HEAD INJURY AND EFFECTIVE MOTORCYCLE HELMETS

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ABSTRACT

The wearing of a motorcycle helmet certified to an appropriate standard has been the most significant step in reducing fatal and serious injury among motorcyclists worldwide. Motorcycle helmets have been shown to be at least 50% effective in reducing fatal head injury in motorcycle crashes [1]. Most motorcycle helmet standard requirements have remained substantially the same for 40 years, while over the same period our understanding of causes of injury to the brain has been rapidly improving. Current international motorcycle helmet standards are based around a translational acceleration energy attenuation test.

Reconstruction of crash involved motorcycle helmet damage in the COST 327 study [2] demonstrated that the AIS 2+ head injuries in helmeted head impacts are more likely to be due to indirect (or head motion induced) rather than direct impact. Occupants of crashed vehicles have also been observed by Gennarelli [3] to have a shift in the type of brain injury treated in the emergency room. This shift has been related to improvements in vehicle safety, especially the use of airbag technology. The improved protection for vehicle occupants in crashes due to airbag controlled head impacts has led to a decreasing incidence of focal (direct) brain injury accompanied by a relative increase in diffuse (indirect) brain injury. In sporting head injury, King et al. [4] have shown that football and bicycle helmets built to the current test requirements reduce translation acceleration of the head but do not necessarily reduce the rotational acceleration of the head of the wearer in an impact. These recent advances in our knowledge of the effects and causes of traumatic brain injury have yet to be carried over to motorcycle helmet standards.

The crash characteristics and injuries to the head sustained by helmeted motorcyclists were examined by reference to data from motorcycle crash studies including:

- COST 327 [2], which reconstructed the helmet impact for n=226 motorcyclists with AIS 2+ head injuries;
- MAIDS [5], which investigated n=921 injurious European motorcycle crashes; and,
- Gibson and Thai [6], which examined the helmets and injuries of n=175 riders in fatal motorcycle crashes in Australia.

The crash data regarding the head injury sustained in helmeted head impacts in motorcycle crashes suggests areas available to improve current motorcycle helmet effectiveness and motorcycle helmet standard test methodologies in reducing brain injury. This study defines some of these areas where motorcycle helmet effectiveness in preventing brain injury can be improved, including:

- Changes to helmet test methodology to include biofidelic rotational as well as translational head motion effects to be measured;
- Development of accepted test requirements to mitigate rotational brain injury, with initial emphasis on reducing traumatic brain injury TBI; and,
- Improved facial impact protection, without increasing neck injury risk.

INTRODUCTION

“Vulnerable” road users such as motorcyclists are at greater risk than vehicle occupants and usually bear the greatest burden of injury. In Australia, motorcyclists make up 3% of registered vehicles but represent 16 % of road user fatalities and 22 % of serious injuries. Motorcycle riders are also the fastest growing sector of road user in Australia with motorcycle registrations increasing by 56 % between 2005 and 2010 [7].

According to the World Health Organisation [8], head trauma is the main cause of death and morbidity in motorised two wheeler users. Head trauma contributes to around 75% of motorised two-wheeler deaths in European countries and between and 55-88% of motorised two wheeler rider deaths in Malaysia.

Head Injury Mechanisms

Head injuries can be classified under four major groups: scalp damage, skull fractures, extra-cerebral bleeding or haematoma, and brain damage [9].

Skull fractures are mainly due to direct impact and the force levels required to cause fracture have been studied by many researchers. In contrast, brain injuries can result directly from impact to the head or indirectly by the motion of the head, even without impact.

Ommaya et al. [10] demonstrated that abrupt rotation with impact could affect sensory responses through experiments with primates. In addition to concussion, other brain injuries such as acute subdural haematoma (SDH) due to ruptured bridging veins [11] and diffuse axonal injuries (DAI) [12] have been experimentally produced in primates by acceleration of the head without requiring a direct impact to the head.

Ommaya [13] also identified the important role of the "contact phenomenon" in causing skull deformation. The angular accelerations required to produce concussion in human surrogates by direct impact to the head were shown to be approximately half of those required to produce concussion by pure inertial loading of the head.

Clinical trends also provide insights into the mechanisms for different types of brain injuries. Gennarelli [3] observed a shift in the type of brain injury treated in the emergency room due to the improvements in vehicle occupant safety. The introduction of airbags and softer impacts to the head has been accompanied by a decreasing incidence of focal (direct) brain injury and an increase in diffuse brain injury.

Head Impact Tolerance Criteria

The Wayne State Tolerance Curve (WSTC), first presented by Lissner et al. [14], presents a relationship between average anterior-posterior acceleration of the skull measured at the occipital bone in forehead impacts and the pulse duration, see Figure 1. The "curve" included six points obtained from different experiments with embalmed cadaver heads and has been developed with subsequent cadaver, animal and volunteer tests. Only translational accelerations were used in producing the WSTC. Despite much criticism and the shortcomings of the WSTC [9], which include being based on translational acceleration only, it is the basis for most currently accepted head injury criterion, including the Head Injury Criterion (HIC) commonly used in automotive research.

Hirsch and Ommaya [15] reported that rotational motion appeared to be more critical to the production of brain injury than translation motion stating that "no evidence has to this date been presented which relates brain injury and concussion to translational motion of the head for short-duration force inputs, whether through whiplash or direct impact."

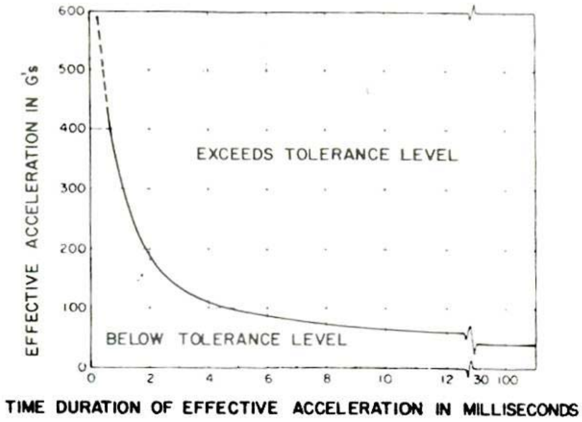


Figure 1. The Wayne State University Concussion Tolerance Curve for linear acceleration, after SAE (1980) [16].

Head injury tolerance to rotational acceleration of the head was investigated by Ommaya [17], who reported that the rotational accelerations necessary to cause concussion and severe diffuse axonal injury (DAI) are 4,500 rad/s^2 and 18,000 rad/s^2 respectively for an adult. Margulies and Thibault [18], using a combination of animal testing and scaling, established tolerance curves for DAI based on peak rotation acceleration and peak change in rotational velocities, see Figure 2.

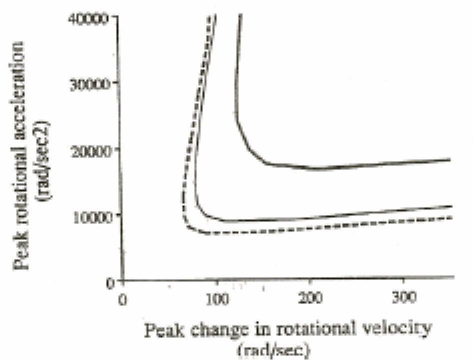


Figure 2. Diffuse axonal injury rotational acceleration and rotational velocity thresholds for infant (500g brain mass, heavy solid line) and adult (1067g brain mass, solid line, 1400g brain mass, dashed line) [18].

Finite element models are increasingly being used as an alternative method for assessing injury risk as they enable investigation of the intracranial response under real world head impact conditions. Deck and Willinger [19] demonstrated that intracranial variables in finite element models demonstrate better correlation with specific injuries than global parameters such as peak linear acceleration and HIC. They reported that intracerebral maximal principal strains, von Mises strains and von Mises stresses are well correlated with both moderate and severe DAI. Similarly, the best correlation with subdural haemorrhage was the minimum pressure within the cerebral spinal fluid.

Motorcycle Helmets

Mandatory motorcycle helmet use is regarded as the single most effective approach for the prevention of traumatic brain injuries among motorcycle users in both developed and developing countries [8]. Motorcycle helmets have been shown to be at least 50% effective in reducing fatal head injury in motorcycle crashes [1].

As explained by van den Bosch [20], *"a motorcycle helmet (to an approved standard) will spread or diffuse any contact impact force and provide for energy absorption beneath that contact point, hence the contact injuries defined by Gennarelli* [skull deformations, coup lesions, epidural haemorrhage] *are those injuries most likely to be prevented - or even excluded - by a motorcycle helmet."* The effect of a helmet on preventing inertial injuries (or indirect injuries due to the motion of the head) is less clear.

The inability of sporting helmets to protect against inertial brain injuries has been demonstrated by other researchers. King et al. [4], for example, demonstrated that American football helmets and bicycle helmets (compliant with current standard test requirements) reduce translational acceleration of the head, but do not necessarily reduce the rotational acceleration of the head in an impact and, in some cases, may increase it.

Performance standards play a large role in the design of helmets. Current international motorcycle helmet standards are based on the WSTC injury criteria and place a limit on the peak linear acceleration and duration of the helmeted headform during an impact (US DOT FMVSS 218, JIS T 8133 and AS 1698 for example) or combined the peak acceleration with a maximum allowable HIC (ECE/UN Regulation 22.05). In the US DOT FMVSS 218, JIS T 8133 and AS 1698 standards the shock absorption test restricts the rotation of the headform by use of a guided drop. On the other hand, the European free flight test allows the headform to rotate, but does not measure or apply limits to the headform rotation.

To assess the protective effectiveness of a helmet in a real (crash based) impact requires the shock absorption test to be a good representation of the actual impact [20]. Figure 3 shows how laboratory (drop) tests attempt to replicate actual crash impacts to correlate the load on the head (form) with the injury. A greater understanding of how the loading to the helmeted head of a motorcyclist in an accident leads to head and brain injury can be used to improve the process for testing the effectiveness of a helmet. However, it must go beyond the deficiencies of the current drop test.

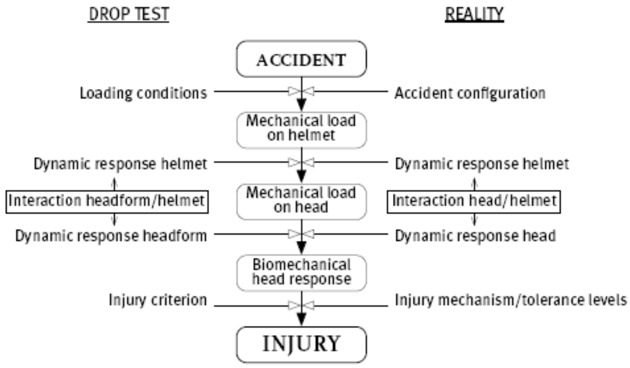


Figure 3 Load-injury scheme for helmeted head impact (van den Bosch [20] modified from Wismans [21])

The first step in this process is to accurately define what happens in real impacts to the helmeted head based on motorcycle crash data.

MOTORCYCLE CRASH DATA

Careful investigation of real world accidents is an integral part of the prevention of injury by the application of biomechanics [22]. In reality, the dynamic helmet and head response (see Figure 3) cannot be directly measured, but crash investigation can indicate the accident configuration and the injury that results.

The MAIDS study [5] of n=921 powered two wheeler accidents in five European countries was carried out by the Association of European Motorcycle Manufacturers. A case control study methodology was used, where data was collected for an additional 923 non-accident involved powered two wheelers. In the crashes, 75% of all powered two wheeler impact speeds were under 50 km/h. When the crash also involved another vehicle, 90% of all other vehicles were to the front of the powered two wheeler rider at impact. The head was the third most injured body region (18.4%) following the lower (31.8%) and upper (24.3%) extremities respectively.

A comprehensive review of the performance of Australian market motorcycle helmets in crashes was performed by Dowdell et al.[23] in NSW, Australia. Cases were included on the basis that the crash was of sufficient severity to have the motorcyclist admitted to hospital and that the motorcyclist was wearing a helmet approved to the current Standard.

200 cases were collected, of which 72 were fatal and 128 non-fatal. More than two thirds of the

impacts to the helmet in these cases were tangential. In the cases where a head or neck injury occurred, 50% of impacts were to the general frontal area of the helmet. Local skull fractures (vault fractures) were associated with impacts adjacent to the fracture site. The authors note that many of the brain injuries were of a type associated with translational or rotational accelerations that are produced by tangential impacts. Brain injuries of this type comprised over 40 percent of the AIS4 injuries. In 42 cases the rider had lost consciousness.

The difficulties which arise when fatal cases are used for motorcycle crash studies are demonstrated by the results of this study [23]. A breakdown of injury severity to the head, neck, face and chest (in Table 1) shows that the fatal cases had a much higher incidence and severity of head and chest injuries.

Table 1.
Comparison of the non-fatal and fatal head, neck, facial and chest injury in motorcycle crashes by severity, based on Dowdell et al. [23].

Body region	Cases with injury	No. of inj.	AIS Injury Severity					
			6	**5**	**4**	**3**	**2**	**1**
Non-fatal cases (n=128)								
Head	58	61	0	0	0	5	46	10
Neck	25	25	0	0	1	1	2	21
Face	15	28	0	0	0	0	14	14
Chest	13	18	0	0	1	7	5	5
Fatal cases (n=72)								
Head	58	143	11	11	43	68	10	0
Neck	16	18	8	2	0	3	3	2
Face	17	25	0	0	0	1	8	16
Chest	62	139	12	17	53	37	18	2

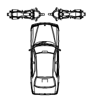
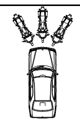
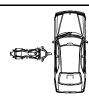
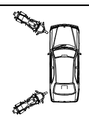
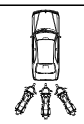
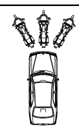
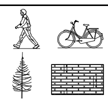
Richter et al. [2] analysed details of 218 European motorcycle accidents which were part of a larger study (COST 327) and examined the head injury mechanisms in these helmeted motorcyclist cases. There were 84 fatalities included, 74 of which suffered fatal head injuries. Of the 205 helmets inspected, there were 196 frontal impacts, including 115 chin bar impacts and 42 impacts to the visor. There were only 2 impacts to the crown. 157 helmets had impacts to the rear and most helmets had lateral impacts. Richter and his co-workers classified the injuries as resulting from either direct force effects or indirect force effects. They found that direct force effects were responsible for a high percentage of skull vault fractures (84.2%), facial fractures (96.3%) and skin injuries (87%), while the majority of brain lesions (96.2%) were the result of acceleration or deceleration forces acting on the head and helmet, i.e. indirect force effects.

A study by Gibson and Thai [6] examined the CASR Head Injury database and abstracted 174, mainly fatal, motorcycle accident cases collected in South Australia between 1983 and 1994. The database included records of the autopsy data (including neuropathology and the incidence of diffuse axonal injury), a helmet inspection and reporting of the crash circumstances. The aim of the study was to investigate basilar skull fracture to helmeted motorcyclists in crashes. The authors reported that 74.7% of cases (n = 174) involved an impact to the helmet or head and almost 50% of the severe impacts to the head were in the facial region. This database was re-analysed in the context of the brain injuries sustained by the fatally injured motorcyclists.

Re-analysis of the CASR Database

The accident types collected in the CASR database are representative of typical motorcyclist crash types. The crash types, from the various studies, are compared in Table 2 based on the classification from the COST 327 study.

Table 2.
Comparison of the COST 327, MAIDS and CASR motorcycle crash type distribution.

Collision Types	Diagram	% (COST 327)	% (MAIDS)	% (CASR)
Type 1		1.8	7.9	7.6
Type 2		8.8	4.1	11.1
Type 3		14.2	20.5	9.0
Type 4		31.0	29.2	25.7
Type 5		5.3	7.2	6.3
Type 6		0	1.7	4.2
Type 7		38.9	29.4	36.1

The CASR fatal motorcycle crash data contains predominantly fatally injured motorcyclists (94%) and so represents only high severity cases. The average estimated impact speed of the motorcycles in the CASR data was approximately 80 km/h. In comparison the average impact speed for COST 327 [2] was 55km/h and for MAIDS [5] 53.6 km/h.

A subset of thirty cases was selected from the CASR database to analyse further, based on the brain injury details being available from the autopsy reports and a helmet inspection being available. For each case, the accident factors and injuries received in the crash were reviewed. The autopsy reports were used to define the injuries. The helmets were visually examined for markings and damage.

The group selected included 27 full face helmets and 3 open face helmets with a total of 40 impacts on the helmets. The distribution of these impacts on the helmets is presented in Figure 4. Further details of the 30 cases are provided in Appendix 1.

The head and brain lesions in the 30 cases were classified as being caused by either direct force effect (DFE) or indirect force effect (IFE), using the same protocol as Richter et al. [2]. These injuries are summarised in Table 3. All coup lesions that were directly caused by a force affecting the damaged structures of the head and brain were defined as DFE, while IFE lesions were all contrecoup lesions and all coup lesions indirectly caused by the effecting force. The lesions were classified by reference to the accident circumstances and the damage to the helmet. In the CASR data, n = 30, the majority of skull vault fracture (77.8%) and facial fracture (100%) were due to direct force effects while most brain lesions (81.3%) were caused by indirect force effects, Table 3. This is similar to general distribution of injury reported by Richter et al. [2].

Table3.

The location and type of the 231 lesions of the head region in the n = 30 CASR fatal motorcycle crashes.

Type of Lesion	Force Effect				Total
	DFE		IFE		
	No.	%	No.	%	
Bone (n = 53)					
Total	**27**	**50.9**	**26**	**49.1**	**53**
Vault	7	77.8	2	22.2	9
Base	0	0	24	100	24
Zygoma	4	100	0	0	4
Orbital	4	100	0	0	4
Nasal	4	100	0	0	4
Maxilla	3	100	0	0	3
Mandible	5	100	0	0	5
Brain (n = 134)					
Total	**25**	**18.7**	**109**	**81.3**	**134**
EDH	0	0	1	100	1
SDH	0	0	9	100	9
SAH	7	25.9	20	74.1	27
Inter ventricular haemorrhage	5	38.4	8	61.6	13
DAI	0	0	6	100	6
Contusion	5	29.4	12	70.6	17
Laceration	8	40	12	60	20
Multi petechial haemorrhage	0	0	19	100	19
Brain Stem	0	0	22	100	22
Skin (n = 44)					
Total	**35**	**79.5**	**9**	**20.5**	**44**
Scalp	13	61.9	8	38.1	21
Face	22	95.7	1	4.3	23
Total	**87**	**37.7**	**144**	**62.3**	**231**

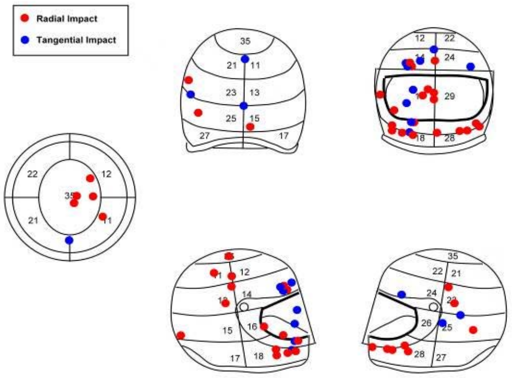


Figure 4. Distribution and type of head impacts in the n=30 fatal motorcyclists cases selected from the CASR Head Injury Study Database.

DISCUSSION

The motorcycle crash investigation studies reviewed here [2, 5, 6, 23] have a similar general distribution of the type of crashes, see Table 2. The crash data defines the accident circumstances and the injuries received by the motorcyclists involved in the crashes. The initial analysis of these available data sources indicates several areas where the effectiveness of motorcycle helmets in powered two wheeler crashes may be improved. The relatively low incidence of skull vault fractures as a result of direct impact on the helmet indicates one area where the current helmet standards are working. The following areas of the current helmets are indicated as worthy of further investigation of possible improvement:

- A high proportion of brain injury results from indirect force effects;
- Both tangential and radial impacts appear to play a part in the causation of indirect brain injuries; and,
- A relatively high incidence of facial fractures and brain injury are the result of direct impacts to the face.

Analytical tools are now available, in the form of human head and neck finite element models which are sufficiently developed to predict brain injuries [20, 28]. Such models allow analysis of the biomechanical response of the head and brain to various types of real crash head impact scenarios and the effect of the helmet on this response (see Figures 3 & 5).

INVESTIGATION METHODOLOGY FOR THE STUDY

The energy absorption "drop" test for helmets is over simplified and a poor representation of a real head impact in a crash. This study will use an advanced dummy, THOR (Test Device for Human Occupant Restraint), combined with a finite element model of the head and neck able to predict the brain injuries which occur in real impacts. The biofidelic THOR head and neck will allow generation of the mechanical loading to the helmet and head from controlled impact tests, while the FE head and neck model will predict the resulting injuries. A flow chart of the methodology is outlined in Figure 5, based upon a modified version of the Wismans biomechanical injury model [21]. A similar methodology has been previously suggested by Deck and Willinger [19].

The test methodology has notable differences to the standard helmet drop test, including human like skull deformation and biofidelic neck responses. The correct neck response is important for reconstructing the correct trajectory of the head after impact and has been found to be necessary for accurately recreating the impacts to COST 327 motorcyclists, American football players and in motorsport (FIA) cases (summarised in [19]).

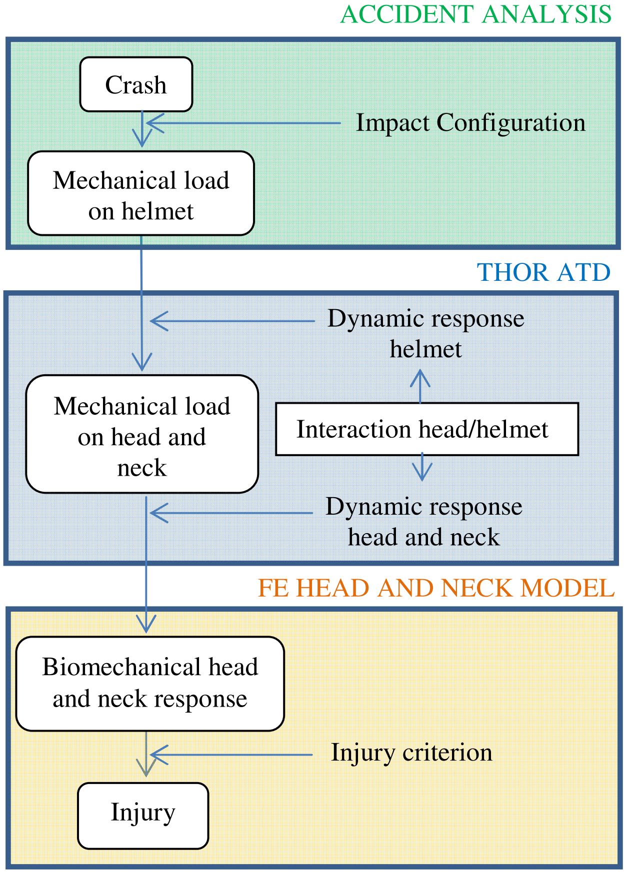


Figure 5 The investigation methodology for the project demonstrating the use of the THOR dummy head and neck and the finite element model.

Impact Testing

An Anthropomorphic Test Device (ATD) or dummy is usually used in vehicle crash testing to predict the injuries sustained by a living person in typical crash circumstances. Such a device must be biofidelic in anthropometry and the response. To assess the protection offered by a vehicle to an occupant in a regulatory frontal crash, biomechanical response data is measured on a Hybrid III crash test dummy. The THOR dummy is an advanced impact dummy, under development by the US National Highway Traffic Safety Administration (NHTSA) since 1993. It is based on more recent and improved biomechanical knowledge than the Hybrid III.

The THOR dummy, to be used in this study, is pictured in a helmeted forehead impact in Figure 6. The THOR head and neck are used in this study for the following reasons:

1. The THOR headform can be instrumented with a 9 accelerometer array (3-2-2-2) for measurement of translational and angular accelerations of the head in 6 axes as a result of impact.
2. A head skin is available with the chin area suitable for wearing a motorcycle helmet.
3. The THOR face has been developed to have human like response to facial impact [24] and is able to be fitted with a complement of force transducers.
4. Finally, the THOR neck has a more biofidelic response than that of the Hybrid III, with improved head lag response and lower stiffness.

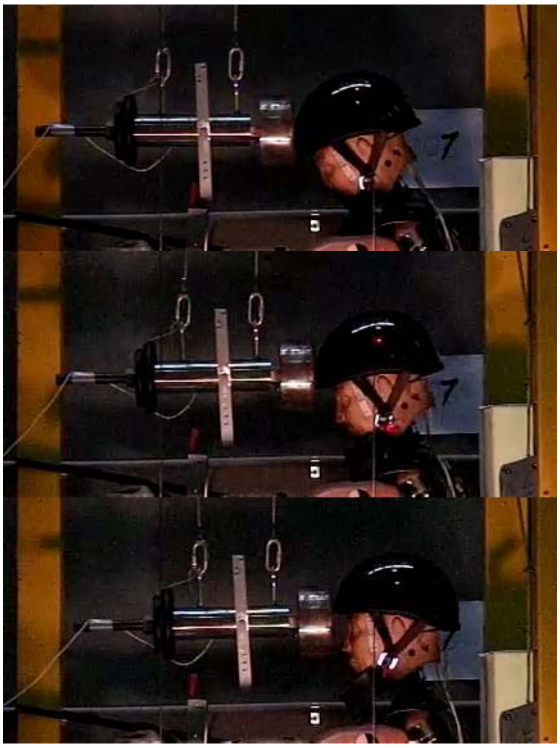


Figure 6 A helmeted THOR head and neck responding to a pendulum impact to the head.

Finite-Element Modelling

Three-dimensional finite element models of the human head have been increasingly used for assessing head injury risk since an early model was developed in 1975 [25]. The development of such finite element models has reached a point which now allows investigation into the intracranial response resulting from an impact to the head.

Injury tolerance limits for intracranial response variables have been proposed and demonstrated for a number of finite element models. As examples, Takhounts and Eppinger [26] suggested a 50% risk of DAI at 55% cumulative strain and a 50% risk of contusion at 7.2% of dilatational damage measure using the SIMon FE model, and the 'Universite Louis Pasteur' (ULP) human head FE model predicts a 50% chance of SDH at cerebral spinal fluid strain energy of 4211 mJ.

This study will use the head and neck components of the finite element H-Model developed by the ESI Group, Figure 7. The model of the human head [27] consists of 48,870 elements, 16 material types and includes the skull (inner and outer table, upper and lower dipole, face bone, mandible), dura, sinus, venous blood, pia, cerebral spinal fluid (CSF), white matter, grey matter, falx cerebri, tentorium, ventricle, cerebellum and brain stem. Contact interfaces are defined between related parts.

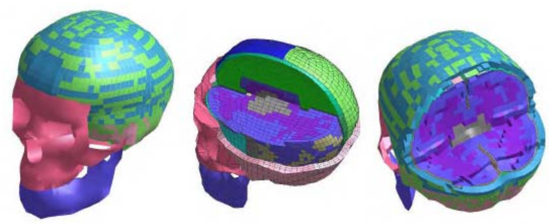


Figure 7 The ESI H-Head finite-element model.

The ESI H-head model has been validated by Xin and Zaouk [28] against the 3D brain motion data of Hardy et al. [29] and the intracranial pressure data of Nahum et al. [30]. The authors subsequently used the ESI H-head model for investigating TBI resulting from blast over-pressure and blast-related impacts. The researchers used the Cumulative Strain Damage Measure (CSDM) and the Dilatation Damage Measure (DDM) to quantify the risk of injury. For this study, the other injury mechanisms defined with the ULP FE model [31] will be investigated on the H-head. These are Von Mises stresses for neurological lesions, global strain in the CSF layer for subdural haemorrhage and local strain energy in the skull for skull fracture.

Discussion

The investigation methodology outlined here provides a realistic approach to testing the effectiveness of a motorcycle helmet in preventing injury to the head, face and brain. Motorcyclist crash data provides a means of real crash validation of the experimental and numerical models, which are themselves independently validated against cadaver, animal and volunteer studies.

The THOR represents one of the most biofidelic mechanical head/neck complexes available. It will allow investigation of a wide range of impact types including the effect of the helmet. The finite element modelling permits prediction of the risk of specific injury determined by the response of the head to the impact, such as skull fracture, subdural haemorrhage and diffuse axonal injury.

SUMMARY

The wearing of a motorcycle helmet certified to an appropriate standard has been the most significant step in reducing fatal and serious injury among motorcyclists worldwide. The following areas of current helmet performance in crashes are worthy of further investigation for possible improvements:

- A high proportion of brain injury results from indirect force effects;
- Both tangential and radial impacts appear to play a part in the causation of indirect brain injuries; and,
- A relatively high incidence of facial fractures and brain injury are the result of direct impacts to the face.

Most helmet standard requirements have remained substantially the same for 40 years, while over the same period our understanding of the mechanisms of brain injury has been rapidly improving. The following suggestions reflect areas available to improve current motorcycle helmet effectiveness and motorcycle helmet standard test methodologies:

- Include measurement of biofidelic rotational as well as translational head motion effects in standard test methodologies;
- Development of accepted test requirements to mitigate rotational brain injury, with initial emphasis on reducing traumatic brain injury TBI; and,
- Improved facial impact protection, without increasing neck injury risk including development of test methods to the facial area.

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APPENDIX 1 – Summary of 30 cases studied in detail from the CASR Database.

Case	Head impact	Region of head impacted	Impacted Object	Direct Brain Injury	Indirect Brain Injury	Skull Vault Fracture	Basilar Skull Fracture
1	Radial	Forehead/ facial	Car roof edge	Y	Y	Y	Y
2	Radial	Right chin bar	Truck	N	Y	Y	Y
3	Radial	Crown	Edge truck tray	Y	Y	Y	Y
4	Tangential	Facial	Road surface/car	N	Y	N	Y
5	Radial	Left chin bar/right frontoparietal	X member behind bumper	N	Y	N	Y
6	Crushing and/or radial	Right chin bar/ right temporo-parietal	Car wheels	Y	Y	N	Y
7	Radial	Right mid facial	Pylon cross-brace	Y	Y	N	Y
8	Radial	Right chin bar/face	Road surface/car	N	N	N	Y
9	Radial	Right chin bar/face	Kerb or road	N	Y	N	Y
10	Radial	Crown	Utility pole	N	Y	N	Y
11	Radial/ Tangential	Forehead/ occipital	Tree/Road surface	N	Y	N	Y
12	Radial	Facial	Car/road surface	N	N	N	Y
13	Radial	Facial	Car	N	Y	N	Y
14	Tangential	Sun Visor, Left Temporal	Truck wheels/ underside	N	Y	N	Y
15	Radial	Left occipital/ chin bar	Utility pole	N	Y	N	Y
16	Radial and Tangential	Right frontal	Car	N	Y	N	Y
17	Tangential	Rear parieto-occipital	Tree	N	Y	N	Y
18	Radial	Right occipital	Truck/road	N	Y	Y	Y
19	Radial	Right facial	Helmet	Y	Y	Y	Y
20	Radial	Crown	Utility pole	Y	Y	Y	Y
21	Radial	Facial/chin bar	Edge truck tray	N	Y		Y
22	Radial	Crown/chin bar	Armco rail/ road	Y	Y	Y	Y
23	Radial	Crown/chin bar	Utility pole	Y	Y	Y	Y
24	No Evidence of Impact			N	Y	N	Y
25	Tangential	Frontal/Facial	Road	N	Y	Y	Y
26	Radial and Tangential	Right frontal/ facial	Utility pole	N	Y	N	N
27	Tangential	Left temporal/ Right chin bar	Road	N	Y	N	N
28	Tangential	Left temporal	Road surface	N	Y	N	N
29	Tangential	Frontal/facial	Road surface	N	Y	N	N
30	Radial	Left/Right temporo-parietal	Van/Road Surface	N	N	N	N