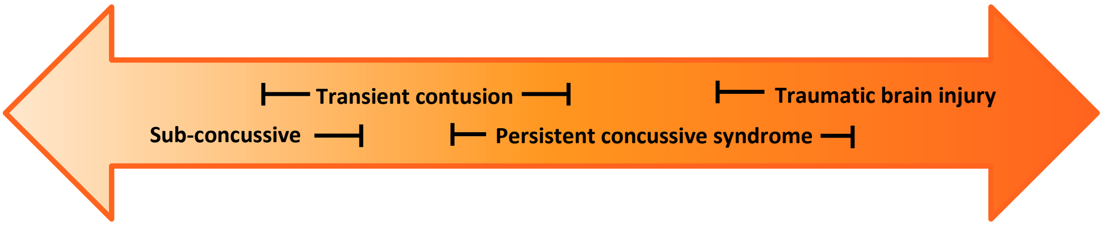

Figure C.2: Illustration of the diffuse concussion severity continuum (adapted from Hoshizaki *et al.* (2013b))

The focus of much research has been around quantifying mechanism of injury in terms of variables such as force and linear and rotational accelerations. (King *et al.*, 2003; Post and Hoshizaki, 2012a). The most commonly researched TBI are contusions, SDH, SAH and DAI particularly due to their high prevalence and mortality rates (Gurdjian, 1975; Bradshaw *et al.*, 2001; Kleiven, 2003; Post, 2013).

Research into the kinematics of impacts has showed that SDH and DAI are caused by a mechanism that involves rotational accelerations (Kleiven, 2003; Post, 2013). These rotations cause high relative motion between the brain and the skull causing the bridging veins to ruptures leading to SDH and also shearing of the white matter within the brain leading to DAI (Al-Bsharat *et al.*, 1999; Bradshaw *et al.*, 2001).

On the other hand contusions and EDH were more closely associated with linear accelerations (Gurdjian, 1976). Modern research has identified that rotational acceleration has more of an influence on the prediction brain stresses and strains than linear acceleration (Forero Rueda *et al.*, 2011; Post *et al.*, 2012b; Post *et al.*, 2013).

C.2 Comparison of Head Injury Criteria

C.2.1 Introduction

This Appendix provides a summary of the state-of-the-art surrounding head injury criteria. It provides an overview of each injury criterion and categorises them by five key categories; localised loading skull fracture criteria, translational head injury criteria, rotational head injury criteria, combined translational and rotational head injury criteria, and brain tissue stress and strain criteria.

This state-of-the-art review is based primarily upon information and discussion abstracted from a recent key review article but has also been supported by several other key research articles (Nahum and Melvin, 2012; Post and Hoshizaki, 2012b; Fernandes and deSousa, 2015; Hernandez *et al.*, 2015; Willinger *et al.*, 2015). The information from these articles has been abstracted and then slightly altered to make this Appendix appropriate to this particular project.

C.2.2 Overview

When the head is exposed to loads that exceed the capacities of its protective features (bone, soft tissues, etc.), the results are head injuries which can be severe and, in the worst cases, fatal. Head injuries are usually sustained from either a direct impact to the head or through purely inertial effects from an indirect impact applied to the head-neck complex during the rapid acceleration or deceleration of the torso. In both instances linear and rotational accelerations are experienced by the head, primarily due to the connections within the head-neck complex (Aare, 2003).

The mechanisms which produce head injuries during impacts have been researched for more than 50 years with the aim of gaining a better understanding of these mechanisms and to establish associated tolerance levels. In order to evaluate effectively the risk of a head injury being sustained, researchers have developed many injury criteria. These are also used to assess the protective potential of equipment such as helmets.

Injury criteria are generally constructed though the testing of animals or cadaveric human specimens. A known stimulus such as force or acceleration is applied and a response is measured. Injury risks curves are then established from the injury inducing threshold levels of the stimulus. Non-injury levels of stimuli can be established through voluntary testing. Anthropometric test devices (ATD) can be used in conjunction with accident data to reconstruct the incident and replicate the mechanism of injury. The outcomes can then be linked to injuries actually sustained during the accident to understand stimuli levels of real world injuries. Theoretical and mathematical models can provide further information on injury criteria.

Currently, numerous studies present many injury criteria and thresholds for the evaluation of injury occurrence and severity. These can be roughly divided into five key categories:

- Injury criteria based on the localised loading of the head
- Injury criteria based on the translational acceleration of the head;
- Injury criteria based on the rotational acceleration of the head;
- Injury criteria based on the combined translational and rotational acceleration of the head; and
- Injury criteria based on the stresses, strains and shearing of the brain tissue.

The various head injury criteria associated with each of the above five categories will be examined in this review and a synthesis of the injury severity thresholds associated with these criteria will be discussed. This will provide information on how the biomechanics of an impact affect injury and injury severity, provide a critical review of the relative merits of the various approaches outlined by the literature and offer guidance on the injury criteria and thresholds that should be adopted by the NCHAP rating scheme.

C.2.3 Localised loading based head injury criteria

Injury criteria for skull fractures can be based on localised loading of the skull and are dependent on skull thickness and shape of impactor. A summary of peak skull fracture forces across different regions of the skull is shown in Table C.1. Depressed skull fractures in the temporal region have been shown to occur when an area under 5 cm^2 is impacted at a localised pressure greater than 4 MPa (Hume *et al.*, 1995). Further experiments to establish stress thresholds of cranial bone demonstrated that in tension compact bone breaks at between 48 and 128MPa and in compression cancellous bone breaks at between 32 and 74 MPa (Robbins and Wood, 1969; McElhaney *et al.*, 1970; Melvin *et al.*, 1970). In terms of global strain energy of the skull, a 50% risk of skull fracture has been correlated to energy of 2.2 J with failure levels for frontal impacts occurring at 22-24 J and 5-15 J for temporal impacts (Raul *et al.*, 2006; Monea *et al.*, 2014).

Table C.1: Peak skull fracture force for different regions of the skull (Fernandes and deSousa, 2015)

Impact Region	Force (kN)	Reference
Frontal	4.0	(Schneider and Nahum, 1972)
	4.2	(Nahum *et al.*, 1968)
	4.3-4.5	(Yoganandan *et al.*, 1994)
	4.7	(Allsop *et al.*, 1988)
	5.5	(Hodgson and Thomas, 1972)
	6.2	(Advani *et al.*, 1975)

Impact Region	Force (kN)	Reference
	15.6	(Voo *et al.*, 1994)
Temporal	2.0	(Schneider and Nahum, 1972)
	3.4-4.4	(Yoganandan *et al.*, 1994)
	3.6	(Nahum *et al.*, 1968)
	5.2	(Allsop *et al.*, 1991)
	6.2	(Voo *et al.*, 1994)
Occipital	11.7-11.9	(Yoganandan *et al.*, 1994)
	12.5	(Advani *et al.*, 1982)
Parietal	3.5	(Hume *et al.*, 1995)
Vertex	3.5	(Yoganandan *et al.*, 1994)

C.2.4 Translation-only based head injury criteria

C.2.4.1 Peak linear acceleration

The simplest form of head injury criterion is to consider purely the peak linear acceleration. For more than fifty years linear accelerations have been used to predict head injury risk (Gurdjian and Lissner, 1961). After an impact the skull deforms and the brain undergoes accelerations which results in a change of pressure (Gurdjian *et al.*, 1966; Thomas *et al.*, 1966). Intracranial pressure differences were observed during testing on animals and these changes in pressure can cause shear stresses to the brain tissue (Gurdjian and Lissner, 1961; Gurdjian *et al.*, 1966; Nusholtz *et al.*, 1987). During testing the kinematics of the impact are monitored in order to produce injury risks.

Although this method ignores the influence of impact durations, some studies present the time durations associated with peak linear acceleration values. These and further thresholds are shown in Table C.2.

Table C.2: Peak linear acceleration thresholds for different injury mechanisms and severities (Adapted from (Fernandes and deSousa, 2015))

Injury	Thresholds	Reference
Head Injury	80 *g* for 3 ms	(Stalnaker *et al.*, 1971; Versace, 1971; Got *et al.*, 1978)
	50% probability AIS2+: 116 *g*	(Peng *et al.*, 2012)
	50% probability AIS3+: 162 *g*	"
	AIS4: 200-250 *g*	(Newman, 1986)
	AIS5: 250-300 *g*	"
	AIS6: >300 *g*	"

Injury	Thresholds	Reference
Skull Fracture	5% risk: 180 *g*	(Mertz *et al.*, 1997)
	40% risk: 250 *g*	"
	50% risk: 135 *g*	(Peng *et al.*, 2012)
mTBI	25% probability: 559 m/s^2	(King *et al.*, 2003)
	50% probability: 778 m/s^2	"
	75% probability: 965 m/s^2	"
	50% probability: 762 m/s^2	(Newman *et al.*, 2000)
	95% probability: 1131 m/s^2	"
	85 *g* for t = 10-30 ms	(Zhang *et al.*, 2004)
	103 *g*	(Brolinson *et al.*, 2006)
	82-146 *g*	(Schnebel *et al.*, 2007)
	103 *g*	(Frechede and McIntosh, 2009)
	90 *g*	(Gurdjian *et al.*, 1966)
Concussion	81 *g*	(Duma *et al.*, 2005)
	60.51-168.71 *g*	(Guskiewicz *et al.*, 2007)
	105±27 *g*	(Rowson and Duma, 2011)
	74±21 *g*	(McAllister *et al.*, 2012)
	50% probability: 65.1 *g*	(McIntosh *et al.*, 2014)
	75% probability: 88.5 *g*	"
SDH	130 *g*	(Willinger and Baumgartner, 2003b)

Linear accelerations have proved to be an effective injury metric and have been successful in reducing sporting head injuries such as skull fractures and traumatic brain injuries (Hoshizaki and Brien, 2004). However, injuries such as concussion still remain prevalent and it is widely thought that may be the result of the rotational kinematics experienced during impact. The injury criteria and thresholds for these will be discussed further in Section 0 onwards.

C.2.4.2 Gadd severity index

The peak linear acceleration method ignores the influence of impact durations which were subsequently found to be significant (Gurdjian *et al.*, 1966). An acceleration-time tolerance curve was developed using animal and cadaver impact data; Wayne State Tolerance Curve (WSTC):

$$\bar{a}^m T = N$$

Where $\bar{a}^m$ is the peak average acceleration measured for the period T.

The curve demonstrated that lower accelerations could be tolerated for a longer duration than high accelerations. Further testing on animals and human cadaveric skulls was conducted in Japan and provided similar results to the WSTC (Ono *et al.*, 1980).

The Gadd Severity Index (GSI) was created from the WSTC to provide a metric that could be more easily applied in the automotive industry for the development of protective head devices (Gadd, 1966). He proposed the use of 2.5 as the exponent for internal head injury taken from an approximation of the slope of a log–log plot of the WSTC thus:

$$GSI = \int_{t_0}^{t} a^{2.5} \cdot dt$$

where a is acceleration.

A threshold value of 1000 was used by Gadd and his colleagues at General Motors Corp. for serious internal head injury. The use of the integral rather than peak values allowed for the whole impact pulse to be incorporated and pulses with a duration of 50 ms or longer were disregarded as usually head impacts were less than 15 ms.

C.2.4.3 Head injury criterion

The head injury criterion (HIC) is currently used worldwide for the regulation of head injuries. It was developed after a the publication of a critical review of the GSI, identifying a number of flaws in the assertions made by Gadd (Versace, 1971). An alternative formula for GSI was then proposed and subsequently adopted by NHTSA[3] for FMVSS 208[4] in 1972. It is expressed by the below equation:

$$HIC = \left\{\left[\left(\frac{1}{t_2 - t_1}\right)\int_{t_1}^{t_2} a(t) \cdot dt\right]^{2.5} (t_2 - t_1)\right\}_{max}$$

where a(t) is the resultant head acceleration in g, t_1 and t_2 are the initial and final times (in seconds) of the interval during which HIC attains a maximum value and the interval t_2-t_1 is the boundary of the time duration intervals that define the HIC analysis, typically 15 ms (ISO). As with GSI a tolerance of 1000 was proposed for HIC.

Prasad and Mertz developed a head injury risk curve demonstrating that a 50 % risk of skull fracture corresponds to a HIC level of 1450 and a 5 % risk of skull fracture to a HIC level of 700 (Prasad and Mertz, 1985).

Originally NHTSA chose to adopt a 36 ms duration with a threshold of 1000 in FMVSS 208. Transport Canada however adopted a 15 ms HIC duration, but set the HIC limit to a more stringent threshold of 700 corresponding to a 5 % risk of skull fracture or serious head injury. In 2003 NHTSA adopted the same values as Transport Canada as this produced a more severe test. Scaling was used to establish the limits of smaller dummies and can be seen in **Table C.3** below(Alliance, 1999; Eppinger *et al.*, 1999).

[3] National Highway Traffic Safety Administration

[4] U.S. Federal Motor Vehicle Safety Standard No. 208, Advanced Airbags

Table C.3: HIC thresholds (15 ms duration) for 5 % risk of skull fracture or serious head injury (Alliance, 1999; Eppinger *et al.*, 1999)

Dummy Size	Thresholds
5th percentile	700
6 year old	700
3 year old	500
CRABI	390

In summary in the HIC two parameters, acceleration and its duration over an impact period, are used as suitable predictors of injury occurrence and severity. The thresholds for this injury criterion are presented in Table C.4 below.

Table C.4: Head injury criteria (HIC) thresholds for different injury mechanisms and severities (Fernandes and deSousa, 2015).

Injury	Thresholds	Reference
Head Injury	Severe but not life-threatening: 1000	(Shuaeib *et al.*, 2002)
	8.5% probability of death: 1000	(Hopes and Chinn, 1989)
	31% probability of death: 2000	"
	65% probability of death: 4000	"
	16% probability of life-threatening injuries: 1000	(Horgan, 2005)
	99% probability of life-threatening injuries: 3000	"
	50% probability of AIS2+: 825	(Peng *et al.*, 2012)
	50% probability of AIS3+: 1442	"
mTBI	25% probability (for HIC_{15}): 136	(King *et al.*, 2003)
	50% probability (for HIC_{15}): 235	"
	75% probability (for HIC_{15}): 333	"
	50% probability (for HIC_{15}): 240	(Nahum *et al.*, 1968)
	95% probability (for HIC_{15}): 485	"
	240	(Zhang *et al.*, 2004)
Skull Fracture	50% risk: 667	(Marjoux *et al.*, 2008)
SDH	50% risk: 1429	(Marjoux *et al.*, 2008)
Neurological injury	50% risk of moderate injury: 533	(Marjoux *et al.*, 2008)
	50% risk of serious injury: 1032	"
Concussion	200	(Duma *et al.*, 2005)

Controversy surrounds HIC, as far as any theoretical basis is concerned; many reasons are advanced, the main one being that HIC is based solely on the measurement of linear accelerations at the centre of gravity of the head and that its units are defined in “seconds”. Rotational accelerations also cause shear deformation in the brain; however, the tolerance limits of rotational accelerations have not been introduced into any regulation. Furthermore, the short-duration aspect of the WSTC is based on unidirectional translational accelerations, measured at the back of the head, and is assumed to be representative of measurements at the centre of gravity of the head. The HIC may therefore not be relevant for a deforming, multi-modal structure composed of fluids and solids and about to fracture. Finally, HIC was developed based on the short duration (2-6 ms) part of the WSTC, with the application of HIC tolerances to long duration events still requiring validation (Newman *et al.*, 1999).

C.2.4.4 Skull fracture criterion

Vander Vorst *et al.* researched linear skull fractures induced via impacts and developed the skull fracture criterion (SFC) as a predictor of skull fracture (Vander Vorst *et al.*, 2003). SFC was defined as the averaged acceleration over the HIC time interval:

$$SFC = \frac{\Delta V_{HIC}}{\Delta T_{HIC}}$$

Where ΔV_{HIC} is the change in velocity over the time interval and ΔT_{HIC} is the time interval (t_2-t_1) that maximizes the integral of the HIC calculation.

The SFC was developed to relate the cranial bone tensile strain with skull fracture; however, Vander Vorst *et al.* found that the strain was difficult to measure and historical test data was not available. To overcome this they constructed a database of skull fracture outcomes of PMHS[5] and correlated these with corresponding Hybrid III headform drop test risk factors. Vander Vorst *et al.* used FE modelling of a simple spherical head model in frontal impact scenarios in order to produce tensile skull strains which could be correlated to SFC (Vander Vorst *et al.*, 2003). This experiment was expanded to also include lateral impacts and varying shapes of impactor with the aim of creating a generalized linear skull fracture criteria (Vander Vorst *et al.*, 2004; Chan *et al.*, 2007). Vander Vorst *et al.* found that the headform change in velocity over the time period used in the HIC calculation (exact period not specified by the authors) was the best correlate to skull fracture. The study primarily focused on flat impact surface as the cylindrical impact surface failed to provide a good correlation between SFC and strain values.

For a 50% probability of skull fracture an SFC value of 155 *g* was proposed, whilst for a 15% risk of skull fracture the criterion is SFC < 120 *g.*

[5] PMHS – post mortem human subject

C.2.5 Rotation-only based head injury criteria

C.2.5.1 Peak rotational acceleration and velocity thresholds

As discussed earlier not all head injuries can be attributed to purely linear impacts, the prevalence of concussive type injures suggests that rotational kinematics of the brain during impact are also responsible for head injury.

However, injuries such as concussion still remain prevalent and it is widely thought that may be the result of the rotational kinematics experienced during impact. The theory that rotational motions attribute to brain injury is not a new one, Holbourn suggested over 75 years ago that head impacts can be analysed in linear and rotational acceleration vectors (Holbourn, 1943).

Severe head injuries were produced in monkey subjects throughout the 1960s (Ommaya *et al.*, 1967). Ommaya *et al.* proposed a scaling strategy for converting the monkey injury tolerances to a concussion threshold for humans. One of the key assumptions used in the scaling process, among several other key limitations and caveats, was that there was geometric similarity between the brains of a series of subhuman primates and the human brain. Despite these limitations, the result of the work by Ommaya *et al.* was to suggest that the cerebral concussion tolerance of 40,000 rad/s^2 observed with rhesus monkeys in sagittal plane rotations equated to 7,500 rad/s^2 for a human.

Gennarelli *et al.* also carried out experiments on monkeys, with the aim of applying purely rotational accelerations to their heads (although there is some debate as to whether pure rotational accelerations were applied) creating injuries ranging from concussions to death (Adams *et al.*, 1981; Gennarelli *et al.*, 1981a). It was found that the severity and type of injury sustained was influenced by the direction of rotation (Gennarelli *et al.*, 1982).

In order to achieve critical brain rotational velocities and considerable displacements between the brain and skull, Löwenhielm demonstrated that rotational accelerations must be applied for a substantial amount of time (Löwenhielm, 1978). Limits of 5000 rad/s^2 for rotational acceleration and 40 rad/s for rotational velocity were proposed by The European Cooperation in Science and Technology (Chinn *et al.*, 2001). It was suggested that cerebral concussion could be caused by rotational velocities in excess of 30 rad/s and rotational accelerations of lower than 1700 rad/s^2 (Ommaya, 1984a).

In more recent times the magnitudes of rotational acceleration and velocity required to induce varying levels of concussion and DAI were estimated (Gennarelli *et al.*, 2003). These values were obtained using the following relationships for rotational acceleration and velocity:

$$\alpha = 2877.8 * AIS$$

$$\omega = 25 * AIS$$

Peng *et al.* further predicted a 50% probability of head injury for resultant rotational acceleration and velocity threshold values (Peng *et al.*, 2012). These threshold values are presented in (Table C.5).

Table C.5: Rotational acceleration and velocity diffuse brain injury severity thresholds

AIS Level	Injury Severity	Rotational Acceleration (rad/s^2)	Rotational Velocity (rad/s)	Reference
1	Mild cerebral concussion	2877.8	25	(Gennarelli *et al.*, 2003)
2	Classical cerebral concussion	5755.6	50	(Gennarelli *et al.*, 2003)
2+	50% head injury risk	11,368.0	50	(Peng *et al.*, 2012)
3	Severe cerebral concussion	8633.4	75	(Gennarelli *et al.*, 2003)
3+	50% head injury risk	18,775.0	55	(Peng *et al.*, 2012)
4	Mild diffuse axonal injury	11,511.2	100	(Gennarelli *et al.*, 2003)
5	Moderate diffuse axonal injury	14,389	125	(Gennarelli *et al.*, 2003)
6	Severe diffuse axonal injury	17,266.8	150	(Gennarelli *et al.*, 2003)

Recently, Zhang *et al.* proposed head injury tolerance levels for temporary brain injuries experienced during impacts lasting between 10-30 ms comprising combined linear and rotational accelerations using advanced FE head modelling. Linear acceleration tolerance levels were observed to be 85 g, whilst rotational acceleration tolerances were 6,000 rad/s^2 (Zhang *et al.*, 2004).

From the body of literature reviewed it seems clear that one should expect a relationship between rotational head accelerations and velocities and haemorrhages within the head, as well as further damage to the substance of the brain (whether that is bleeding or axonal disruption). Testing with rats, Stemper *et al.* identified that increasing the magnitude of rotational acceleration produced longer periods of unconsciousness, which were used to assess acute injury severity (Stemper *et al.*, 2015). Stemper *et al.* further determined that longer duration rotational accelerations produced changes in the 'emotionality' of the rats, as measured using the Elevated Plus Maze assessment (Stemper *et al.*, 2015). This suggests that it is also important to monitor the pulse duration as well as the magnitude of the rotational acceleration if neurological sequelae are to be investigated as well as the acute injury severity.

A summary of brain injury thresholds for rotational acceleration and velocity, at particular pulse durations, is presented in Table C.6. it is important to note that rotational motion was induced in the sagittal plane in the majority of these studies.

Table C.6: Rotational acceleration and velocity thresholds for different injury mechanisms and severities

Injury	Thresholds	Reference
Brain Surface Shearing	α = 2,000–3,000 rad/s^2	(Advani *et al.*, 1982)
Bridging Vein Rupture	α = 4,500 rad/s^2 or ω = 50-70 rad/s	(Löwenhielm, 1974a; Löwenhielm, 1975)
	α = 5,000 rad/s^2 or ω = 50 rad/s	(Löwenhielm, 1978)
	α = 10,000 rad/s^2 for t < 10 ms	(Monea *et al.*, 2014)
Concussion	50% probability: α = 1,800 rad/s^2 for t < 20 ms	(Ommaya *et al.*, 1967; Ommaya and Hirsch, 1971)
	50% probability: ω = 30 rad/s for t ≥ 20 ms	"
	99% probability: α > 7,500 rad/s^2 for t > 6.5 ms	"
	99% probability: α = 14,000 rad/s^2 for t = 11 ms	(Unterharnscheidt, 1969)
	99% probability: α = 13,000 rad/s^2 for t = 11 ms	(Ono *et al.*, 1980)
	99% probability: α = 20,000 rad/s^2 for t = 18 ms	(Gennarelli and Thibault, 1982)
	99% probability: α = 13.6-16 krad/s^2 and ω = 25-48 rad/s	(Pincemaille *et al.*, 1989)
	99% probability: α = 18,000 rad/s^2 for t = 18 ms	(Thibault *et al.*, 1990)
	50% probability: α = 6,200 rad/s^2	(Newman and Shewchenko, 2000)
	50% probability: α = 6,322 rad/s^2	(Newman *et al.*, 2000)
	95% probability: α = 9,267 rad/s^2	(Newman *et al.*, 2000)
	α = 6,400 rad/s^2 and ω = 35 rad/s	(Viano *et al.*, 2005)
	α = 6,200 rad/s^2	(Fijalkowski *et al.*, 2006)
	α = 7,600 rad/s^2 for t = 15 ms	(Fijalkowski *et al.*, 2007)
	α = 7,300 rad/s^2 for t = 23 ms	"
	α = 1,800 rad/s^2	(Kleiven, 2007a)
	α = 6,432 rad/s^2	(Pellman *et al.*, 2003)
	α = 5,022 rad/s^2	(Rowson *et al.*, 2012)
	α = 5,582.3 rad/s^2	(Broglio *et al.*, 2010)
	50% probability: α = 1,747 rad/s^2	(McIntosh *et al.*, 2014)
DAI	α = 20,000 rad/s^2 for t = 18 ms	(Gennarelli and Thibault, 1982)
	α = 19,000 rad/s^2 for t = 20 ms	(Gennarelli *et al.*, 1987)
	α = 10,000 rad/s^2	(Gennarelli and Thibault, 1989)
	ω = 100 rad/s	(Margulies and Thibault, 1992)
	α = 18,000 rad/s^2	(Ommaya *et al.*, 2002)

Injury	Thresholds	Reference
	α = 8,000 rad/s^2 or ω = 70 rad/s	(Kleiven, 2007a)
	α = 10,000 rad/s^2 for t > 4 ms or ω = 19 rad/s	(Davidsson *et al.*, 2009)
	α = 16,000 rad/s^2	(Ommaya *et al.*, 1967)
Mild DAI	α = 12,500–15,500 rad/s^2	(Ommaya *et al.*, 2002)
SDH	α = 32,000 rad/s^2 for t = 14 ms	(Gennarelli and Thibault, 1982)
	α = 10,000 rad/s^2	(Yoganandan *et al.*, 2005)
	α = 10,000 rad/s^2 for t > 10 ms	(Depreitere *et al.*, 2006)
	α = 4,500 rad/s2	(Löwenhielm, 1974b)
TBI	α = 1,700 rad/s^2 or ω = 60-70 rad/s	(Ewing *et al.*, 1975)
	AIS0: α < 4,500 rad/s^2 or ω < 30 rad/s	(Ommaya, 1984b)
	AIS2+: α > 1,700 rad/s^2 or ω > 30 rad/s	"
	AIS3+: α > 3,000 rad/s^2 or ω > 30 rad/s	"
	AIS4+: α > 3,900 rad/s^2 or ω > 30 rad/s	"
	AIS5+: α > 4,500 rad/s^2 or ω > 30 rad/s	"
	α = 25,000 rad/s^2 for short durations	(Tarriere, 1987)
	α > 5,000 rad/s^2	(Thomson *et al.*, 2001)
	4,500 < α < 5,000 rad/s^2 and ω = 60 rad/s	(Shuaeib *et al.*, 2002)
mTBI	25% probability: α = 4,384 rad/s^2	(King *et al.*, 2003)
	50% probability: α = 5,757 rad/s^2	"
	75% probability: α = 7,130 rad/s^2	"
	α = 3,000-4,000 rad/s^2	(Willinger and Baumgartner, 2003b)
	α = 6,000 rad/s^2 for 10 < t < 30 ms	(Zhang *et al.*, 2004)
	25% probability: α = 4,600 rad/s^2	"
	50% probability: α = 5,900 rad/s^2	"
	80% probability: α = 7,900 rad/s^2	"
	α = 1,800 rad/s^2	(Ommaya *et al.*, 1967)
	α = 8,020 rad/s^2	(Frechede and McIntosh, 2009)
Head Injury	50% probability AIS2+: α = 11,368 rad/s^2 and ω = 40 rad/s	(Peng *et al.*, 2012)
	50% probability AIS3+: α = 18,775 rad/s^2 and ω = 55 rad/s	"

C.2.5.2 Rotational injury criterion

The rotational injury criterion (RIC) substitutes the resultant linear acceleration of the HIC equation for the resultant rotational acceleration (Kimpara and Iwamoto, 2012). Kimpara and Iwamoto proposed that a 50% probability of mTBI can be represented with a RIC value of 1.03×10^7 based on head impact data from NFL players who suffered concussive injuries. The RIC equation is shown as below:

$$RIC = \left\{ \left[\left(\frac{1}{t_2 - t_1} \right) \int_{t_1}^{t_2} \alpha(t) \cdot dt \right]^{2.5} (t_2 - t_1) \right\}_{max}$$

where $\alpha(t)$ is the resultant rotational head acceleration, t_1 and t_2 are the initial and final times of the interval during which RIC attains a maximum value and the interval t2-t1 is the boundary of the time duration intervals that define the RIC analysis, typically 36 ms.

C.2.5.3 Power rotational head injury criterion

The power rotational head injury criterion (PRHIC) is calculated as the integrated power of rotational head motion (Kimpara and Iwamoto, 2012). The six degrees of freedom kinematics at the centre of gravity of the head are used in order to predict head injuries associated with angular head accelerations (Kimpara *et al.*, 2011). Kimpara and Iwamoto proposed a 50% probability of mTBI can be represented with a PRHIC threshold value of 8.70×10^5 (Kimpara and Iwamoto, 2012). The expression of this criterion is provided by the below equation:

$$PRHIC = \left\{ \left[\left(\frac{1}{t_2 - t_1} \right) \int_{t_1}^{t_2} \left(I_{xx}\alpha_x \int_{t_1}^{t_2} \alpha_x(t) + I_{yy}\alpha_y \int_{t_1}^{t_2} \alpha_y(t) + I_{zz}\alpha_z \int_{t_1}^{t_2} \alpha_z(t) \right) \cdot dt \right]^{2.5} (t_2 - t_1) \right\}_{max}$$

where x, y and z correspond to coronal, sagittal, horizontal axes for rotational acceleration, $\alpha_{x,y,z}(t)$ is the rotational head acceleration about each axis, I_{xx}, I_{yy} and I_{zz} represent the appropriate mass moments of inertia of the human head about each axis and $[I_{xx}, I_{yy}, I_{zz}]$ = [0.016, 0.024, 0.022] kg.m^2, t_1 and t_2 are the initial and final times of the interval during which PRHIC attains a maximum value and the interval t_2-t_1 is the boundary of the time duration intervals that define the PRHIC analysis, typically 36 ms.

C.2.5.4 Brain rotational Injury Criterion

The Brain rotational Injury Criterion (BrIC) was proposed by (Takhounts *et al.*, 2008). 50th percentile male dummies were impacted to generate head kinematic data from crash events (Takhounts *et al.*, 2013). They then used the outputs from the head instrumentation of the dummies as inputs to the SIMon finite element model to produce a Cumulative Strain Damage Measure (CSDM) value for each test. Critical values were developed for each test to ensure that a BrIC value of 1 corresponds to a 30% probability of DAI (or an AIS4+ injury) occurring. This criterion is expressed in the following equation:

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2}$$

where x, y and z correspond to coronal, sagittal, horizontal axes for rotational acceleration, $\omega_{x,y,z}$ is the peak rotational head acceleration about each axis, W_{xC}, W_{yC} and W_{zC} represent the corresponding critical values determined from frontal dummy impacts and [W_{xC}, W_{yC}, W_{zC}] = [66.2, 59.1, 44.2] rad/s.

C.2.6 Combined rotational and translational head injury criteria

In real world situations it is highly unlikely that a head injury is sustained just by purely linear or rotational accelerations, in fact it is largely recognised that brain injuries are caused by a combination of both. In order to accurately predict the response of brain tissue in an impact it is important that the kinematics of the impact event are represented in three-dimensions.

C.2.6.1 Generalized acceleration model for brain injury threshold

The first injury criterion that attempted to combine both linear and rotational responses into one injury criterion was proposed by Newman as a generalized acceleration model for brain injury threshold (GAMBIT) (Newman, 1986). The validity of this criterion was tested against all known head injury databases in which three-dimensional kinematics were reported. Assuming that linear and rotational accelerations equally and independently contribute to head injury, the GAMBIT expression was calculated to be:

$$G(t) = \left[\left(\frac{a(t)}{a_C}\right)^n + \left(\frac{\alpha(t)}{\alpha_C}\right)^m\right]^{1/s}$$

where a(t) and α(t) refer to the instantaneous values of linear and rotational accelerations and expressed respectively in *g* and rad/s^2, n, m and s are empirical constants and a_C and α_C represent critical tolerance levels for those accelerations.

Whilst many variations of the GAMBIT equation as well as various values for the constants and critical tolerance levels have been presented by several researchers (Table C.7), GAMBIT was never validated extensively as an injury criterion (Newman, 1986),(Chinn *et al.*, 2001),(Newman and Shewchenko, 2000),(Mellor and St Clair, 2005). Kramer proposed that a 50% probability of an irreversible head injury can be represented with a GAMBIT of value 1. Other thresholds proposed throughout the literature are shown in Table C.8 (Kramer, 1998).

Table C.7: GAMBIT constants and critical tolerances (Fernandes and deSousa, 2015)

n	m	s	a_C (g)	α_C (rad/s^2)	Reference
2	2	2	250	25,000	(Newman, 1986)
2	2	2	250	10,000	(Chinn *et al.*, 2001; Mellor and St Clair, 2005)

Table C.8: GAMBIT thresholds (Fernandes and deSousa, 2015)

Injury	Thresholds	Reference
Head Injury	50% probability of AIS4+: G = 1	(Newman *et al.*, 2000)
	50% probability of AIS4+: G = 1.5-2	(Chinn *et al.*, 2001)
Concussion	50% probability: G ≥ 0.4	(Newman *et al.*, 2000)
	95% probability: G ≥ 0.56	"

C.2.6.2 Head impact power

Newman *et al.* deduced that head injuries could be assessed using the rate of change of the translational and rotational kinetic energy (Newman and Shewchenko, 2000). Based on a general expression for this kinetic energy function, but setting appropriate coefficients based on the mass and mass moments of inertia for each axis of the human head, Newman *et al.* developed the Head Impact Power (HIP) injury criterion:

$$HIP = \left\{\left(m \cdot a_x \int a_x \cdot dt\right) + \left(m \cdot a_y \int a_y \cdot dt\right) + \left(m \cdot a_z \int a_z \cdot dt\right) + \left(I_{xx} \cdot \alpha_x \int \alpha_x \cdot dt\right) + \left(I_{yy} \cdot \alpha_y \int \alpha_y \cdot dt\right) + \left(I_{zz} \cdot \alpha_z \int \alpha_z \cdot dt\right)\right\}_{max}$$

where x, y and z correspond to anterior, left and superior for linear acceleration, x, y and z correspond to coronal, sagittal, horizontal axes for rotational acceleration, $a_{x,y,z}(t)$ is the linear head acceleration along each axis, $\alpha_{x,y,z}(t)$ is the rotational head acceleration about each axis, m represents the mass of the human head (4.5 kg) and I_{xx}, I_{yy} and I_{zz} represent the appropriate mass moments of inertia of the human head about each axis and $[I_{xx}, I_{yy}, I_{zz}]$ = [0.016, 0.024, 0.022] kg.m^2.

Newman *et al.* continued to discuss whether the different tolerance to power absorption of the head in different directions needed inclusion (Newman *et al.*, 2000). They suggested that the HIP would need to include additional coefficients reflecting directional tolerances. An evaluation of American football head impact cases in their mTBI database supported these assertions regarding directional tolerances (Newman *et al.*, 1999; Newman *et al.*, 2000; Newman and Shewchenko, 2000). It also seemed to support the conclusion that the maximum HIP appeared to correlate better than existing head injury assessment functions (i.e. HIC) with the mild traumatic brain injury data (Newman *et al.*, 2000). They therefore proposed the maximum head impact power HIP as a new head injury assessment function, though they did not propose the directional tolerance coefficients and further validation at higher impact severities was still required. Furthermore, the data upon which HIP was based had limitations and approximations were made with regard to the impact vectors; the error was calculated to be in excess of 20%.

A variety of proposed injury thresholds associated with HIP are presented in Table C.9.

Table C.9: Head impact power (HIP) injury thresholds (Fernandes and deSousa, 2015)

Injury	Thresholds	Reference
Skull Fracture	50% probability: HIP = 38 kW	(Marjoux *et al.*, 2008)
SDH	50% probability: HIP = 55 kW	"
Moderate Neurological Injury	50% probability: HIP = 24 kW	"
Severe Neurological Injury	50% probability: HIP = 48 kW	"
Concussion	50% probability: HIP = 12.8 kW	(Newman *et al.*, 2000)
	95% probability: HIP = 20.88 kW	"

C.2.6.3 Combined probability

A study by Rowson and Duma (2011) introduced a new injury metric which considered both linear and rotational head acceleration. The metric was derived using a multivariate logistic regression analysis of American football head impact data obtained using the Head Impact Telemetry System (HITS) for instrumenting helmets. The dataset consisted of peak linear and rotational accelerations for 62,974 sub-concussive events and 37 impacts where concussion was diagnosed.

Based on this data, and making an adjustment to account for underreporting of concussion injuries a generalised linear model calculating the combined probability (CP) of concussion was defined by the following equation:

$$CP = \frac{1}{1 + e^{-(\beta_0 + \beta_1 \cdot a + \beta_2 \cdot \alpha + \beta_3 \cdot a \cdot \alpha)}}$$

Where β_0, β_1, β_2 and β_3 are regression coefficients and $[\beta_0, \beta_1, \beta_2, \beta_3] = [-10.2, 0.0433, 0.000873, -0.000000920]$, a is the peak relative linear acceleration, α is the peak relative rotational acceleration and CP is the combined probability of concussion.

This study has two key limitations. First, the analysis is specific to the type of impact mode (football helmet impacts) and neither dataset included impacts predominantly comprising of rotational accelerations. Second, the underreporting of concussion injuries, and the adjustment of the HITS database to account for this, may also have affected the analysis. Unreported concussions may result in conservative estimates of specificity, where the true value of the false positive rates would be lower. These datasets were, however, the best human subject datasets available at the time for analysing the biomechanics of concussion.

C.2.6.4 Principal component score

The Principal Component Score (PCS) is an empirically weighted summation of linear and rotational accelerations, HIC and GSI to calculate the probability of concussion (Greenwald *et al.*, 2008).

$$PCS = 10 \cdot \left(\left(0.4336 \cdot s(a) + 0.2164 \cdot s(\alpha) + 0.4742 \cdot s(HIC) + 0.4718 \cdot s(GSI)\right) + 2\right)$$

where s(X) is a standardised value defined as s(X) = (X-x̅)/σ (x̅ is the population mean, σ is the population standard deviation), a is the peak relative linear acceleration, α is the peak relative rotational acceleration, HIC is the head injury criterion and GSI is the Gadd severity index. Furthermore, impact location weighting coefficients were derived based on the 99th percentile PCS for each location bin. These coefficients were 1.00, 0.95, 0.62, and 0.48 for impacts to the side, front, back and top of the head, respectively.

Key limitations include only using a dataset including only 17 concussion events, of which all were used in training the logistic regression analysis, the use of instrumentation that assumed the rotation of the head occurred in two-dimensions only and the potential for the underreporting of concussion injuries.

C.2.6.5 Brain injury threshold surface

The brain injury threshold surface (BITS) is a global head injury criterion, with associated injury thresholds, that accounts for the time-dependent, combined, linear and rotational kinematics of the head (Antona-Makoshi *et al.*, 2016). A generic BITS equation was developed to define a global three-dimensional injury risk iso-surface based on three second-order variables; linear head accelerations, rotational head accelerations and the duration of the acceleration pulse. This approach combines scaled data from animal experiments and finite element model reconstructions to establish and evaluate the effect of the variables on the positive predictive value of the BITS equation in classifying injurious and non-injurious brain trauma. Injury was defined both as the occurrence of injury during the experiment and as a maximum principle strain level of >23% during the FE modelling. The BITS injury criterion is defined by the below equation, with a BITS value of >1 indicating the risk of a brain injury:

$$BITS = \left(\frac{a}{c_1}\right)^2 + \left(\frac{\alpha}{c_2}\right)^2 - \left(\frac{c_3}{\Delta T}\right)^2$$

Where c_1, c_2 and c_3 are regression coefficients, a is the peak resultant linear acceleration, α is the peak resultant rotational acceleration and ΔT is the duration of the acceleration pulse. For classifying the injury occurrence [c_1, c_2, c_3] = [550 g, 30,000 rad/s^2, 7 ms], whilst for classifying maximum principle strain [c_1, c_2, c_3] = [450 g, 35,000 rad/s^2, 5 ms].

The key limitation of this approach is that the BITS injury criteria cannot yet be applied to humans as the surface has not yet been calculated for, or scaled to represent, that for the human brain.

C.2.7 Stress and strain based head injury criteria

In recent times it seems apparent that the more popular predictors of head injury are largely based on responses at head tissue level rather than on kinematics even though brain injuries have been correlated well with stress, strain and strain rate (Lee and Haut, 1989; Viano and Lovsund, 1999). These parameters can be challenging to measure in practice but can this be achieved through the use of accurate, highly detailed FEM of the head and brain (Van Den Bosch, 2006). Injury parameters can be computed using simulated stresses and strains and several injury predictors have been suggested based on the use of FEM.

C.2.7.1 Intracranial pressure

This head injury criterion is based on calculations of pressure within the brain. Thresholds for this predictor, shown in Table C.10., have been published in many studies.

It had been demonstrated through the use of FE modelling that for impacts with a very short duration intracranial pressure has a superior sensitivity to HIC (Liu and Fan, 1998). Despite this, intracranial pressure was found to poorly correlate with some brain injuries; in particular the prediction of diffuse axonal injuries (Kang *et al.*, 1997; Miller *et al.*, 1998).

Table C.10: Intracranial pressure injury thresholds (Fernandes and deSousa, 2015)

Injury	Intracranial Pressure (kPa)	Reference
Moderate	172.3	(Nahum *et al.*, 1977)
Severe or Fatal	234.4	"
Minor or Absent	≤ 173	(Ward and Chan, 1980)
Severe	≥ 235	"
Brain Injuries	200	(Willinger *et al.*, 1999; Baumgartner, 2001; Raul *et al.*, 2006)
Coup Pressure Brain Injury	180	(Yao *et al.*, 2006)
	256	(Yao *et al.*, 2008)

C.2.7.2 Brain von Mises stress

Brain von Mises stress is based on the theory that the principal cause of brain damage is the stresses (in shear, tension and compression) experienced by the brain. Some of the proposed thresholds are given below in Table C.11.

Table C.11: von Mises stress injury thresholds (Fernandes and deSousa, 2015)

Injury	Von Mises Stress (kPa)	Reference
Brain Injury	12	(Yao *et al.*, 2006)
	14.8	(Yao *et al.*, 2008)
Severe Brain Injury	11-16.5	(Kang *et al.*, 1997)
	27	(Anderson, 2000)
Concussion	22	(Baumgartner, 2001)
	20	(Willinger *et al.*, 2000)
	40	(Deck *et al.*, 2003)
Long Duration Concussion	20	(Chinn *et al.*, 2001)
Short Duration Concussion	10	"
Severe Neurological Injury	46	(Baumgartner *et al.*, 2001)
50% Moderate Neurological Injury	18	(Willinger and Baumgartner, 2003b; Willinger and Baumgartner, 2003a)
50% Severe Neurological Injury	38	"
50% Mild DAI	26	(Deck and Willinger, 2008)
50% Severe DAI	33	"
50% Probability of Concussion	8.4 in corpus callosum	(Kleiven, 2007b)
Severe and Irreversible TBI	14.8 ± 4.5	(Yao *et al.*, 2008)

C.2.7.3 Strain

In mathematics there tend to be two ways of expressing strain: in natural (or Eulerian) terms it is the instantaneous change in length divided by the instantaneous length; however, the often more familiar expression is known as the Lagrangian strain, and it is the difference between the current and original length, divided by the original length. Morrison *et al.* (2003) applied mechanical injuries to organotypic hippocampal slice cultures and quantified the resultant cell death. They concluded that:

- Cell injury is dependent on the magnitude and rate of application of tissue deformation
- Mechanical deformations ≤ 0.1 Lagrangian strain are not injurious when applied at strain rates between 5 and 50 s^{-1}
- Mechanical deformations ≥ 0.2 Lagrangian strain induce significant levels of cell injury, noting that the time course for the damage was dependent on the strain rate of the applied deformation.

C.2.7.4 Strain rate

Concussive events that occurred during US American football (National Football League) games were quantified and duplicated in the laboratory using helmeted dummies. Linear and rotational accelerations measured during the reconstructions were used as inputs into the finite element Wayne State University Head Injury Model (WSUHIM) (Zhang *et al.*, 2003). A variety of brain response parameters were computed for both the concussed and non-concussed players. A total of 53 cases were studied of which there were 22 cases of concussion, as diagnosed by the on-site physician. Strain rate was manually calculated by differentiating the maximum principal strain versus time curves for elements with the highest values of strain. The rate varied from 23 to 140 s^{-1} with an average value of 84 s^{-1} for injury cases and from 11- 67 s^{-1} with an average value of 38 s^{-1} for non-injurious cases. The product of strain and strain rate was also suggested as a local tissue response measure that could be a mechanical criterion for neurological injury. Based on values from three significance tests, the product of strain and strain rate at the midbrain region provided the strongest correlation with mild traumatic brain injury in the WSUHIM (Zhang *et al.*, 2003). Strain rate was also a good injury predictor in this model.

C.2.7.5 Strain correlated with both accelerations

Kleiven and Von Holst found that changes in rotational velocity corresponded best to changes in intracranial strains, whilst HIC and HIP had the best correlation with the strain levels during linear impacts (Kleiven and von Holst, 2003). From this Aare *et al.* developed a criterion correlating linear and rotational accelerations with strains in the brain tissue using the following expression:

$$\varepsilon = k_1 \cdot \Delta\omega + k_2 \cdot HIC$$

where ε is the maximum strain component in the brain tissue, $\Delta\omega$ is the maximum change in rotational velocity, and k_1 and k_2 are constants obtained by regression analysis for each impact condition (Aare *et al.*, 2004).

Aare *et al.* investigated three head impact conditions. Impact 1 was to the vertex inducing sagittal plane rotation, impact 2 was to the lateral aspects of the head inducing axial plane rotation and, finally, impact 3 was to the lateral aspects of the head inducing coronal plane rotation. The k-constants derived for these impact conditions are presented in the Table C.12 below.

Table C.12: Brain tissue strain correlation regression constants (Aare *et al.*, 2004)

Constant	Impact 1	Impact 2	Impact 3
k_1	$6.14*10^{-3}$	$7.26*10^{-3}$	$3.92*10^{-3}$
k_2	$1.32*10^{-5}$	$3.50*10^{-5}$	$7.41*10^{-5}$

Although no injury threshold values were proposed by Aare *et al.*, there are a number of studies in the literature where strain injury thresholds are proposed for the brain. These are presented in Table C.13.

Table C.13: Strain injury thresholds (Fernandes and deSousa, 2015)

Injury	Thresholds	Reference
Brain Tissue Damage	0.5	(Prange and Margulies, 2002; Franceschini *et al.*, 2006)
	0.15	(Thibault *et al.*, 1990)
Cerebral Contusions	50% risk: 0.19	(Shreiber *et al.*, 1997)
DAI	0.18-0.21	(Bain and Meaney, 2000; Morrison *et al.*, 2003)
	0.18	(Wright and Ramesh, 2012)
	0.2	(Kleiven, 2007a)
	Moderate to severe: 0.05-0.1	(Margulies and Thibault, 1992)
	Maximum principle strain: 0.25	(Takhounts *et al.*, 2008)
	50% probability of mild: 0.31	(Deck and Willinger, 2008)
	50% probability of severe: 0.4	"
	50% probability in corpus callosum: 0.21	(Kleiven, 2007b)
	50% probability in grey matter: 0.26	"
	0.1	(Thibault, 1993)
mTBI and Concussion	0.35	(King *et al.*, 2003; Zhang *et al.*, 2003; Viano *et al.*, 2005; Zhang *et al.*, 2008)
	AIS1 concussion: 0.3	(Zhang *et al.*, 2008)
	AIS2 concussion: 0.35	"
	Concussion: 0.1	(Kleiven, 2007a)
	Concussion: 0.1	(Thibault, 1993)
	mTBI: 0.1	(Kimpara and Iwamoto, 2012)
	50% probability in thalamus: 0.13	(Patton *et al.*, 2013)
	50% probability in corpus callosum: 0.15	"
	50% probability in white matter: 0.26	"

C.2.7.6 SIMon injury criteria

The Simulated Injury Monitor (SIMon) FE head model developed by Takhounts *et al.* (2003) is based on three component-level injury metrics developed by DiMasi *et al.* (1995) and Bandak (1995); (1997). The cumulative strain damage measure criteria (CSDM), the dilatation damage measure (DDM) and the relative motion damage measure (RMDM) are representative of general brain injuries; (DAI, contusions and SDH respectively) (Takhounts *et al.*, 2008).

C.2.7.7 Cumulative strain damage measure

The CSDM can be used to evaluate brain damage which is strain-related (Bandak and Eppinger, 1994). This measure was based on an association between DAI and the cumulative volume of brain matter experiencing tensile strains exceeding a critical threshold at some point during an impact event. The measure therefore calculates the volume of model elements that experienced a tensile strain above a prescribed threshold value for each time increment and gives a maximum cumulative value after the event.

To select the critical values of strain and volume for the CSDM injury metric, data from animal experiments were used to relate the CSDM levels to the observed occurrence of DAI. Mild DAI and moderate DAI were found to correspond to CSDM levels of 5 and 22 respectively, meaning that a critical level of strain (15% from Thibault et al. (1990)) was exceed in 5% and 22% of the brain tissue volume (Zhang et al., 2007). Further work has shown that a ≥15% strain level experienced by 55% of brain tissue volume was linked to a 50% probability of concussion (Takhounts et al., 2003). Further suggested values of brain strain critical levels are presented in Table C.13.

C.2.7.8 Dilatation damage measure

The dilation damage measure (DDM) is an intracranial pressure-wave based injury criterion, which evaluates the potential for brain contusions caused by large dilatational stresses (Bandak, 1997). This measure assesses regions where dilatational pressure-waves cause stress states in the brain model that lead to large negative pressures. Specified negative pressure levels experienced by the brain are monitored through the cumulative volume fraction readings. The DDM criterion assesses the damage fraction caused by contusions and brain tissue damage (as may be found in contre-coup injuries). The DDM is calculated by measuring the volume of the elements at each time step which are experiencing a negative pressure level lower than the set threshold value. Suggested threshold pressures are presented in Table C.14.

Table C.14: DDM pressure thresholds

DDM (%)	Threshold (kPa)	Injury risk	Reference
5.0	-101	-	(Zhang *et al.*, 2007)
7.2	-101	50% probability of contusion	(Takhounts *et al.*, 2003)
-	-130	contre-coup Injury	(Yao *et al.*, 2006)
-	-152	contre-coup Injury	(Yao *et al.*, 2008)
-	-186	contre-coup Injury	(Ward and Chan, 1980)

C.2.7.9 Relative motion damage measure

The relative motion damage measure (RMDM) is a correlate for acute SDH and can be used to assess tangential motion of the brain surface resulting from linear and rotational accelerations (Bandak, 1997). The RMDM is thought to be an appropriate predictor of SDH as long as the brain-skull interface is modelled accurately (Marjoux et al., 2008; Takhounts et al., 2008). SDH are caused by the rupture of the bridging veins within the brain (Marjoux et al., 2008). Table C.15. shows the limits associated with the rupture of these veins.

Table C.15: RMDM pressure thresholds

Injury metric	Strain	Reference
Ultimate strain in tension	0.5	(Lee and Haut, 1989)
Failure strain	0.2-1	(Löwenhielm, 1974b)
	0.3-0.6	(Monson *et al.*, 2003; Morrison *et al.*, 2003)
	1	(Takhounts *et al.*, 2003)
Occurrence of SDH	5mm elongation, 25% stretch limit	(Monea *et al.*, 2014)

C.2.7.10 The Strasbourg University finite element head model criteria

The Strasbourg University finite element head model (SUFEHM) performs state-of-the-art FE Analysis to biomechanically model the head during impact. The software uses the 6 degree-of-freedom (6DoF) motion of a head (rotational and linear accelerations) in order to provide an injury risk assessment for three types of head injury; SDH and DAI and skull fracture (Willinger and Baumgartner, 2003a). SUFEHM outputs the brain Von Mises stresses, Cerebrospinal fluid and deformable skull strain energies to give the percentage risk for neurological, SDH and skull fracture injuries respectively. The tolerance limits set out for the model are presented in Table C.16.

Table C.16: SUFEHM tolerance limits

Metric	Injury	Value	Reference
Maximum von Mises stress	Moderate DAI	27 kPa	(Marjoux *et al.*, 2008)
		28 kPa	(Deck *et al.*, 2007; Deck and Willinger, 2008)
	Mild DAI	39 kPa	(Marjoux *et al.*, 2008)
		53 kPa	(Deck *et al.*, 2007; Deck and Willinger, 2008)
Brain von Mises strain	Moderate DAI	30%	"
	Mild DAI	57%	"
Brain first principal strain	Moderate DAI	33%	"
	Mild DAI	67%	"
Maximum global internal strain energy	SDH	4211 mJ	(Marjoux *et al.*, 2008)
		4950 mJ	(Deck *et al.*, 2007)
	Skull fracture	833 mJ	(Marjoux *et al.*, 2008)
		544 mJ	(Sahoo *et al.*, 2013)
		448 mJ	(Sahoo *et al.*, 2014b)

SUFEHM has been found to provide reliable results and has been validated to provide axonal elongations as well as skull damage (Sahoo *et al.*, 2013; Sahoo *et al.*, 2014a). Other high-quality FE head models are available, such as the Wayne State University brain injury model, the KTH FE human Head Model and the Dartmouth subject-specific FE human head model (Zhang *et al.*, 2004; Kleiven, 2007b; McAllister *et al.*, 2012).

C.3 References

Aare M (2003). *Prevention of Head Injuries-focusing Specifically on Oblique Impacts.* Farkost och flyg.

Aare M, Kleiven S and Halldin P (2004). *Injury tolerances for oblique impact helmet testing.* International Journal of Crashworthiness 9(1). 15-23.

Adams J, Graham D and Gennarelli T (1981). *Acceleration induced head injury in the monkey. II. Neuropathology*. In Experimental and Clinical Neuropathology. Springer.

Advani S, Powell W, Huston J and Ojala S (1975). *Human head impact response—experimental data and analytical simulations.* Proc. Intern. Conf. on Biomechanics of Serious Trauma. pp.153-62.

Advani S, Ommaya A, Yang W, COLLINS R, Kivity K, Kalliers D, Mathern R, Schmidt G, Mohan D and Kingn D (1982). *Human Body Dynamics: Impact, Occupational and Athletic Aspects*. Human Body Dynamics: Impact, Occupational and Athletic Aspects.

Al-Bsharat AS, Hardy WN, Yang KH, Khalil TB, Tashman S and King AI (1999). *Brain/skull relative displacement magnitude due to blunt head impact: new experimental data and model.* SAE Technical Paper.

Alliance (1999). *Dummy response limits for FMVSS 208 compliance testing. Docket No. NHTSA-99-6407.*

Allsop DL, Warner CY, Wille MG, Schneider DC and Nahum AM (1988). *Facial impact response—a comparison of the Hybrid III dummy and human cadaver.* SAE Transactions. 1224-40.

Allsop DL, Perl TR and Warner CY (1991). *Force/deflection and fracture characteristics of the temporo-parietal region of the human head.* SAE Transactions. 2009-18.

Alves W, Macciocchi SN and Barth JT (1993). *Postconcussive symptoms after uncomplicated mild head injury.* The Journal of Head Trauma Rehabilitation.

Anderson RWG (2000). *A study on the biomechanics of axonal injury.*

Antona-Makoshi J, Davidsson J, Ejima S and Ono K (2016). *Development of a comprehensive injury criterion for moderate and mild traumatic brain injuries.* International journal of automotive engineering 7(2). 69-75.

Bain AC and Meaney DF (2000). *Tissue-level thresholds for axonal damage in an experimental model of central nervous system white matter injury.* Journal of biomechanical engineering 122(6). 615-22.

Bandak F (1995). *On the mechanics of impact neurotrauma: a review and critical synthesis.* Journal of neurotrauma 12(4). 635-49.

Bandak F (1997). *Impact traumatic brain injury: a mechanical perspective.* Neurotraumatology–biomechanic aspects, cytologic and molecular mechanisms. Schmidt-Römhild, Lübeck. 59-83.

Bandak FA and Eppinger RH (1994). *A three-dimensional finite element analysis of the human brain under combined rotational and translational accelerations.* SAE Transactions. 1708-26.

Baumgartner D (2001). *Mécanismes de lésion et limites de tolérance au choc de la tête humaine—Reconstruction numérique et expérimentale de traumatismes crâniens.* Ph. D. dissertation, Université Louis Pasteur Strasbourg I, France.

Baumgartner D, Willinger Rm, Shewchenko N and Beusenberg M (2001). *Tolerance limits for mild traumatic brain injury derived from numerical head impact replication.* Proceedings of the 2001 International IRCOBI Conference on the Biomechanics of Impacts. pp.353-55.

Bradshaw D, Ivarsson J, Morfey C and Viano DC (2001). *Simulation of acute subdural hematoma and diffuse axonal injury in coronal head impact.* Journal of Biomechanics 34(1). 85-94.

Broglio SP, Schnebel B, Sosnoff JJ, Shin S, Feng X, He X and Zimmerman J (2010). *The biomechanical properties of concussions in high school football.* Medicine and science in sports and exercise 42(11). 2064.

Brolinson PG, Manoogian S, McNeely D, Goforth M, Greenwald R and Duma S (2006). *Analysis of linear head accelerations from collegiate football impacts.* Current sports medicine reports 5(1). 23-28.

Brooks JC, Strauss DJ, Shavelle RM, Paculdo DR, Hammond FM and Harrison-Felix CL (2013). *Long-term disability and survival in traumatic brain injury: results from the National Institute on Disability and Rehabilitation Research Model Systems.* Archives of physical medicine and rehabilitation 94(11). 2203-09.

Brooks JC, Shavelle RM, Strauss DJ, Hammond FM and Harrison-Felix CL (2015). *Long-term survival after traumatic brain injury Part II: life expectancy.* Archives of physical medicine and rehabilitation 96(6). 1000-05.

Chan P, Lu Z, Rigby P, Takhounts E, Zhang J, Yoganandan N and Pintar F (2007). *Development of generalized linear skull fracture criterion.* 20th International Technical Conference on the Enchanced Safety of Vehicles, Lyon, France.

Chee CP and Ali H (1991). *Basal skull fractures. A prospective study of 100 consecutive admissions.* Australian and New Zealand Journal of Surgery 61(8). 597-602.

Chinn B, Canaple B, Derler S, Doyle D, Otte D, Schuller E and Willinger R (2001). *COST 327 Motorcycle safety helmets, Final Report of the Action, European Commission.* URL: http://ec. europa. eu/transport/roadsafety_library/publications/cost327_final_report. pdf (diakses 21 Juni 2010).

Davidsson J, Angeria M and Risling M (2009). *Injury threshold for sagittal plane rotational induced diffuse axonal injuries.* Proceedings of the International Research Conference on the Biomechanics of Impact (IRCOBI). pp.43-56.

Deck C, Willinger R and Baumgartner D (2003). *Helmet optimisation based on head-helmet modelling.* WIT Transactions on The Built Environment 67.

Deck C, Baumgartner D and Willinger R (2007). *Head injury prediction tool for protective systems optimisation.* Infats In: Proceedings of the 5th International Forum of Automotive Traffic Safety.

Deck C and Willinger Rm (2008). *Improved head injury criteria based on head FE model.* International Journal of Crashworthiness 13(6). 667-78.

Depreitere B, Van Lierde C, Sloten JV, Van Audekercke R, Van Der Perre G, Plets C and Goffin J (2006). *Mechanics of acute subdural hematomas resulting from bridging vein rupture.* Journal of Neurosurgery 104(6). 950-56.

DeWitt DS, Perez-Polo R, Hulsebosch CE, Dash PK and Robertson CS (2013). *Challenges in the development of rodent models of mild traumatic brain injury.* Journal of neurotrauma 30(9). 688-701.

DiMasi FP, Eppinger RH and Bandak FA (1995). *Computational analysis of head impact response under car crash loadings.* 0148-7191. SAE Technical Paper.

Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, Greenwald RM, Chu JJ and Crisco JJ (2005). *Analysis of real-time head accelerations in collegiate football players.* Clinical journal of sport medicine 15(1). 3-8.

Eppinger R, Sun E, Bandak F, Haffner M, Khaewpong N, Maltese M, Kuppa S, Nguyen T, Takhounts E and Tannous R (1999). *Development of improved injury criteria for the assessment of advanced automotive restraint systems–II.* National Highway Traffic Safety Administration. 1-70.

Ewing CL, Thomas D, Lustick L, Becker E and Williams G (1975). *The effect of the initial position of the head and neck on the dynamic response of the human head and neck to-Gx impact acceleration.* NAVAL AEROSPACE MEDICAL RESEARCH LAB DETACHMENT MICHOUD NEW ORLEANS LA.

Fernandes FA and deSousa RJA (2015). *Head injury predictors in sports trauma–A state-of-the-art review.* Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine 229(8). 592-608.

Fijalkowski RJ, Ellingson B, Stemper BD, Yoganandan N, Gennarelli TA and Pintar FA (2006). *Interface parameters of impact-induced mild traumatic brain injury.* Biomedical sciences instrumentation 42. 108-13.

Fijalkowski RJ, Stemper BD, Pintar FA, Yoganandan N and Gennarelli TA (2007). *Influence of angular acceleration duration on functional outcomes following mild diffuse brain injury.* 2007 IRCOBI conference. pp.161-71.

Forero Rueda MA, Cui L and Gilchrist MD (2011). *Finite element modelling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs.* Computer methods in biomechanics and biomedical engineering 14(12). 1021-31.

Franceschini G, Bigoni D, Regitnig P and Holzapfel GA (2006). *Brain tissue deforms similarly to filled elastomers and follows consolidation theory.* Journal of the Mechanics and Physics of Solids 54(12). 2592-620.

Frechede B and McIntosh AS (2009). *Numerical reconstruction of real-life concussive football impacts.* Medicine and science in sports and exercise 41(2). 390-98.

Gadd CW (1966). *Use of a weighted-impulse criterion for estimating injury hazard.* 0148-7191. SAE Technical Paper.

Gennarelli TA, Adams J and Graham D (1981a). *Acceleration induced head injury in the monkey. I. The model, its mechanical and physiological correlates*. In Experimental and Clinical Neuropathology. Springer.

Gennarelli TA, Ono K, Kikuchi A, Nakamura M, Kobayashi H and Nakamura N (1981b). *Mechanistic approach to the head injuries: Clinical and experimental studies of the important types of injury.* Head and Neck Injury Criteria: A Consensus Workshop, Washington, DC, March 26-27, 1981. US Department of Transportation, National Highway Traffic Safety Administration, pp.20.

Gennarelli TA and Thibault LE (1982). *Biomechanics of acute subdural hematoma.* The Journal of trauma 22(8). 680-86.

Gennarelli TA, Thibault LE, Adams JH, Graham DI, Thompson CJ and Marcincin RP (1982). *Diffuse axonal injury and traumatic coma in the primate.* Annals of Neurology: Official Journal of the American Neurological Association and the Child Neurology Society 12(6). 564-74.

Gennarelli TA, Thibault LE, Tomei G, Wiser R, Graham D and Adams J (1987). *Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration.* SAE Transactions. 1355-59.

Gennarelli TA and Thibault L (1989). *Clinical rationale for a head injury angular acceleration criterion.* Head injury mechanisms: The need for an angular acceleration criterion, Association for the Advancement of Automotive Medicine. 5-8.

Gennarelli TA, Pintar FA and Yoganandan N (2003). *Biomechanical tolerances for diffuse brain injury and a hypothesis for genotypic variability in response to trauma.* Annual Proceedings/Association for the Advancement of Automotive Medicine. Association for the Advancement of Automotive Medicine, pp.624.

Got C, Patel A, Fayon A, Tarriere C and Walfisch G (1978). *Results of experimental head impacts on cadavers: the various data obtained and their relations to some measured physical parameters.* 0148-7191. SAE Technical Paper.

Greenwald RM, Gwin JT, Chu JJ and Crisco JJ (2008). *Head impact severity measures for evaluating mild traumatic brain injury risk exposure.* Neurosurgery 62(4). 789-98.

Gurdjian ES and Lissner H (1961). *Photoelastic confirmation of the presence of shear strains at the craniospinal junction in closed head injury.* Journal of Neurosurgery 18(1). 58-60.

Gurdjian ES, Roberts V and Thomas LM (1966). *Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury.* Journal of Trauma and Acute Care Surgery 6(5). 600-04.

Gurdjian ES (1975). *Re-evaluation of the biomechanics of blunt impact injury of the head.* Surgery, gynecology & obstetrics 140(6). 845-50.

Gurdjian ES (1976). *Cerebral contusions: re-evaluation of the mechanism of their development.* The Journal of trauma 16(1). 35-51.

Guskiewicz KM, Mihalik JP, Shankar V, Marshall SW, Crowell DH, Oliaro SM, Ciocca MF and Hooker DN (2007). *Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion.* Neurosurgery 61(6). 1244-53.

Hernandez F, Wu LC, Yip MC, Laksari K, Hoffman AR, Lopez JR, Grant GA, Kleiven S and Camarillo DB (2015). *Six degree-of-freedom measurements of human mild traumatic brain injury.* Annals of biomedical engineering 43(8). 1918-34.

Hodgson VR and Thomas LM (1972). *Breaking strength of the human skull vs. impact surface curvature.*

Holbourn A (1943). *Mechanics of head injuries.* Lancet 242. 438-41.

Hopes P and Chinn B (1989). *Helmets: A New Look At Design And Possible Protection. Proceedings Of The 1989 International Ircobi Conference On The Biomechanics Of Impacts, Held In Stockholm, 13-15 September 1989.* Publication of: IRCOBI-Secretariat.

Horgan TJ (2005). *A finite element model of the human head for use in the study of pedestrian accidents.* University College Dublin Ireland.

Hoshizaki B, Pos A, Kendall M, Karton C and Brien S (2013a). *The Relationship between Head Impact Characteristics and Brain Trauma.* J Neurol Neurophysio 5(181).

Hoshizaki B, Post A, Kendall M, Karton C and Brien S (2013b). *The relationship between head impact characteristics and brain trauma.* J Neurol Neurophysiol 5(1). 1-8.

Hoshizaki TB and Brien SE (2004). *The science and design of head protection in sport.* Neurosurgery 55(4). 956-67.

Hume A, Mills N and Gilchrist A (1995). *Industrial head injuries and the performance of the helmets.* Proceedings of the international IRCOBI Conference on Biomechanics of Impact, Switzerland.

Hwang K and You SH (2010). *Analysis of facial bone fractures: An 11-year study of 2,094 patients.* Indian journal of plastic surgery: Official publication of the Association of Plastic surgeons of India 43(1). 42.

Jager TE, Weiss HB, Coben JH and Pepe PE (2000). *Traumatic brain injuries evaluated in US emergency departments, 1992-1994.* Academic Emergency Medicine 7(2). 134-40.

Kang H-S, Willinger Rm, Diaw BM and Chinn B (1997). *Validation of a 3D anatomic human head model and replication of head impact in motorcycle accident by finite element modeling.* 0148-7191. SAE Technical Paper.

Kimpara H, Nakahira Y, Iwamoto M, Rowson S and Duma S (2011). *Head injury prediction methods based on 6 degree of freedom head acceleration measurements during impact.* International journal of automotive engineering 2(2). 13-19.

Kimpara H and Iwamoto M (2012). *Mild traumatic brain injury predictors based on angular accelerations during impacts.* Annals of biomedical engineering 40(1). 114-26.

King AI, Yang KH, Zhang L, Hardy W and Viano DC (2003). *Is head injury caused by linear or angular acceleration.* IRCOBI Conference. Lisbon, Portugal, pp.1-12.

King RE, Scianna JM and Petruzzelli GJ (2004). *Mandible fracture patterns: a suburban trauma center experience.* American journal of otolaryngology 25(5). 301-07.

Kleiven S (2003). *Influence of impact direction on the human head in prediction of subdural hematoma.* Journal of neurotrauma 20(4). 365-79.

Kleiven S and von Holst H (2003). *Review and evaluation of head injury criteria.* RTO Specialist Meeting, the NATO's Research and Technology Organization (RTO).

Kleiven S (2007a). *Head injury biomechanics and criteria.* Biomechanics and Neuronics, course litterature, KTH.

Kleiven S (2007b). *Predictors for traumatic brain injuries evaluated through accident reconstructions.* SAE Technical Paper.

Kramer F (1998). *Rechnerische Simulation*. In Passive Sicherheit von Kraftfahrzeugen. Springer.

Langlois JA, Rutland-Brown W and Wald MM (2006). *The epidemiology and impact of traumatic brain injury: a brief overview.* The Journal of Head Trauma Rehabilitation 21(5). 375-78.

Lee M-C and Haut RC (1989). *Insensitivity of tensile failure properties of human bridging veins to strain rate: implications in biomechanics of subdural hematoma.* Journal of Biomechanics 22(6). 537-42.

Liu-Shindo M and Hawkins DB (1989). *Basilar skull fractures in children.* International journal of pediatric otorhinolaryngology 17(2). 109-17.

Liu D and Fan C (1998). *Applied pressure tolerance to evaluate motorcycle helmet design.* Proceedings of IJCRASH 98.

Löwenhielm P (1974a). *Strain tolerance of the Vv. Cerebri Sup.(bridging veins) calculated from head-on collision tests with cadavers.* Zeitschrift für Rechtsmedizin 75(2). 131-44.

Löwenhielm P (1974b). *Dynamic properties of the parasagittal bridging beins.* Zeitschrift für Rechtsmedizin 74(1). 55-62.

Löwenhielm P (1975). *Mathematical simulation of gliding contusions.* Journal of Biomechanics 8(6). 351-56.

Löwenhielm P (1978). *Tolerance level for bridging vein disruption calculated with a mathematical model.* Journal of bioengineering 2(6). 501.

Ma VY, Chan L and Carruthers KJ (2014). *Incidence, prevalence, costs, and impact on disability of common conditions requiring rehabilitation in the United States: stroke, spinal cord injury, traumatic brain injury, multiple sclerosis, osteoarthritis, rheumatoid arthritis, limb loss, and back pain.* Archives of physical medicine and rehabilitation 95(5). 986-95. e1.

Margulies SS and Thibault LE (1992). *A proposed tolerance criterion for diffuse axonal injury in man.* Journal of Biomechanics 25(8). 917-23.

Marjoux D, Baumgartner D, Deck C and Willinger R (2008). *Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria.* Accident Analysis & Prevention 40(3). 1135-48.

Marshall S, Bayley M, McCullagh S, Velikonja D and Berrigan L (2012). *Clinical practice guidelines for mild traumatic brain injury and persistent symptoms.* Canadian Family Physician 58(3). 257-67.

McAllister TW, Ford JC, Ji S, Beckwith JG, Flashman LA, Paulsen K and Greenwald RM (2012). *Maximum principal strain and strain rate associated with concussion diagnosis correlates with changes in corpus callosum white matter indices.* Annals of biomedical engineering 40(1). 127-40.

McElhaney JH, Fogle JL, Melvin JW, Haynes RR, Roberts VL and Alem NM (1970). *Mechanical properties of cranial bone.* Journal of Biomechanics 3(5). 495-96.

McIntosh A, Patton D, Frechede B, Pierre P, Ferry E and Barthels T (2014). *The biomechanics of concussion in unhelmeted football players in Australia: A case-control study.* Br Med J 4(5). 1-9.

McKee AC, Gavett BE, Stern RA, Nowinski CJ, Cantu RC, Kowall NW, Perl DP, Hedley-Whyte ET, Price B and Sullivan C (2010). *TDP-43 proteinopathy and motor neuron disease in chronic traumatic encephalopathy.* Journal of Neuropathology & Experimental Neurology 69(9). 918-29.

Meaney DF and Smith DH (2011). *Biomechanics of concussion.* Clinics in sports medicine 30(1). 19-31.

Mellor A and St Clair V (2005). *Advanced motorcycle helmets.* Proceedings: International Technical Conference on the Enhanced Safety of Vehicles. National Highway Traffic Safety Administration, pp.13p-13p.

Melvin JW, McElhaney J and Roberts V (1970). *Development of a mechanical model of the human head-determination of tissue properties and synthetic substitute materials.* 0148-7191. SAE Technical Paper.

Mertz HJ, Prasad P and Irwin AL (1997). *Injury risk curves for children and adults in frontal and rear collisions.* 0148-7191. SAE Technical Paper.

Miller RT, Margulies SS, Leoni M, Nonaka M, Chen X, Smith DH and Meaney DF (1998). *Finite element modeling approaches for predicting injury in an experimental model of severe diffuse axonal injury.* 0148-7191. SAE Technical Paper.

Monea AG, Van der Perre G, Baeck K, Delye H, Verschueren P, Forausebergher E, Van Lierde C, Verpoest I, Vander Sloten J and Goffin J (2014). *The relation between mechanical impact parameters and most frequent bicycle related head injuries.* Journal of the mechanical behavior of biomedical materials 33. 3-15.

Monson KL, Goldsmith W, Barbaro NM and Manley GT (2003). *Axial mechanical properties of fresh human cerebral blood vessels.* Journal of biomechanical engineering 125(2). 288-94.

Morrison B, Cater HL, Wang CC, Thomas FC, Hung CT, Ateshian GA and Sundstrom LE (2003). *A tissue level tolerance criterion for living brain developed with an in vitro model of traumatic mechanical loading.* SAE Technical Paper.

Nahum AM, Gatts JD, Gadd CW and Danforth J (1968). *Impact tolerance of the skull and face.* 0148-7191. SAE Technical Paper.

Nahum AM, Smith R and Ward CC (1977). *Intracranial pressure dynamics during head impact.* 0148-7191. SAE Technical Paper.

Nahum AM and Melvin JW (2012). *Accidental injury: biomechanics and prevention*. Springer Science & Business Media, Accidental injury: biomechanics and prevention.

Newman J, Beusenberg M, Fournier E, Shewchenko N, Withnall C, King A, Yang K, Zhang L, McElhaney J and Thibault L (1999). *A new biomechanical assessment of mild traumatic brain injury: Part 1—Methodology.* Proceedings of the International Research Conference on the Biomechanics of Impacts (IRCOBI). pp.17-36.

Newman J, Barr C, Beusenberg MC, Fournier E, Shewchenko N, Welbourne E and Withnall C (2000). *A new biomechanical assessment of mild traumatic brain injury. Part 2: results and conclusions.* Proceedings of the International Research Council on the Biomechanics of Injury conference. International Research Council on Biomechanics of Injury.

Newman JA (1986). *A generalized acceleration model for brain injury threshold (GAMBIT).* Proceedings of International IRCOBI Conference, 1986.

Newman JA and Shewchenko N (2000). *A proposed new biomechanical head injury assessment function-the maximum power index.* SAE Technical Paper.

NICE (2014). *Head Injury: Triage, Assessment, Investigation and Early Management of Head Injury in Children, Young People and Adults.*

Nusholtz G, Kaiker P and Gould W (1987). *Two factors critical in the pressure response of the impacted head.* Aviation, space, and environmental medicine 58(12). 1157-64.

Oeur A, Karton C, Post A, Rousseau P, Hoshizaki T, Marshall S and Cusimano M (2013). *An examination of headform dynamic response for concussive and traumatic brain injuries.* Helmet Performance and Design.

Ommaya A, Hirsch AE, Yarnell P and Harris EH (1967). *Scaling of experimental data on cerebral concussion in sub-human primates to concussion threshold for man.* David W Taylor Naval Ship Research and Development Centre.

Ommaya A and Hirsch A (1971). *Tolerances for cerebral concussion from head impact and whiplash in primates.* Journal of Biomechanics 4(1). 13-21.

Ommaya A (1984a). *Head injury biomechanics. In: George G. Snively Memorial Lecture.* Denver: AAAM, .

Ommaya A (1984b). *The head: kinematics and brain injury mechanisms.* Elsevier: Amsterdam.

Ommaya A, Goldsmith W and Thibault L (2002). *Biomechanics and neuropathology of adult and paediatric head injury.* British journal of neurosurgery 16(3). 220-42.

Ono K, Kikuchi A, Nakamura M, Kobayashi H and Nakamura N (1980). *Human Head Tolerance to Sagittal Impact—Reliable Estimation Deduced from Experimental Head Injury Using Subhuman Primates and Human Cadaver Skulls.* SAE Transactions. 3837-66.

Pappachan B and Alexander M (2006). *Correlating facial fractures and cranial injuries.* Journal of Oral and Maxillofacial Surgery 64(7). 1023-29.

Patton DA, McIntosh AS and Kleiven S (2013). *The biomechanical determinants of concussion: finite element simulations to investigate brain tissue deformations during sporting impacts to the unprotected head.* Journal of applied biomechanics 29(6). 721-30.

Pellman EJ, Viano DC, Tucker AM, Casson IR and Waeckerle JF (2003). *Concussion in professional football: reconstruction of game impacts and injuries.* Neurosurgery 53(4). 799-814.

Peng Y, Deck C, Yang J, Otte D and Willinger Rm (2012). *A study of kinematics of adult pedestrian and head impact conditions in case of passenger car collisions based on real world accident data.* International Research Council on the Biomechanics of Injury (IRCOBI) Conference. pp.766-78.

Pincemaille Y, Trosseille X, Mack P, Tarriere C, Breton F and Renault B (1989). *Some new data related to human tolerance obtained from volunteer boxers.* SAE Transactions. 1752-65.

Post A and Hoshizaki T (2012a). *Mechanical properties describing brain impact injuries: a review.* Trauma 14(4). 327-49.

Post A and Hoshizaki TB (2012b). *Mechanisms of brain impact injuries and their prediction: a review.* Trauma 14(4). 327-49.

Post A, Hoshizaki TB, Gilchrist M and Brien S (2012a). *Analysis of the influence of independent variables used for reconstruction of a traumatic brain injury incident.* Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology 226(3-4). 290-98.

Post A, Oeur A, Hoshizaki T and Gilchrist MD (2012b). *The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research.* Proc of IRCOBI, Dublin, Ireland, Sept. 10-14.

Post A (2013). *The influence of dynamic response characteristics on traumatic brain injury.* University of Ottawa Ottawa, ON, Canada.

Post A, Oeur A, Hoshizaki B and Gilchrist MD (2013). *Examination of the relationship between peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts.* Computer methods in biomechanics and biomedical engineering 16(5). 511-19.

Post A, Hoshizaki TB, Gilchrist MD, Brien S, Cusimano MD and Marshall S (2014). *The influence of dynamic response and brain deformation metrics on the occurrence of subdural hematoma in different regions of the brain.* Journal of Neurosurgery 120(2). 453-61.

Prange MT and Margulies SS (2002). *Regional, directional, and age-dependent properties of the brain undergoing large deformation.* Journal of biomechanical engineering 124(2). 244-52.

Prasad P and Mertz HJ (1985). *The position of the United States delegation to the ISO Working Group 6 on the use of HIC in the automotive environment.* SAE Transactions. 106-16.

Raul J-S, Baumgartner D, Willinger R and Ludes B (2006). *Finite element modelling of human head injuries caused by a fall.* International Journal of Legal Medicine 120(4). 212-18.

Robbins D and Wood J (1969). *Determination of mechanical properties of the bones of the skull.* Exp Mech 9(5). 236–40.

Rowson S and Duma SM (2011). *Development of the STAR evaluation system for football helmets: integrating player head impact exposure and risk of concussion.* Annals of biomedical engineering 39(8). 2130-40.

Rowson S, Duma SM, Beckwith JG, Chu JJ, Greenwald RM, Crisco JJ, Brolinson PG, Duhaime A-C, McAllister TW and Maerlender AC (2012). *Rotational head kinematics in football impacts: an injury risk function for concussion.* Annals of biomedical engineering 40(1). 1-13.

Sahoo D, Deck C, Yoganandan N and Willinger R (2013). *Anisotropic composite human skull model and skull fracture validation against temporo-parietal skull fracture.* Journal of the mechanical behavior of biomedical materials 28. 340-53.

Sahoo D, Deck C and Willinger R (2014a). *Development and validation of an advanced anisotropic visco-hyperelastic human brain FE model.* Journal of the mechanical behavior of biomedical materials 33. 24-42.

Sahoo D, Deck C, Yoganandan N and Willinger R (2014b). *Composite FE human skull model validation and development of skull fracture criteria.* IRCOBI Conference Proceedings.

Schnebel B, Gwin JT, Anderson S and Gatlin R (2007). *In vivo study of head impacts in football: a comparison of National Collegiate Athletic Association Division I versus high school impacts.* Neurosurgery 60(3). 490-96.

Schneider DC and Nahum AM (1972). *Impact studies of facial bones and skull.* 0148-7191. SAE Technical Paper.

Shreiber DI, Bain AC and Meaney DF (1997). *In vivo thresholds for mechanical injury to the blood-brain barrier.* 0148-7191. SAE Technical Paper.

Shuaeib FM, Hamouda A, Umar RR, Hamdan M and Hashmi M (2002). *Motorcycle helmet: Part I. Biomechanics and computational issues.* Journal of Materials Processing Technology 123(3). 406-21.

Stalnaker R, McElhaney J and Roberts V (1971). *MSC tolerance curve for human heads to impact. ASME Paper No.* 71WA/BHF-10 0971.

Stemper BD, Shah AS, Pintar FA, McCrea M, Kurpad SN, Glavaski-Joksimovic A, Olsen C and Budde MD (2015). *Head rotational acceleration characteristics influence behavioral and* *diffusion tensor imaging outcomes following concussion.* Annals of biomedical engineering 43(5). 1071-88.

Takhounts EG, Eppinger RH, Campbell JQ, Tannous RE, Power ED and Shook LS (2003). *On the development of the SIMon finite element head model.* SAE CONFERENCE PROCEEDINGS P. SAE; 1999, pp.107-34.

Takhounts EG, Ridella SA, Hasija V, Tannous RE, Campbell JQ, Malone D, Danelson K, Stitzel J, Rowson S and Duma S (2008). *Investigation of traumatic brain injuries using the next generation of simulated injury monitor (SIMon) finite element head model.* SAE Technical Paper.

Takhounts EG, Craig MJ, Moorhouse K, McFadden J and Hasija V (2013). *Development of brain injury criteria (BrIC).* SAE Technical Paper.

Tarriere C (1987). *'Relationships between experimental measuring techniques and real world injuries.* Report, Head Injury Mechanism Symposium, AAAM/Volvo Car Corporation, New Orleans, Louisiana. pp.47-85.

Thibault L (1993). *Brain injury from the macro to the micro level and back again: what have we learned to date?* Proceedings of the International Research Council on the Biomechanics of Injury conference. International Research Council on Biomechanics of Injury, pp.3-25.

Thibault LE, Gennarelli TA, Margulies SS, Marcus J and Eppinger R (1990). *The strain dependent pathophysiological consequences of inertial loading on central nervous system tissue.* International Conference on the Biomechanics of Impacts, Bron.

Thomas L, Roberts V and Gurdjian E (1966). *Experimental intracranial pressure gradients in the human skull.* Journal of neurology, neurosurgery, and psychiatry 29(5). 404.

Thomson R, Lovsund P, Norin H, Jakobsson L, Bostrom O and Haland Y (2001). *Brain injuries in real world accidents–a multidisciplinary investigation.* Proceedings of the IRCOBI Conference, Isle of Man.

Unterharnscheidt F (1969). *Traumatic leasions of the brain and spinal cord due to non-deforming angular acceleration of the head.* Tex Rep Biol Med 27(1). 127-66.

Van Den Bosch HLA (2006). *Crash helmet testing and design specifications.*

Vander Vorst M, Stuhmiller J, Ho K, Yoganandan N and Pintar F (2003). *Statistically and biomechanically based criterion for impact-induced skull fracture.* Annual Proceedings/Association for the Advancement of Automotive Medicine. Association for the Advancement of Automotive Medicine, pp.363.

Vander Vorst M, Chan P, Zhang J, Yoganandan N and Pintar F (2004). *A new biomechanically-based criterion for lateral skull fracture.* Annual Proceedings/Association for the Advancement of Automotive Medicine. Association for the Advancement of Automotive Medicine, pp.181.

Versace J (1971). *A review of the severity index.* 0148-7191. SAE Technical Paper.

Viano DC and Lovsund P (1999). *Biomechanics of brain and spinal-cord injury: analysis of neuropathologic and neurophysiology experiments.* Traffic Injury Prevention 1(1). 35-43.

Viano DC, Casson IR, Pellman EJ, Zhang L, King AI and Yang KH (2005). *Concussion in professional football: brain responses by finite element analysis: part 9.* Neurosurgery 57(5). 891-916.

Voo L, Pintar FA, Yoganandan N, Sanees Jr A, Ewing CL, Thomas DJ and Snyder RG (1994). *Biomechanical analysis of tractor induced head injury.* SAE Transactions. 178-83.

Ward C and Chan M (1980). *'Rotation Generated Shear Strains in the Brain.* Proceedings, 8th Annual International Workshop on Human Subjects for Biomechanical Research. pp.11-29.

Willinger R, Kang H-S and Diaw B (1999). *Three-dimensional human head finite-element model validation against two experimental impacts.* Annals of biomedical engineering 27(3). 403-10.

Willinger R and Baumgartner D (2003a). *Numerical and physical modelling of the human head under impact-towards new injury criteria.* International journal of vehicle design 32(1-2). 94-115.

Willinger R and Baumgartner D (2003b). *Human head tolerance limits to specific injury mechanisms.* International Journal of Crashworthiness 8(6). 605-17.

Willinger R, Halldin P, Bogerd C, Deck C and Fahlstedt M (2015). *Final report of Working Group 3: Impact engineering. A COST Action TU1101/HOPE collaboration.*

Willinger Rm, Baumgartner D, Chinn B and Neale M (2000). *Head tolerance limits derived from numerical replication of real world accidents.* Proceedings of the International Research Council on the Biomechanics of Injury conference. International Research Council on Biomechanics of Injury, pp.209-21.

Wright RM and Ramesh K (2012). *An axonal strain injury criterion for traumatic brain injury.* Biomechanics and modeling in mechanobiology 11(1-2). 245-60.

Yao J, Yang J and Otte D (2006). *Investigation of brain injuries by reconstructions of real world adult pedestrian accidents.* 2006 IRCOBI conference. pp.241-52.

Yao J, Yang J and Otte D (2008). *Investigation of head injuries by reconstructions of real-world vehicle-versus-adult-pedestrian accidents.* Safety science 46(7). 1103-14.

Yoganandan N, Sances Jr A, Pintar FA, Walsh PR, Ewing CL, Thomas DJ, Snyder RG, Reinartz J and Droese K (1994). *Biomechanical tolerance of the cranium.* SAE Transactions. 184-89.

Yoganandan N, Pintar FA, Zhang J, Gennarelli TA and Beuse N (2005). *Biomechanical aspects of blunt and penentrating head injuries.* IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications. Springer, pp.173-84.

Yoganandan N, Nahum A and Melvin J (2015). *Accidental Injury: Biomechanics and Prevention.* New York: Springer Science & Business Media., Accidental Injury: Biomechanics and Prevention.

Zhang J, Yoganandan N, Pintar FA, Guan Y and Gennarelli TA (2007). *Biomechanical differences between contact and non-contact head impacts in vehicle crash tests.* Proceedings 20th International Conference on ESV. Paper. Citeseer.

Zhang L, Yang KH, King AI and Viano DC (2003). *A new biomechanical predictor for mild traumatic brain injury–A preliminary finding.* Proc. 2003 Summer Bioengineering Conference. pp.137-38.

Zhang L, Yang KH and King AI (2004). *A proposed injury threshold for mild traumatic brain injury.* Journal of biomechanical engineering 126(2). 226-36.

Zhang L, Yang K and Gennarelli TA (2008). *Mathematical modeling of cerebral concussion: Correlations of regional brain strain with clinical symptoms.* Proceedings of Biomechanics of Impact. 123-32.

Development of a New Cycle Helmet Assessment Programme (NCHAP): Literature Review

This literature review focused on three key research themes that underpin the design of evidence-based helmet safety performance testing and assessment protocols. These included an analysis of literature describing the characteristics of real-world cyclist collisions, a critical review of current international cycle helmet testing standards and a review of the state-of-the-art in traumatic brain injury risk criteria.

The cyclist accidentology review analysed a range of research literature, collision databases and cycling ridership statistics to quantify the key characteristics of cyclist collisions. The key demographics of cyclist casualties were defined to quantify age, gender, height and weight. Helmet impact locations, impact angles and injury reduction levels as well as causes of collisions and collision speeds were reported.

The critical appraisal of current cycle helmet testing and assessment standards reviewed and compared seven standards currently in force across the world. The appraisal focussed on summarising the differences between the testing and assessment approaches and the key characteristics of each safety performance requirement. Based on this appraisal, this review proposed several recommendations to further inform the initial development of the NCHAP testing and assessment protocols.

The final literature review provides an overview of the theory underpinning the head injury continuum and summarises the state-of-the-art in head injury criteria. It provides an overview of each head injury criterion and identifies the associated injury risk thresholds to provide a complete overview of all relevant head injury criteria and risk thresholds that may be utilised by the NCHAP protocols.

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