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Technical Response to the Unpublished Paper 'Critical Evaluation of the SHARP Motorcycle Helmet Rating' by NJ Mills

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Transport Research Laboratory

PUBLISHED PROJECT REPORT PPR452

Technical Response to the Unpublished Paper 'Critical Evaluation of the SHARP Motorcycle Helmet Rating' by NJ Mills

by B Chinn and D Hynd (TRL)

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Client:	**Department for Transport, Transport Technology and Standards**
	(Bernie Frost)

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Executive Summary

The SHARP motorcycle helmet safety rating scheme was launched by the Department for Transport (DfT) in June 2008 to provide motorcyclists with objective information on the impact protection offered by motorcycle helmets in the event of an accident. The rating scheme is based on considerable previous research regarding head injury mechanisms, motorcycle accident investigations, motorcyclist head impact accident reconstructions, and the development of an advanced helmet that demonstrated the potential for considerable improvement in the protection offered by helmets.

Recently, an unpublished paper (Critical Evaluation of the SHARP Motorcycle Helmet Rating, by NJ Mills) has criticised the approach taken by SHARP to the rating of helmet impact performance. The authors have been asked by the DfT to provide a technical response to this unpublished paper, which is contained in the present report.

The primary evidence sources for the SHARP protocols are the 'European Co-operation in the Field of Scientific and Technical Research, Action 327' (COST 327), 'New Helmet Designs: Performance Assessment and Cost Benefit Analysis' (S100/L) and 'Motorcyclists' Helmets and Visors – Test Methods and New Technologies' (S0232). The Final Report of the COST 327 Action [Chinn *et al.*, 2001] contains detailed information on:

- Motorcycle accident causation;
- Injury distribution to all body regions;
- Detailed information on head injury severity, head injury location and helmet impact location in motorcycle accidents;
- Information on human tolerance to head impact based on detailed laboratory reconstruction of real-world motorcycle accidents; and
- Assessment of test tools and test procedures.

The S100/L and S0232 projects, undertaken by TRL on behalf of the DfT, were focussed on implementing improvements in helmet design and testing, based on the knowledge generated in the COST 327 project. An advanced helmet was developed that demonstrated the potential reduction in head injuries that could be achieved through improved helmet design. It was estimated that up to 100 lives per year could be saved if helmets of equivalent performance were worn by all motorcyclists. The S0232 project also developed a draft consumer information test programme that could be used to rate the impact performance of motorcycle helmets. This draft programme has been updated to provide the SHARP test and evaluation protocols.

It is understood that the DfT will publish a separate paper that describes the details of the SHARP test and evaluation protocols. It is not the purpose of this report to duplicate this exercise, but instead to explain the technical foundations of SHARP and hopefully address Dr Mills' concerns. Some of Dr Mills' comments are based on misunderstandings of the COST 327 data and the SHARP evaluation protocol, and it is hoped that the information in this report will help to clarify these issues.

The current understanding of head injury mechanisms are discussed in the report. Based on this information, it is apparent that there are three main factors that, for a given person and head impact site, affect the risk of serious head injury:

- Distribution of impact forces;
- Linear head acceleration; and
- Rotational head acceleration.

A motorcycle helmet can help to protect against injuries due to all three of these factors. The main helmet design factors that contribute to this are improved padding and energy absorption from the shell and liner to reduce the linear and rotational head acceleration in an impact, and a lower coefficient of friction to further reduce the rotational head acceleration. The SHARP rating is based on 32 flat plate, kerb-type and oblique impact tests per helmet model, which assess the performance of the helmet across a range of impact severities for which improved protection has been shown to be possible. No single test type dominates the assessment, so that a rounded approach to motorcycle helmet safety is encouraged.

In order to derive the helmet rating, the test results are weighted according to the best available motorcycle accident data. This weights the likelihood of impacts occurring to different regions of the helmet, of impacts occurring at different speeds, and of impacts with flat, kerb and oblique loading conditions, all based on the comprehensive accident studies in COST 327. The side and the rear of the helmet were found to be commonly impacted, and to have a strong correlation between impact location and injury. The side of the head was also found to be particularly vulnerable to injury. The weighting of the SHARP results according to real-world accident data ensures that improved helmet designs are targeted to where they will make the most difference to motorcyclist safety.

This report demonstrates that SHARP strongly encourages improved linear acceleration response and a lower coefficient of friction. Both of these, particularly a reduction in linear head acceleration, are recommended by Dr Mills in his paper. Furthermore, the report demonstrates that the reduction in each of these parameters encouraged by SHARP is in proportion to their contribution to the different mechanisms of head injury.

It should be noted that good helmet fit is also very important for the safety of the wearer. In addition to the impact safety rating, SHARP provides guidance on how to choose a helmet that fits well and is comfortable.

1 Introduction

The authors have been asked by the Department for Transport (DfT) to provide a technical response to the unpublished paper “Critical Evaluation of the SHARP Motorcycle Helmet Rating”, by NJ Mills. The paper by Dr Mills is based on an unpublished description of the performance evaluation protocol used to rate motorcycle helmets in the DfT’s motorcycle Safety Helmet Assessment and Rating Programme (SHARP).

It is understood that the DfT will publish a separate paper that describes the details of the SHARP test and evaluation protocols, the reasons for them, and how the test results are used to determine the rating for each helmet model. The purpose of this report is not to repeat that exercise, rather it is to explain the technical origins of SHARP and hopefully to allay the misgivings of Dr Mills and, in turn, motorcyclists who we believe can greatly benefit from SHARP. Furthermore, it appears that some of Dr Mills’ comments are based on misunderstandings of the performance evaluation protocol, and it is hoped that the information in this report will help to clarify these issues.

SHARP is a complex but robustly defensible combination of test results based on linear and rotational tests. Dr Mills’ paper describes only his views on rotation, the assessment of which in SHARP is based upon an oblique impact test. Hence, this paper concentrates primarily on that matter. Indeed, the importance of rapid rotational motion as a frequent cause of and contributor to serious and fatal brain injury is well established. This report also gives some information on the linear impact tests in SHARP in order to put the rotational (oblique) impact tests in the context of the overall helmet impact safety assessment.

The report starts with an overview of the international research effort that has led to the development of the SHARP rating scheme (Section 2). This is followed in Section 3 by a brief description of head anatomy, head injuries and injury mechanisms, and the effects of rotation on the head.

Dr Mills is critical of the SHARP mathematical model for head/helmet rotation and claims that it is insufficiently representative of a real helmet to justify incorporation within SHARP. Section 4 describes the model in more detail and gives examples of test results using motorcycle helmets against equivalent values predicted by the model. It also discusses sliding and rolling of a helmet in an impact.

COST 327 was a Europe-wide Action on motorcycle accidents and head and neck injuries. Much of the derivation of SHARP relied upon this published work; hence Section 5 is devoted to a brief description of the most relevant results from COST 327. This includes extracts from the Test Procedures chapter of the COST 327 Final Report [Chinn *et al.*, 2001] to illustrate the correlation between peak rotational acceleration and tangential force which a the critical link between brain injury and helmet tests.

Section 6 gives a brief overview of how the SHARP performance evaluation protocol combines the accident and test data to rate each helmet, and discusses the remaining issues raised in Dr Mills paper that were not addressed in the preceding chapters. The overall findings of this technical response are discussed in Section 7 and conclusions are drawn in Section 8.

2 Background

A collaborative European study, 'European Co-operation in the Field of Scientific and Technical Research, Action 327' (known as COST 327) included a detailed analysis of real world accidents and the protection offered by helmets. This novel research included the reproduction of helmet damage by replicating accident conditions in a laboratory environment. The severity of the head impact was determined using instrumented headforms and these data were related to the injuries known to have been sustained by the accident casualty. COST 327 set a number of recommendations for improved helmet safety including enhanced test requirements. These included having a higher head impact speed in helmet testing and also taking into account the effects of rotational acceleration in an impact, and therefore the coefficient of friction of the helmet.

In response to the COST 327 findings, TRL, funded by the DfT, developed an advanced technology helmet as part of the project 'New Helmet Designs: Performance Assessment and Cost Benefit Analysis' (S100L/VF) [Mellor *et al.*, 2004]. This project also assessed the potential life-saving capabilities of the advanced helmet, using an injury risk function developed using accident replications from COST 327. The advanced helmet designed for S100L/VF had a lightweight carbon composite shell fitted with a high-efficiency expanded polystyrene energy absorbing liner and a low friction sacrificial shell surface. The project concluded that an advanced helmet could reduce Abbreviated Injury Scale[1] (AIS) 6 injuries to AIS4, AIS 5 and 4 to AIS level 3. AIS 3, 2 and 1 injuries would be maintained. These reductions were estimated to be deliver an associated casualty prevention value of £52.7M (at 2003 values), based on a 10% wearing rate of the enhanced helmet.

Following the completion of this project, the DfT commissioned TRL to investigate ways of improving helmet performance and devise a possible consumer information scheme based on the information from COST 327 and S100L/VF. This project, 'Motorcyclists' Helmets and Visors – Test Methods and New Technologies' (S0232) [Mellor *et al.*, 2007], used the injury risk function and injury statistics from COST 327 as well as an exposure population, as the bases of an assessment protocol. The protocol could then be used to estimate the number of fatalities that would occur over a year wearing a certain helmet. This number was then compared to the number of fatalities that was predicted for a 'baseline' helmet. This baseline helmet was chosen as a representation of a typical motorcycle helmet on the market at that time. S0232 suggested that the introduction of a helmet consumer information programme could save up to 100 lives a year with improved helmet designs.

Project S0614/V8, Motorcycle Helmets: Test and Assessment Protocol Prove Out [StClair and McCarthy, 2007], is the most recent piece of DfT funded TRL work on motorcycle helmets. This project subjected five helmet models compliant with UNECE Regulation 22.05 (the dominant European legal requirement) [UNECE Reg 22, 2002] to a series of linear and oblique impact tests specified by the DfT. The objective was to ensure that the test and assessment protocols proposed for the basis of a consumer information programme were robust and suitable for implementation. The assessment protocol from S0232 was used to estimate the number of fatalities for the range of helmets tested and it was concluded that a comparative assessment could be made of helmet performance up to an impact velocity of 8.5 m/s. Limiting the impact speed to 8.5m/s carries a risk that the assessment would not consider the benefits that advanced helmet technology could offer in more severe impacts (e.g. up to 10 $m.s^{-1}$). However, testing at higher speeds is limited by practical considerations. Measuring the protection at 8.5 $m.s^{-1}$ aims to drive improvements in protection up to at least this speed.

[1] The Abbreviated Injury Scale is described in Section 5.4.1.

2.1 Helmet performance evaluation strategy

From the previous work described earlier, a strategy for assessing motorcycle helmet performance had become clear:

Helmets of various sizes should be assessed at a range of impact velocities up to and including 8.5 m/s and peak headform acceleration measurements taken

↓

The coefficient of friction of the helmet should be measured during oblique impacts to give an indication of rotational performance of the helmet and this performance should make up part of the assessment methodology

↓

Injury statistics relating to impact location distribution, impact shape distribution, head impact velocity distribution and exposure population should be used as part of the evaluation process so that not only energy attenuation performance but also accident risk is taken into account

↓

Helmet performance should be assessed based on risk of fatality while wearing the helmet and a rating assigned to a helmet model according to this performance measurement

This strategy has been implemented in the SHARP motorcycle helmet safety rating scheme. Some of the background knowledge developed during the above-mentioned projects is collated in this report to demonstrate the scientific basis for the rating scheme and address the issues raised in Dr Mills' paper.

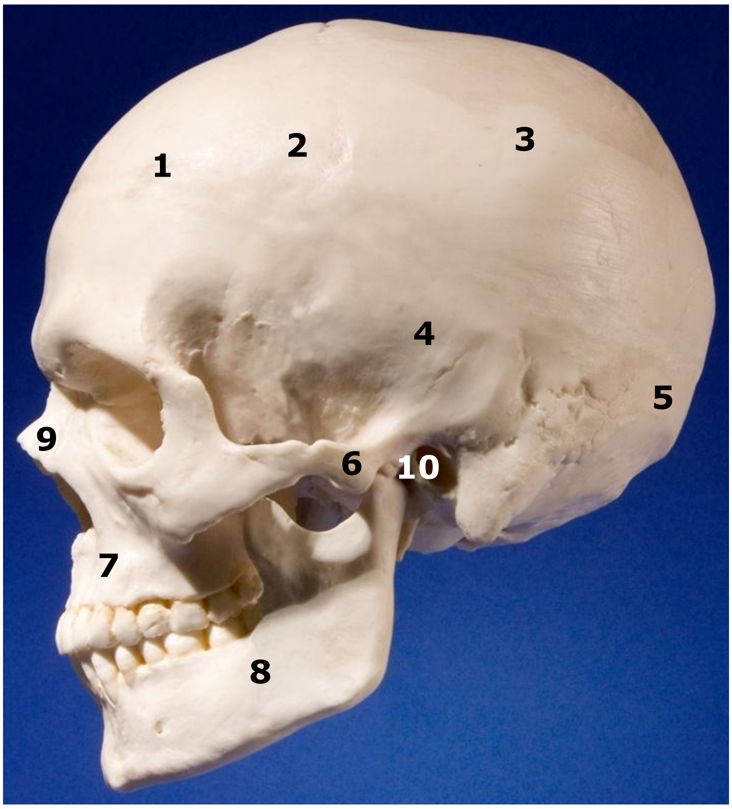
3 Head Anatomy and Injury

The head is a complex collection of bones and soft tissues. Head injury may refer to injuries to any of these tissues, and multiple head injuries may occur in a single accident. SHARP is focussed on reducing the risk of a fatal head injury, so this section of the report is focused on the most severe and life threatening types of head injury.

3.1 Head Anatomy

The skull comprises the cranial vault and the bones of the nose and jaw. Lateral and inferior views of the skull are shown in Figure 3.1 and Figure 3.2. The cranial vault is comprised primarily of several curved sections of bone - the frontal, left and right parietal, occipital, left and right temporal, sphenoid and ethmoid bones - that are joined together along irregularly-shaped sutures. The interior surface of the upper part of the cranial vault is relatively smooth, while the lower part has a number of projections that may interact with the brain in an impact. At the base of the cranial vault is the foramen magnum through which the brain stem and spinal cord intersect (labelled '3' in Figure 3.2).

The cranial vault tends to be thinnest at the side of the head and the posteroinferior (lower rear) part of the skull posterior to the foramen magnum. In addition to being relatively thin, the side of the cranial vault is relatively flat and is therefore particularly vulnerable to fracture.



1 Frontal bone

2 Coronal suture

3 Parietal bone

4 Temporal bone

5 Occipital bone

6 Zygomatic arch

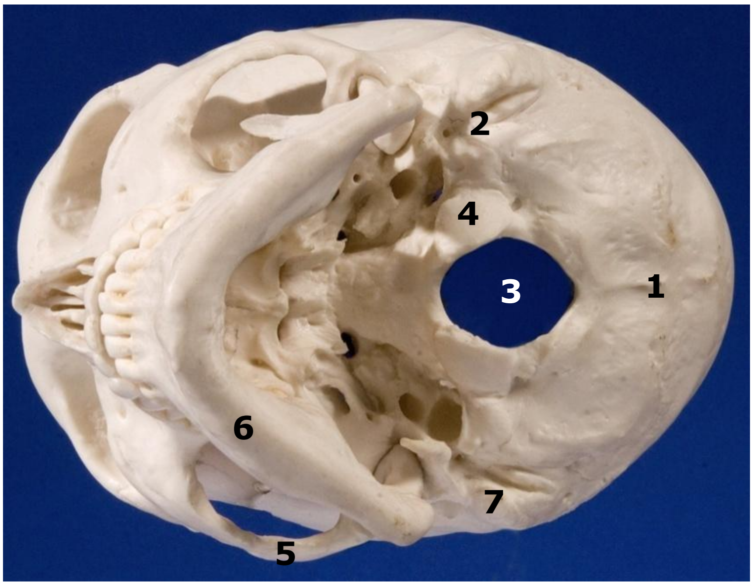
7 Maxilla

8 Mandible

9 Nasal bone and orbit

10 External auditory meatus

Figure 3.1: Key regions of the skull – lateral view



1 Occipital bone

2 Temporal bone

3 Foramen magnum

4 Occipital condyle

5 Zygomatic arch

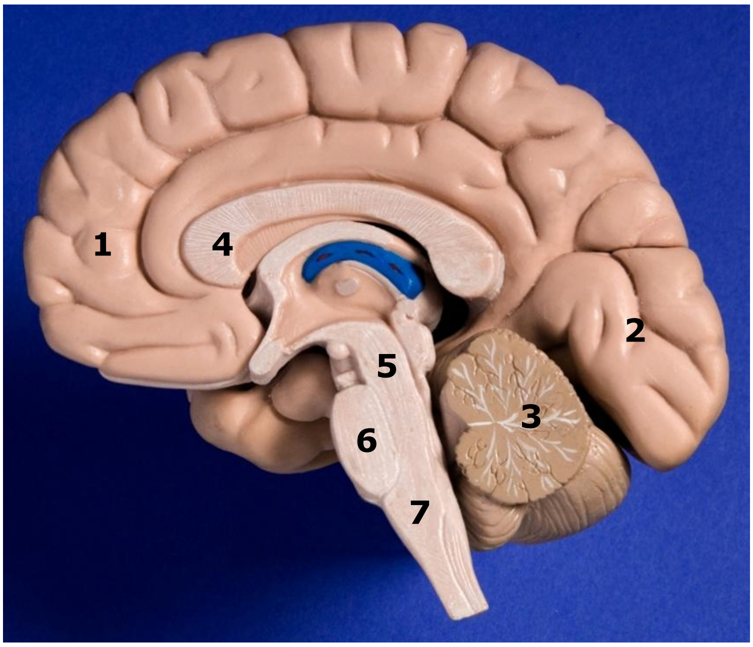
6 Mandible

7 Mastoid process

Figure 3.2: Key regions of the skull – inferior view

Much of the cranial vault is covered by the scalp. The outer layer of skin and connective tissue is separated from the periostium of the cranium by a layer of loose connective tissue that forms a shear plane between the skin and the cranium [Moore, 1985]. The scalp is well supplied with blood vessels and bleeding from scalp lacerations can be profuse.

Between the cranium and the brain are a series of protective coverings called the cranial meninges. The outer layer is the dura mater, the middle layer is the arachnoid and the inner layer is the pia mater, which adheres closely to the surface of the brain, dipping in to the fissures and carrying small blood vessels with it. An extension of the dura mater, called the falx cerebri, separates the left and right hemispheres of the cerebrum. Some of the key regions of the brain are shown in Figure 3.3.



1 Frontal lobe of cerebrum

2 Occipital lobe of cerebrum

3 Cerebellum

4 Body of corpus callosum

5 Mid-brain

6 Pons

7 Medulla oblongata

Figure 3.3: Key regions of the brain – mid-sagittal seciton

The brain is well supplied with oxygen and nutrients through a network of blood vessels. Blood vessels that enter the brain tissue first pass along the surface of the brain and, as they penetrate inwards, they are surrounded by a loose-fitting layer of pia mater.

3.2 Types of Head Injury

There are three types of head injury that are most relevant to the improved motorcycle helmet performance that SHARP is intended to encourage: cranium fracture; focal brain injuries; and diffuse brain injuries The mechanisms and likely consequences of these injury types have been considered in detail in numerous publications [e.g. Gennarelli, 1985; Melvin *et al.*, 1993] and a comprehensive review was undertaken as part of the COST 327 Action. A summary is given here.

3.2.1 Skull Fracture

Skull fractures may be simple or complex and occur due to direct impact of the head with another object. Simple linear fractures are generally considered to have little significance for brain injury, although dangerous complications may occur [Melvin *et al.*, 1993]. More severe impact forces may lead to comminuted or depressed fractures, where fragments of bone may be pushed into the underlying soft tissues causing damage to the blood vessels or brain tissue. Even when skull fracture does not occur, bending of the skull may be sufficient to damage underlying blood vessels and brain tissue.

3.2.2 Focal Brain Injuries

Focal (localised) brain injuries consist of epidural haematomas, subdural haematomas (sub-arachnoid haematomas), intracerebral haematomas, and coup or contrecoup contusions. Most focal injuries are due to direct contact with bone fragments from skull fractures, or to relative motion between different parts of the skull and the brain. Such relative motion may be due to linear or rotational acceleration of the skull.

Figure 3.4 shows some of the possible effects of brain movement relative to the skull. Such large relative movements have been observed in experiments [Shelden, 1944; Pudenz and Shelden, 1946; Gurdjian and Lissner, 1961; Gosch, 1970]. In most of these experiments the whole upper half of the skull had been replaced by a lucite calvarium and because the dura had been removed, the effects of tethering the brain at the vault could not be obtained.

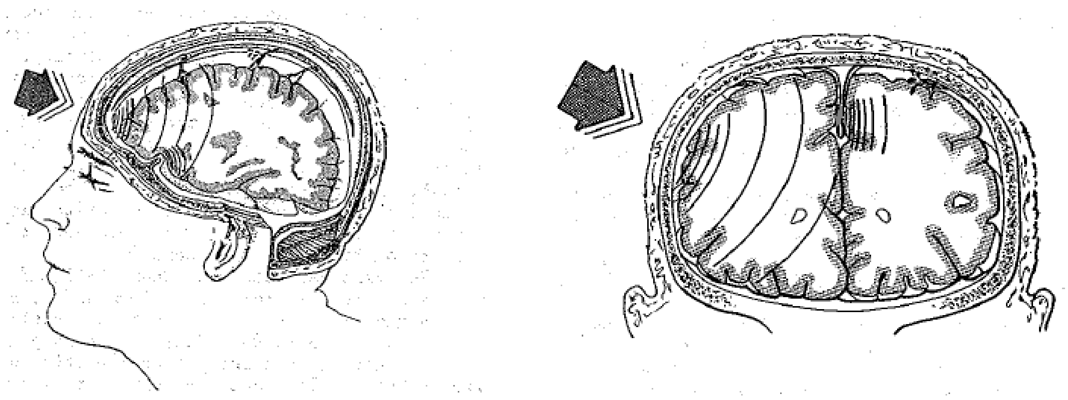


Figure 3.4: Some of the possible effects of relative brain movement [Viano, 1988]

The skull is smooth at the vertex, but highly irregular at the base. Therefore, sliding of the brain against the internal surface of the skull is facilitated at the vertex, but is impeded at the skull base where it can lead to high shear strains in the meningeal and cortical tissues. Sufficient shearing of these tissues will cause lacerations, contusions and haematomas in the cortex.

The shape of the base of the skull is much more irregular in the frontal and temporal regions than in the occipital region. This explains why most cerebral contusions occur at the frontal and temporal lobes [Gurdjian, 1966], regardless of whether the site of impact is frontal or occipital [Gurdjian, 1955].

The relative movement between the skull and the brain is always toward the site of impact. Because of this, intracranial tissue is compressed at the site of impact and strained at the contra lateral site. This leads to an increase in pressure at the site of impact and a reduction in pressure at the opposite site (Figure 3.5).

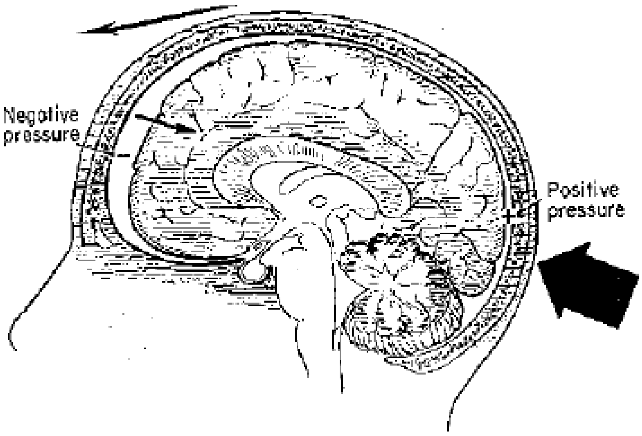


Figure 3.5: Intracranial pressure changes due to relative movement between brain and skull [Douglass *et al.*, 1968]

It should be noted that focal brain injuries are reportedly highly correlated with fatality [Melvin *et al.*, 1993]. For example, Gennarelli and Thibault [1982] reported an incidence of acute subdural haematoma of 30%, with an associated mortality rate of 60%.

3.2.3 Diffuse Brain Injuries

Diffuse injuries consist of concussion, swelling of the brain and diffuse axonal injury (DAI). Mild concussion may include disorientation and confusion, with moderate (often referred to as 'classical' concussion) concussion leading to loss of consciousness for up to 24 hours. Recovery rates from mild and moderate concussion are good, but severe deficit in brain function may result in a small minority of cases. The clinical outcome for patients with moderate concussion is dependent on any other head injuries received [Melvin *et al.*, 1993]. Loss of consciousness greater than 24 hours is associated with a much higher rate of brain deficit and even fatality. Melvin *et al.* [1993] reported that close to 2% of patients with loss of consciousness greater than 24 hours may have a severe deficit and 2% may have moderate deficit.

DAI is associated with widespread disruption of the axons in the cerebral hemispheres, mid-brain and brainstem. DAI involves loss of consciousness lasting at least 24 hours and possibly weeks. 55% of patients are likely to have died one-month post-trauma, 3% may have vegetative survival and 9% may have severe deficit [Gennarelli, 1981; Melvin *et al.*, 1993].

Brain swelling due to an increase in intravascular blood within the brain may worsen the effects of primary brain injury due to increased intracranial pressure. This increased pressure may force the brain and brainstem downwards through the foramen magnum causing further damage to the tissues. In this way, focal injuries may greatly increase the risk of fatality for patients with DAI.

Holbourn [1943] found in his gelatin model of the brain that in rotational motion the highest shear strains occurred in the anterior part of the temporal lobe near the base of the skull.

Rotation of the skull relative to the brain presses the highly irregular skull base towards the brain. This leads to a combined compression and shearing of the meningeal and cortical tissues in this area, which increases the effects of the sliding of the brain over the skull base. The effects of this relative rotation are most severe when the head is subjected to a rapid forward or rearward motion relative to the torso.

The main functions of the connections between the skull and the brain at the vertex are to tether the brain to prevent excessive movements and to protect the blood supply to the cortical tissues. The functions of the tissues crossing the skull-brain interface at the skull base are much more vital. All of main blood vessels supplying the entire brain and the neurological connections between the brain and the rest of the body pass through this location. Shearing of these blood vessels and neurological connections due to rotation of the brain relative to the cranium can cause serious injury to these structures.

3.3 Rotational Motion: the Dynamics of Impact

Injury is caused when one particle of the body moves relative to the adjacent particle such that the elastic limit of the joining material is exceeded and damage occurs; the brain is no exception. Relative movement can occur only if adjacent particles are differently accelerated over time and although an impact to the head results in an acceleration (deceleration is simply negative acceleration) injury will not occur if the acceleration is evenly applied to all particles of the brain. In practice injury does occur and to understand why, it is important to consider the motion of a particle in space and for simplicity this will be confined to a plane (Figure 3.6).

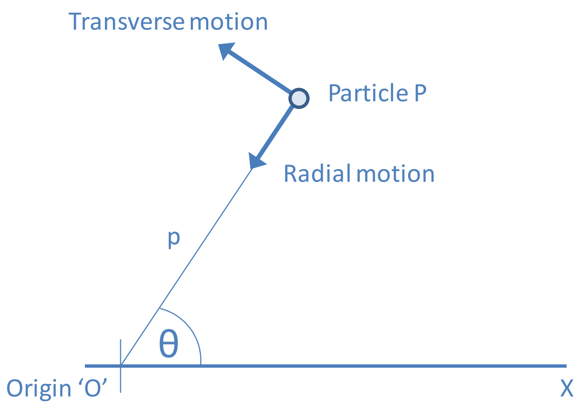


Figure 3.6: Representation of a particle in space using polar coordinates

With O as the pole, let OX be the initial line, let the polar coordinates of particle P be p, length of vector **p**, and θ being the angle between **p** and OX, positive in the anticlockwise sense. Thus $\dot{\theta}$ is defined as the angular velocity of P about O. The direction of the position vector of P is the radial direction and perpendicular to this, with θ increasing, is the transverse direction.

It can be shown using conventional vector analysis that the radial and transverse acceleration components are as follows:

radial component $p\ddot{\theta} + 2\dot{p}\dot{\theta}$

transverse component $\dot{p} \bullet p \ \dot{\theta}^2$

If the motion is circular about O with radius r and the particle passes through the position P with speed *v* then the tangential component becomes $r\ddot{\theta}$ and the normal component becomes $\frac{v^2}{r}$.

Thus this has illustrated mathematically that when a body rotates with uniform angular velocity there is a force toward the centre of rotation proportional to the distance from this centre and the mass. Hence, an acceleration can be defined acting along a line passing through the centre. If the angular velocity changes then an angular acceleration is induced and this, in turn, gives rise to a linear acceleration normal to the line through the centre and proportional to the distance from the centre. It is this gradation of linear acceleration that gives rise to tangential sheer forces within the brain causing severe injury as is described in Section 3.2.3.

4 Factors Influencing the Potential for Rotational Motion in an Impact to the Head: a Theoretical and Practical Analysis

4.1 The Causes of Head Rotational Acceleration

A number of authors have commented on the relative contributions to brain injury of translational and rotational head accelerations. As noted in Section 3.2.3, rotational acceleration is generally considered to be associated with a range of injuries from concussion through to diffuse axonal injury, the effects of which may be very severe in terms of risk of death or poor long-term outcome for survivors.

The head will rotate in an impact if the resultant impact reaction force does not intersect with the centre of gravity of the head. This may occur in a vertical impact if the contact point, and therefore the normal reaction force F_N, is offset horizontally from the centre of gravity of the head (Figure 4.1). In a head impact with a lateral velocity component, friction between the head (or helmet) and the impacted surface will cause a tangential force F_T that will also cause a rotation (Figure 4.2).

The tangential force is proportional to the normal force multiplied by the coefficient of friction between the helmet and the impact surface, denoted by the symbol μ. Sliding friction is constant, whatever the relative velocity of the two contact surfaces. Therefore, halving the normal force will halve the tangential force and commensurately reduce the rotational acceleration. Halving the coefficient of friction will half the tangential force component and therefore halve its contribution to rotational acceleration.

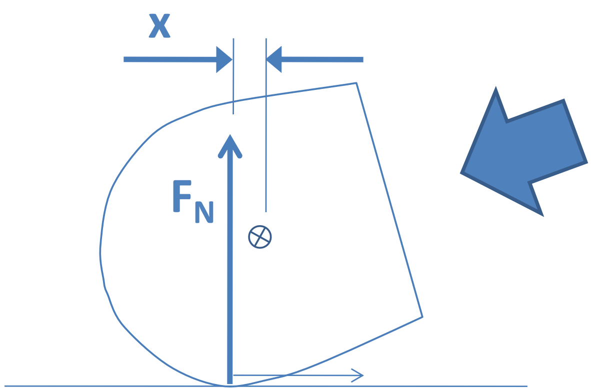


Figure 4.1: Example of head rotation due to head centre of gravity offset from impact point

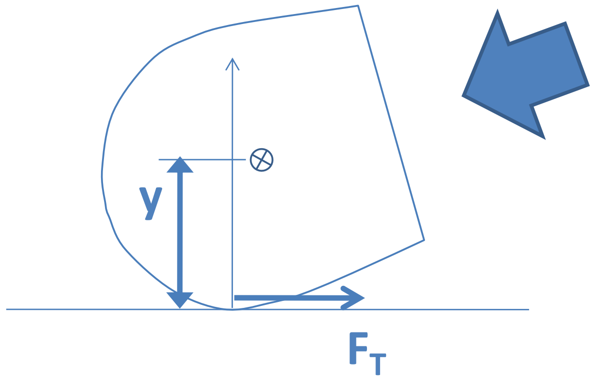


Figure 4.2: Example of head rotation due to sliding friction causing a tangential force

Finan *et al.* [2008] discuss the implications of these effects in some detail. In some circumstances these rotational components will combine to increase the total rotational moment on the head, while in other circumstances the components may tend to cancel each other out and, if evenly balanced, result in zero rotational moment on the head. Reducing friction between the helmet and the impacted surface may therefore increase or decrease the rotational acceleration of the head depending on the trajectory of the impact.

The authors also undertook physical helmet tests to confirm their theoretical discussion. They concluded that '*while friction may be beneficial in a particular impact, in an averaged sense it is never beneficial and may be quite costly.*' That is, there may be some impact configurations in which the forces due to friction will protect the brain from rotational acceleration, but in the majority of cases the friction will increase the rotational acceleration and therefore the head injury risk. Finan *et al.* therefore recommended a reduction in the coefficient of friction of the helmet in order to provide the best overall protection against rotational brain injury.

4.2 Angular Acceleration of a Sphere Related to Surface Friction and an Asymmetric Centre of Gravity

4.2.1 Surface Friction

It is important to understand not only how the brain may be affected by rotational motion but also how rotational motion may be imparted and what fundamental variables can influence the outcome. It is assumed, for simplicity, that the head is a sphere of uniform mass "m" kg and of radius "r" metres. It is then assumed that the sphere is moving and strikes a plane surface such that the angle between the surface and the direction of travel is not 90°; the coefficient of friction between the sphere and the plane surface is defined to be µ. Thus, there will be a force normal and a force tangential to the surface of the sphere at the point of contact (see Figure 4.3). The rotational acceleration in Figure 4.3 is denoted as $\ddot{\theta}$

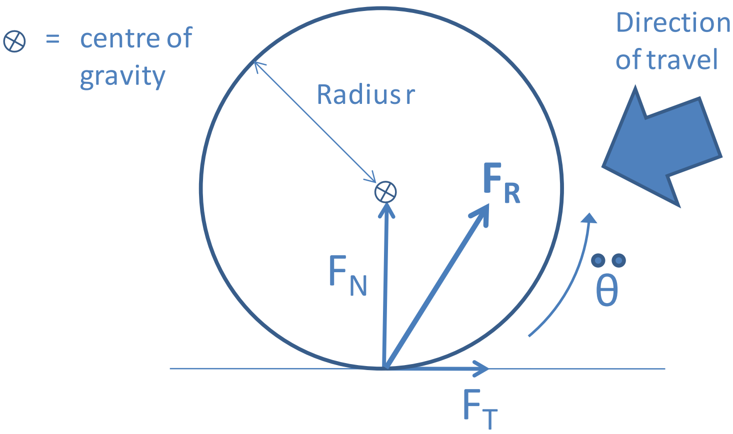


Figure 4.3: Schematic of a spherical head impacting a flat surface obliquely

If the force normal to the sphere at the point of contact is assumed to be F_N then the tangential force is $\mu.F_N$ and the resultant (total) force F_R is:

(i) $$F_R = \sqrt{\left[F_N^{\,2} + (\mu.F_N)^2\right]}$$

Which can be written:

(ii) $$F_R = F_N.\sqrt{\left[1 + \mu^2\right]}$$

Hence, using F = ma, the linear acceleration at the centre of the sphere at any instant is:

(iii) $$a = \frac{F_N}{m} \cdot \sqrt{\left[1+\mu^2\right]}$$

Where m is the mass of the sphere and a is the linear acceleration at the centre of the sphere. Rearranging this gives:

(iv) $$F_N = \frac{ma}{\sqrt{\left[1+\mu^2\right]}}.$$

Furthermore, the moment of inertia I of a sphere about a diameter is:

(v) $$\frac{2mr^2}{5}$$

and

(vi) $$\frac{F_N}{I} = \frac{5a}{2r^2\sqrt{1+\mu^2}}$$

and the angular acceleration is:

(vii) $$\therefore \quad \ddot{\theta} = \frac{5}{2}\frac{a}{r}\frac{\mu}{\sqrt{1+\mu^2}}$$

It should be noted that when the coefficient of friction is reduced to zero then the rotational acceleration of a uniform sphere is zero irrespective of other conditions. At this point in the analysis it is worth mentioning that Dr Mills draws a clear distinction between rolling and sliding, and this is discussed further in Section 4.4.

4.2.2 Asymmetric Centre of Gravity (Assuming Unchanged Moment of Inertia)

In the equations developed above it was assumed that the centre of gravity was in the centre of the sphere and, therefore, the force normal to the surface acted through the centre of gravity and the resulting rotational acceleration was dependent only upon the tangential force (and therefore the coefficient of friction). However, in practice, because of the irregularity of the head it is likely that the force normal to the surface of the head at the point of impact will not pass through the centre of gravity and the consequences are examined below.

It is difficult to represent an irregular body mathematically in this context, so the problem was examined by considering a sphere of the same radius and the same magnitude of moment of inertia at the instant of impact, but with the centre of gravity offset from the centre of the sphere. Again, let the normal force be F_N and the offset be 'x' as illustrated in Figure 4.6.

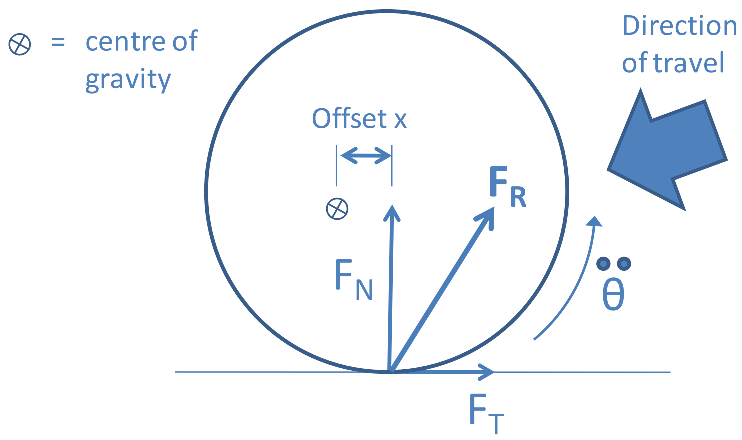


Figure 4.4: Moment Nx caused by offset centre of gravity

Hence:

(viiii) $$\ddot{\theta} = \frac{\mu F_N r + F_N x}{I}$$

and, therefore, from Equation (v) above:

(ix) $$\frac{F_N}{I} = \frac{5a}{2r^2\sqrt{1+\mu^2}}$$

(x) $$\therefore \ddot{\theta} = \frac{5a(\mu r + x)}{2r^2\sqrt{(1+\mu^2)}}$$

It should be noted that the offset 'x' of the normal force F_N from the centre of gravity of the sphere will change as the sphere rotates, and the offset may be zero at some instant during the rotation of the sphere. It should also be noted that Figure 4.6 illustrates the offset such that the moment generated will increase the tendency for rotational acceleration. It is equally possible that the opposite could occur, but it was thought appropriate to consider the worst case.

4.3 The Effects of Friction, Offset Centre of Gravity and Moment of Inertia on Rotational Motion

4.3.1 Unhelmeted Human Head

Values for the mass and moment of inertia of human heads have been developed from tests with human subjects, and these have been used to determine values for the equivalent sized dummy heads. It is the dummy head values, which are representative of a human of the same size, that were used in the following analysis such that the results could be compared with experimental data.

The equations presented in Section 4.2.1 were used to investigate the effects of surface friction and linear acceleration on rotational acceleration during an oblique impact. The radius of the sphere was chosen to be 90 mm, based upon a Hybrid II[2] headform, and the acceleration range was 50 *g* to 300 *g*. The friction coefficient was allowed to vary from 0 to 1. It is clear from Figure 4.5 that as the coefficient of friction rises so does the rotational acceleration. It is also clear that, for a given coefficient of friction, the rotational acceleration increases as the linear acceleration increases. For example, for a linear acceleration of 100 *g* the rotational acceleration rises from below 10,000 $rad.s^{-2}$ for a coefficient of friction of 0.2, to above 25,000 $rad.s^{-2}$ for a value of 0.9. This is from below the generally accepted human tolerance value (i.e. 10,000 $rad.s^{-2}$) to well above it. Similar dramatic variation can be seen for the changes in linear acceleration.

The equations presented in Section 4.2.2 were used to investigate the effect of an offset centre of gravity, in addition to surface friction, on rotational acceleration during an oblique impact (Figure 4.6). The radius of the sphere was 90 mm and the offset x was allowed to vary from 0 mm to 25 mm, which is typical of the maximum offset that may be achieved by a head of similar size. The linear acceleration was 150 *g* for this analysis and the friction coefficient μ was, again, allowed to vary from 0 to 1. It can be seen in this example that the rotational acceleration is less sensitive to a change in offset, even though the range used is greater than that found in practice.

Table 4.1: Variable ranges for Figure 4.5 and Figure 4.6

Variable	Range and unit
Radius of head:	90 mm
Mass of head	5 kg
Linear acceleration (range for Figure 4.5):	50 *g* to 300 *g*
Offset of centre of gravity (range for Figure 4.6):	0 to 25 mm

[2] The Hybrid II is was developed in the early 1970 for automotive safety test procedures and introduced into US car crash Regulations in 1973. The Hybrid II headform was evaluated in COST 327.

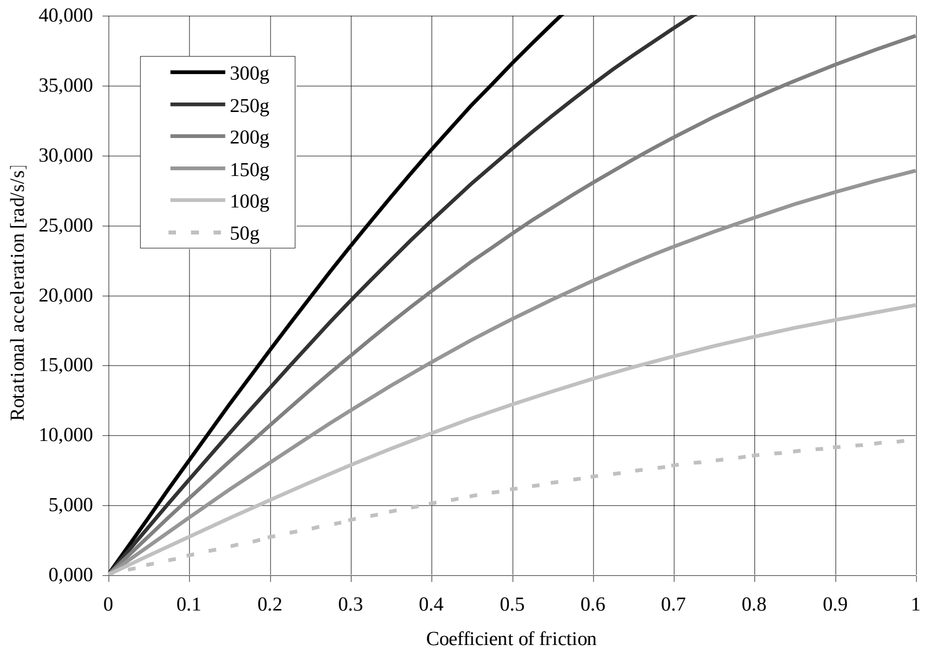


Figure 4.5: Theoretical response of a uniform sphere during an oblique impact (rotational acceleration vs. friction coefficient for different values of acceleration)

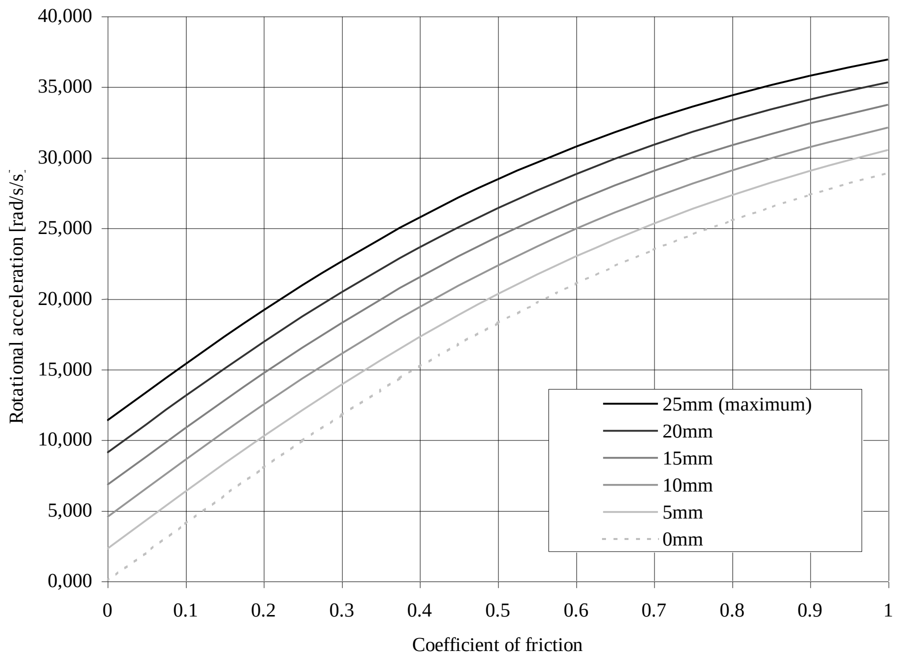


Figure 4.6: Theoretical response of sphere with offset centre of gravity during an oblique impact (rotational acceleration vs. friction coefficient for different values of offset, based on linear acceleration of 150g)

4.3.2 Helmeted Head

The foregoing analysis related to the human head alone and the analysis was repeated but with a motorcycle helmet included. The theory is identical to that above and only the values for radius and moment of inertia needed to be changed. However, helmets vary in size and mass and thus inertia and, therefore, it was considered important to investigate ranges of values at least as great as those spanned by current helmets. It is also true that the force normal to the helmet surface, and therefore the linear acceleration, will vary for different helmets and different impact conditions and this was also investigated.

Table 4.2 list the ranges used for each variable which are plotted subsequently in Figure 4.7 to Figure 4.9. It is not surprising that Figure 4.7 and Figure 4.8 show similar trends to those for the human head although the addition of a helmet, which increases the rotational inertia, has made the system a little less sensitive to a variation in friction. Figure 4.9 shows a range of angular inertia from 0 to 0.04 km.m^2, which is a far greater range than for helmets on the market. Nevertheless, it is interesting that as the angular moment decreases the system becomes more sensitive to variation in friction and it is important when designing a helmet for low mass to ensure that the benefits of low mass are not offset by the possible increase in rotational motion.

Table 4.2: Variable ranges for Figure 4.7 to Figure 4.9

Variable	Range and unit
Radius of helmet (full faced):	140 mm
Moment of inertia of helmet (full faced):	0.02 kg.m^2
Moment of inertia of helmet (range for Figure 4.9)	0 to 0.04 kg.m^2
Mass of head (Hybrid II dummy)	4.3 kg
Moment of inertia of head (Hybrid II dummy)	0.0147 kg.m^2
Linear acceleration range:	50 *g* to 300 *g*
Offset of centre of gravity:	0 to 80 mm

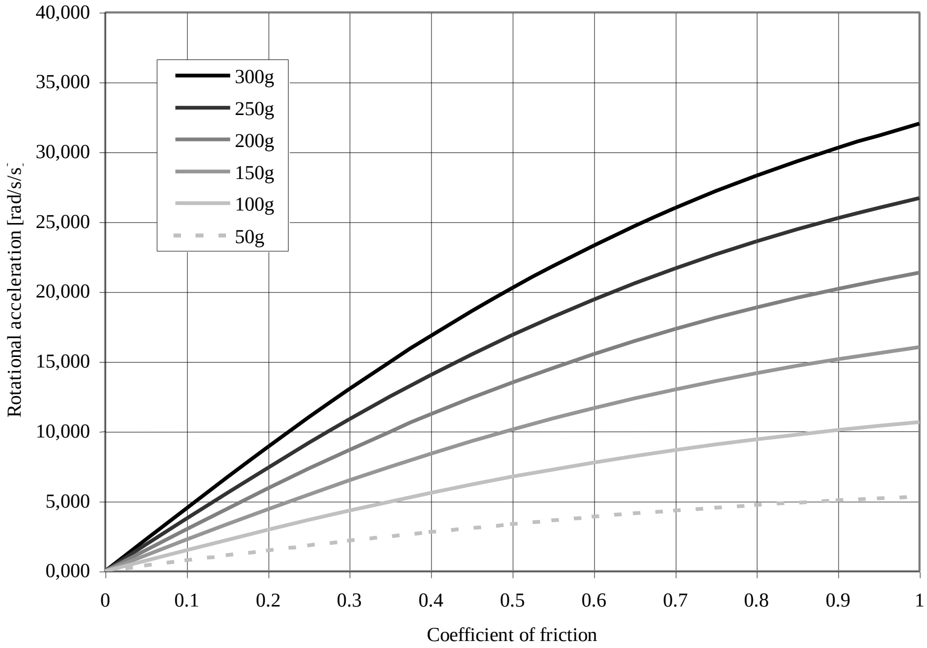


Figure 4.7: Theoretical rotational acceleration of helmeted head vs friction coefficient for different values of linear acceleration

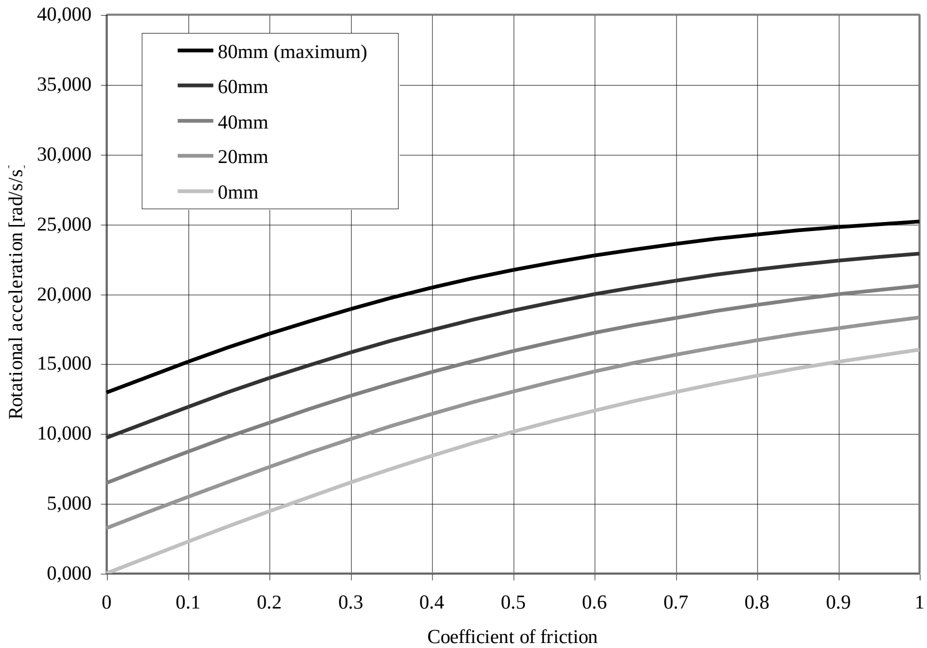


Figure 4.8: Theoretical rotational acceleration of helmeted head vs friction coefficient for different values of offset (based on linear acceleration of 150g)

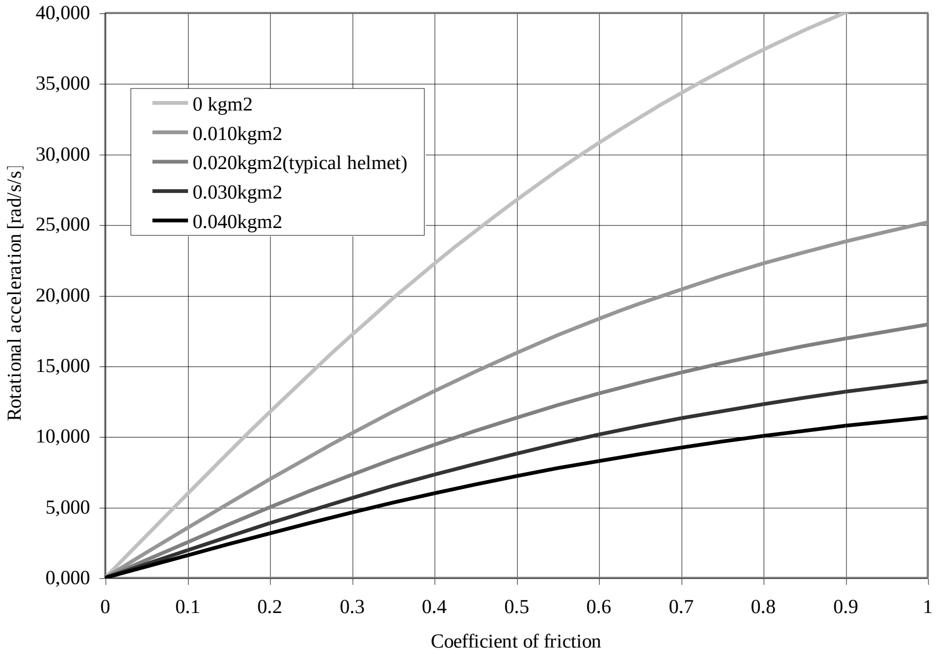


Figure 4.9: Theoretical rotational acceleration of helmeted head vs friction coefficient for different values of helmet angular inertia (based on linear acceleration of 150g)

4.3.3 Comparisons with Experimental Data

A test programme was devised to investigate the correlation between rotational acceleration and peak tangential force and the effect low friction and offset of the centre of gravity on rotational acceleration and tangential force [Chinn *et al.*, 2001]. Four helmet types were chosen: full faced and open faced helmets with GRP (glass reinforced plastic) shells, and full faced and open faced helmets thermoplastic shells. A Hybrid II dummy headform as described in the previous section was fitted to the helmets.

The results are included here so that the foregoing theoretical results may be compared with practical tests. It is not possible to plot the practical results in exactly the same way as for the theory because some parameters that are held constant theoretically, for example linear acceleration, are not constant in practice. Figure 4.10 shows the results of numerous tests onto a flat anvil angled at 15° to the vertical (as used in UNECE Regulation 22.05, BS 6658:1985 and SHARP), where the impact velocity, and hence the linear acceleration, was varied. The graph also shows lines based upon the theoretical calculations predicted for helmets of two different, but typical, moments of inertia (representing an open-faced and full-face helmet respectively) and a coefficient of friction of 0.7.

It is clear that the test results lie very close to the theoretical lines and with a nominal friction coefficient of 0.7 the rotational acceleration varied from 2,000 rad.s^{-2} to above 10,000 rad.s^{-2}.

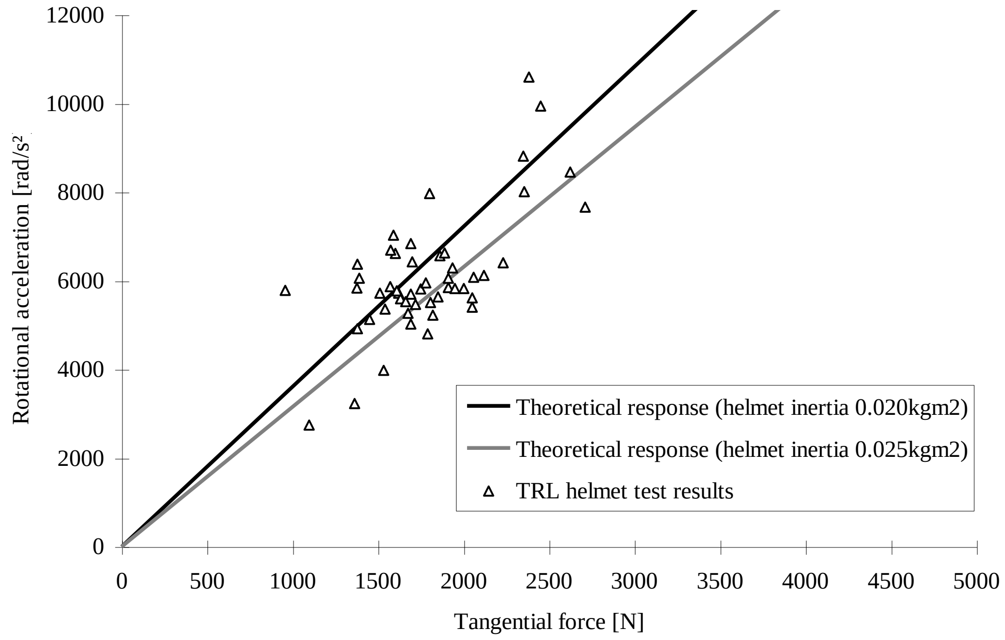


Figure 4.10: Comparison of theoretical response of helmet headform with laboratory helmet test results (oblique impact onto an abrasive surface)

4.4 Rolling and Sliding: the Conundrum

Dr Mills draws a clear distinction between sliding and rolling. Such a distinction is not relevant, and moreover is erroneous. For a friction coefficient of zero it is inevitable that that the helmet can only slide. For all other values there will be both sliding and rolling. Only when the friction coefficient is infinite will the helmet not slide. What matters is the rotational acceleration, which is the injurious factor, and the graphs above have shown the relationship between this and the tangential force for low and high values of friction for theoretical results based upon the simple model and experimental results for motorcycle helmets.

As an example, consider a motorcycle which travels from A to B through twists and turns in the road because the tyres grip the road. If the road is covered in ice, the tyres no longer grip and acceleration simply spins the wheel without forward movement and braking locks the wheel with no retardation. However, even on a high friction dry surface, where the coefficient of friction can be almost one, the wheels will always be moving at a slightly different speed to that of the motorcycle: slightly faster during acceleration and slightly slower during light braking. Therefore, for all practical purposes there is always slip in combination with roll. Only when the friction is infinite such as on a funicular railway does the "driving" wheel move at the same sped as the vehicle with no slip.

Dr Mills further criticises the method because he believes that the assumptions will break down for different impact sites. He has provided a picture showing the deformation of a helmet for an impact with a vertical velocity of 7.5 $m.s^{-1}$ and a horizontal velocity of 9.8 $m.s^{-1}$. This represents a fall from 2.86 m at a horizontal velocity of 22 mile/h; hardly typical of a rider falling from a motorcycle and his helmeted head striking the ground.

In the 15° oblique impact test the normal velocity is much lower than in the linear impact tests (2.2 $m.s^{-1}$ in the oblique test at 8.5 $m.s^{-1}$, compared with 6 to 8.5 $m.s^{-1}$ in the linear impact tests) and the deformation much less. Hence the moment of inertia, which is dependant on the mass and radius, changes little: the mass will not change and the radius changes by only a small amount (due to the small deformation). Helmets are not spherical but the variation in radius (excluding the chin guard, which is not tested in SHARP) is not sufficient to give a variation in the moment of inertia to invalidate the model. The variation that can be expected is shown in the two lines in Figure 4.10.

Dr Mills states that the values used by SHARP for the normal velocity are some 2.7 to 3.9 times those used in a series of oblique tests reported in COST 327. It appears that Dr Mills bases this on the assumption that for a given vertical velocity and hence force the horizontal velocity and hence tangential force must automatically increase. Such a premise was not assumed by SHARP and the vertical force used was simply to determine the characteristics of the shell and liner at higher velocities. Nevertheless, the vertical velocity and hence vertical force remains within values that are relevant to helmet testing and accidents at speeds within the range that it is known that a helmet can reduce injury potential. TRL has tested numerous helmets and some gave values as high as 0.9 just within the range of 0 to 1 evaluated as part of the theoretical study.

Dr Mills extends his argument to state that the equation derived by Halewood and Hynd (actually from [Mellor *et al.*, 2007]) for the purpose of assessing rotational acceleration ceases to be valid if the helmet rolls on the road surface.

When the helmet first strikes the anvil it is not rotating. Thereafter, it begins to rotate and as the impact progresses the helmet will slide and roll to a varying extent dependant upon the factors discussed above. During the impact the rotational acceleration will increase to a peak and then reduce to zero as the helmet departs from the anvil. It is the peak rotational acceleration which is the injurious parameter and this is proportional to the tangential and normal force, the measurement of which is the basis for the SHARP ratings.

Furthermore, a helmet impact lasts no more than about 15 milliseconds. This is the duration of the injurious impact, which will inevitably be a combination of rolling and sliding - it cannot be otherwise - when a helmet first strikes the road or other surface. Indeed, Figure 9 from Mills *et al.* [2009] shows exactly this, both for series of images from high-speed film of an oblique impact test, and for the matching FE simulation.

Dr Mills also contends that the coefficients of friction in the initial SHARP data set (which ranged from 0.54 to 0.86 with a mean of 0.68) seemed rather high, but provided only data from two helmet models from one manufacturer in support of this. The range in SHARP is as measured in the oblique impact tests; further more, they are in good the baseline helmet in Mellor *et al.* [2007]. There are also differences in the way that the coefficient of friction is calculated in different studies, which may lead to different results. SHARP uses the coefficient of friction calculated from the ratio of the normal and tangential forces *at the time of peak tangential force*, because the peak tangential force has been shown to be highly correlated with rotational acceleration (see Section 5.5).

5 The Contribution of COST 327 to SHARP

5.1 Introduction

The COST 327 Action was an international European project in which nine countries participated and 14 organisations across Europe contributed. It is not the purpose of this report to describe the research, the details of which are fully published in the final report of the Action available on the EU CORDIS website [Chinn *et al.*, 2001][3]. Rather it is to select those parts that were particularly relevant to SHARP and show how they were used in the derivation of the motorcycle helmet rating scheme.

5.2 Accident Analysis

The analysis showed that in over 60% of the casualties rotational motion was a contributory factor, and in over 30% it was the sole cause of brain injury, hence the importance of rotation.

It is perhaps unfortunate that the final report of the action contains only a summary of each of the working group reports. This has led to misunderstandings on the part of Dr Mills regarding the distribution of the impact site location on the helmets.

The findings from COST 327 were that the location of damage is distributed fairly evenly with 26.9% lateral right, 26.3% lateral left, 23.6% frontal and 21.0% to the rear, slightly fewer than the other regions. The crown, section 35, received only 2.2% of the impacts. It is *not* true that the definition of 'lateral left' covers all sites to the left of the mid plane. Each of the numbered sections apart from 35 (the crown) is divided into left and right sides and either front or rear depending upon the section. This means that the helmet was divided into eight regions radially when viewed from above, which enables the impact location distribution cited in the S0232 project [Mellor *et al.*, 2007] and used in SHARP to be determined very accurately.

To clarify this, Table A.1 in Appendix A shows the detailed head impact location distribution data from the Accident Data Task Group interim report [Otte *et al.*, 1998][4], which illustrates this. The table shows fewer than half of the final number of accident cases that were assessed, but shows the more detailed impact location distribution that was available in the COST 327 database that was summarised for the COST 327 Final Report [Chinn *et al.*, 2001].

Furthermore, the COST 327 Final Report notes that:

> *'It is clear that injuries to the side of the head (lateral injuries) and injuries to the rear correlate exactly with the damage location. However, injuries to the face, upper and lower, occur not only with frontal impacts as may be expected, but also with lateral impacts. The reason for this is not clear, but it is possible that loads to the side of the helmet are transmitted to the face. Damage to the upper part of the helmet seems to be evenly distributed around the helmet and probably correlates with the injury location.'*

Dr Mills reported that 79% of impacts (those impacts that were with rounded objects) were omitted from the SHARP analysis, and that the 60% oblique, 38.4% flat, 1.6% kerb impact type distribution proposed in Mellor *et al.* [2007] and used in the SHARP protocol was therefore incorrect. This is a misinterpretation of the COST 327 accident data and how it was used to develop SHARP. The COST 327 Final Report notes that 60% of impacts were oblique, which leaves 40% where linear impact was the primary loading mechanism. The flat and rounded impacts in Table 3.6 of the COST 327 Final Report

[3] http://ec.europa.eu/transport/roadsafety_library/publications/cost327_final_report.pdf
[4] http://www.cordis.europa.eu/cost-transport/src/cost-327.htm

were grouped together as being best represented by the flat anvil test procedure. Rounded was quite broadly defined and included, for instance, car body panels which deform and are far better represented by the flat anvil test than either the kerb anvil test of Regulation 22.05 or the hemispherical anvil test of BS 6658:1985. The remaining edge-type impacts, which included, for example, the corner of an A-pillar or cant rail, were considered to be best represented by the kerb anvil test. This gives the impact type distribution used in SHARP.

5.3 Accident Reconstruction

As part of the COST 327 Action, TRL replicated over 20 motorcycle accidents in which impacts to the helmet occurred. In each case the accident details were studied to determine what the rider’s helmeted head had struck – e.g. the road, or part of a vehicle such as the wheel or bonnet - during the accident. The relevant parts were obtained and set up in a laboratory; further, brand new helmets of the same type worn by the rider at the time of the accident were purchased. A fully instrumented dummy head fitted with the duplicate helmets was dropped onto the component part at various velocities until the damage to the helmet matched the damage on the original accident helmet. In addition to the instrumented headform the components were mounted onto a load cell to measure the force normal and tangential to the surface of the helmet. For each test a new helmet was used. Figure 5.1 shows an example of a reconstruction test onto a car door.

Figure 5.1: Example of a TRL accident reconstruction test

The accident data included details of the brain injuries as recorded by a Consultant Neuropathologist. For the fatal injury cases a histological examination of the brain provided the details on the injuries; for those that were not fatal a CT scan was taken. Figure 5.2 is an example of the brain injuries recorded for a fatal accident.

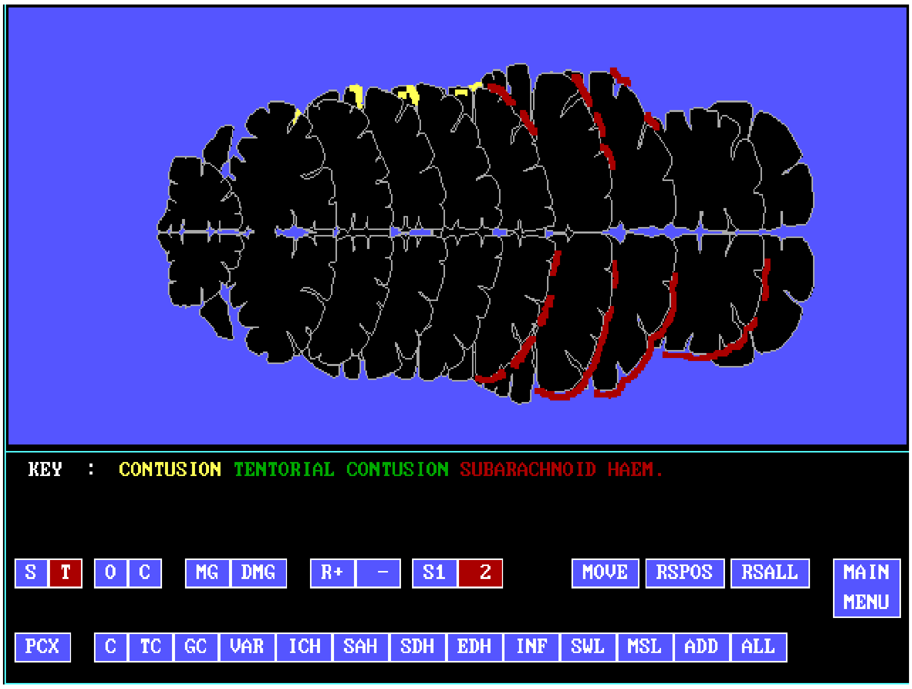


Figure 5.2: An example of brain injuries to a fatally inured motorcyclist

The information obtained from the brain injury analysis and the accident reconstructions were used in a finite element (FE) model of the brain developed by Strasbourg University. This enabled a wide range of accident conditions and the consequential injury potential to be investigated. This exercise was not specifically included as part of the analysis to develop the SHARP protocol; nevertheless it relates directly to an investigation of the causes of DAI (diffuse axonal injury) and other brain injuries.

The finite element simulation was correlated against the TRL replications and provided a link between the TRL accident replications, the distortion in the brain, rotational acceleration and tangential forces. It is worth noting that FEA, if properly calibrated and validated, is an invaluable tool for a parametric study across a wide range of variability not feasible through tests; this is then used to determine the much narrower range for practical investigation. It is rarely if ever wise to rely upon the results of one or two examples from FEA to support a hypothesis for which only practical tests can truly be relied upon.

5.4 Human Tolerance

5.4.1 Brain Injury Related to Criteria

Of critical importance to the derivation of SHARP was to relate the variables measured to human injury tolerance. Human response to a given dose or injurious parameter varies across a range of the population. The dose-response curve tends to be 'S' (sigmoid) shaped such that as the magnitude of the injurious parameter increases so does the proportion of the population that sustains an injury of a given severity. Thus, a family of 'S' curves can be generated for a range of injury severity (such as AIS) and a measurement (such as peak linear head acceleration) or injury criterion (such as HIC, the Head Injury Criterion). AIS is the Abbreviated Injury Scale, which is used to code the severity of injuries arising from road traffic accidents [AAAM, 1990], and much of the COST 327 and other research refers to this scale. The scale goes from 0 (no injury) to 6 (virtually unsurvivable). AIS ≥ 3 injuries are usually considered to be 'serious' or 'life threatening'.

To guide the reader Appendix B gives examples of the type of head and brain injury that correspond to a given value of AIS. The 1990 version of the AIS, which was used in the COST project, contains nearly 200 detailed physical head injury codes (not including the face) and nearly 40 additional codes relating to loss of consciousness and neurological deficit.

COST 327 analysed statistically a wide range of parameters and formulae that may represent the potential for brain injury. Figure 5.3 shows the statistical probability of head injury ≥ AIS 3 for headform speed up to 80 km/h (22.3 $m.s^{-1}$). Hence it includes velocities normal to the helmet surface somewhat greater than those used by SHARP and hence illustrates that the correlation of injury potential with velocity normal to the helmet surface was sufficiently well known to justify the combination of normal and tangential velocity as used by SHARP. Dr Mills criticises SHARP for apparent extrapolation of data beyond 12 $m.s^{-1}$ for the oblique impact test velocity and whilst the above graph does not relate specifically to oblique impacts it is sufficiently general to include such impacts with a velocity well above 12 $m.s^{-1}$.

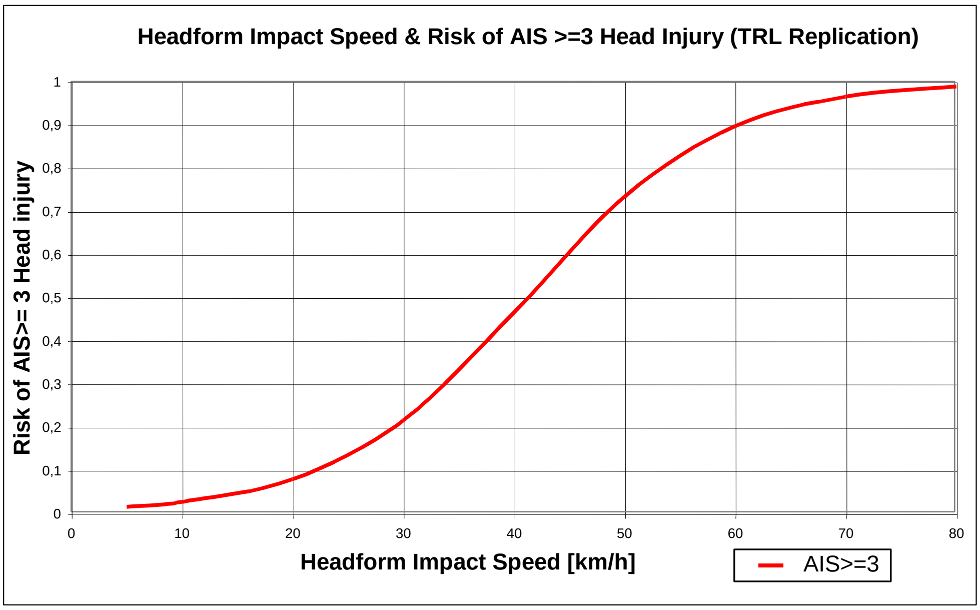


Figure 5.3: Headform impact speed vs probability of a head injury ≥ AIS 3 (COST 327 data)

5.5 Test Procedures

The research of the test procedures working group included the investigation of the correlation of rotational acceleration and tangential force measured in the oblique impact test. A Hybrid II dummy headform was used for the initial series of tests. Four different helmet types were used at five different velocities, ranging from 6.0 to 12.0 $m.s^{-1}$. This range considerably exceeds the velocity range for the tests in SHARP, which are from 6.0 to 8.5 $m.s^{-1}$ for the linear impact tests and at 8.5 $m.s^{-1}$ for the oblique impact tests. The results are given in Figure 5.4 for which a correlation coefficient of 0.94 was calculated.

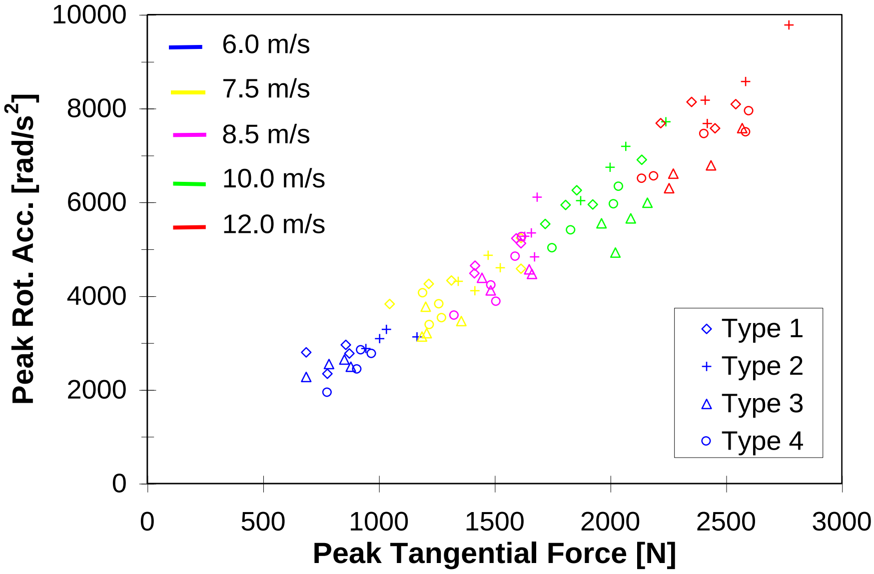


Figure 5.4: Peak rotational acceleration versus peak tangential force for impacts of a helmeted dummy headform onto the abrasive anvil

Similar tests were undertaken with a full dummy at impact velocities from 4.4 to 6.0 m.s^{-1}, and the results are given in Figure 5.5 for which the correlation coefficient was 0.90. The combined headform and dummy results are shown in Figure 5.6.

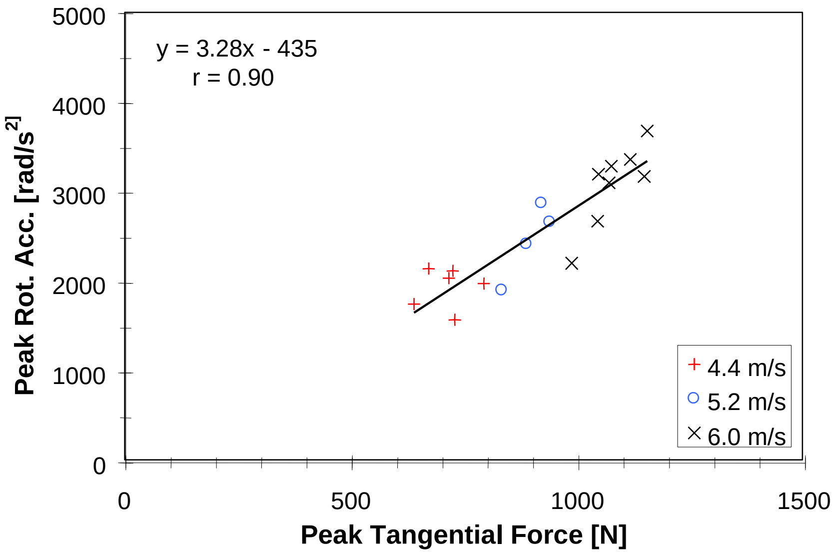


Figure 5.5: Peak rotational acceleration versus peak tangential force for dummy impacts onto the oblique abrasive anvil

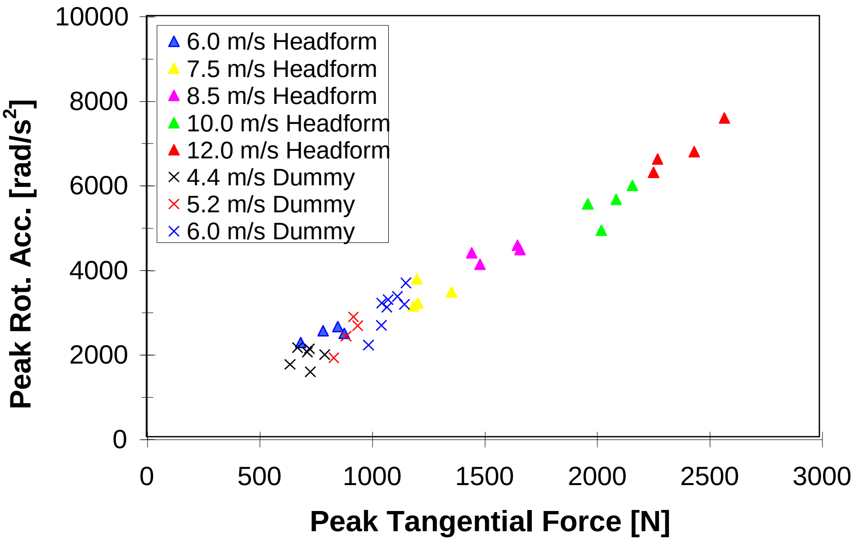


Figure 5.6: Peak rotational acceleration versus peak tangential force for dummy and headform impacts onto the oblique abrasive anvil

The excellent correlation shown in these tests is, when linked to the work of the human tolerance group a clear indication that peak tangential force can be correlated directly with the potential for brain injury.

All of the above tests were with the grade 80 closed-coat aluminium oxide abrasive paper impact surface as specified in BS 6658 and Regulation 22.05. Dr Mills paper criticised its use as a friction surface for oblique motorcycle helmet tests. It is not clear why he, as chairman of the British Standards Institute committee responsible for BS 6658, does not support its use. SHARP has simply used a well-established test surface as a means of achieving repeatable and reproducible test results.

6 How SHARP Uses Test and Accident Data

6.1 Background

The SHARP performance evaluation protocol is based on information from work funded or co-funded by the DfT including the COST327, S100L/VF, S0232/VF and S0614/V8 projects. Additional information from other sources was also considered and used where appropriate.

The focus for the SHARP evaluation protocol is on encouraging improved motorcycle helmet safety, based on the most up-to-date understanding of real-world motorcyclist accidents and head injuries.

The structure of the performance evaluation protocol is such that if new information becomes available, this can easily be incorporated in the performance evaluation protocol for future phases. For example, an updated injury risk function for peak head acceleration or distribution of helmet impact sites for fatal injuries may be published. If these are considered to be better than the functions and distributions used here, they could be substituted for the current functions quite easily.

6.2 Overview of the SHARP Helmet Test Programme

6.2.1 Test Matrix

Thirty-two impact tests per helmet model (thirty linear and two oblique) are conducted for the DfT by its test contractor. The linear tests are undertaken at three impact energies, with impacts at each energy level being performed with a different helmet size (medium, large and extra-large). Each helmet is fitted with the appropriate headform (nominally size J, M and O respectively). Where the nominal headform size is found to be too large for the helmet, the next smaller size of headform is used. To compensate for the lower mass of the smaller headform, the impact velocity is increased to give the same impact energy for all helmets of a given size. The test matrix for a single helmet model is shown in Table 6.1 overleaf.

6.2.2 Evaluation Parameters

Based on the recommendations from the previous research summarised in the earlier chapters of this report, as well as a statistical analysis of the discriminative ability of a number of criteria that was conducted on the results of the first 28 helmet models that were tested in SHARP, the following parameters are used in the assessment of each helmet's performance:

- Peak vertical headform acceleration from the linear impacts;
- Peak tangential anvil force measured in the oblique impacts;
- Normal anvil force measure in the oblique impacts at the time of peak tangential force.
- The tangential and normal force measurements are used to determine the coefficient of friction in the oblique impacts.

Table 6.1: Helmet test matrix

Nominal impact speed ($m.s^{-1}$)	**6**	**7.5**	**8.5**
Helmet size	**M**	**XL**	**L**
Nominal headform size	**J**	**O**	**M**
Impact location and anvil		**Test number**	
Flat Front	1	6	11
Flat Side L	2	7	12
Flat Side R	3	8	13
Flat Crown	4	9	14
Flat Rear	5	10	15
Kerb Front	16	21	26
Kerb Side L	17	22	27
Kerb Side R	18	23	28
Kerb Crown	19	24	29
Kerb Rear	20	25	30

Oblique	**Medium**	**J Headform**	**8.5 $m.s^{-1}$**
Side L		31	
Side R		32	

6.3 The SHARP Performance Evaluation Protocol

An outline of the evaluation protocol is shown in Figure 6.1. It can be split into two main stages:

1. Injury risks are estimated for each helmet based on injury and accident statistics and a number of injuries for each helmet is calculated using this risk and an appropriate population (the blue boxes in the figure);
2. Helmets are rated based on this injury number and the current injury statistics for the UK (the green box in the figure).

A detailed description of the test and evaluation protocols can be found in the SHARP Technical Manual that it is understood that the DfT will publish. This section of the report focuses on the remaining issues raised in the paper by Dr Mills that have not been addressed in earlier chapters.

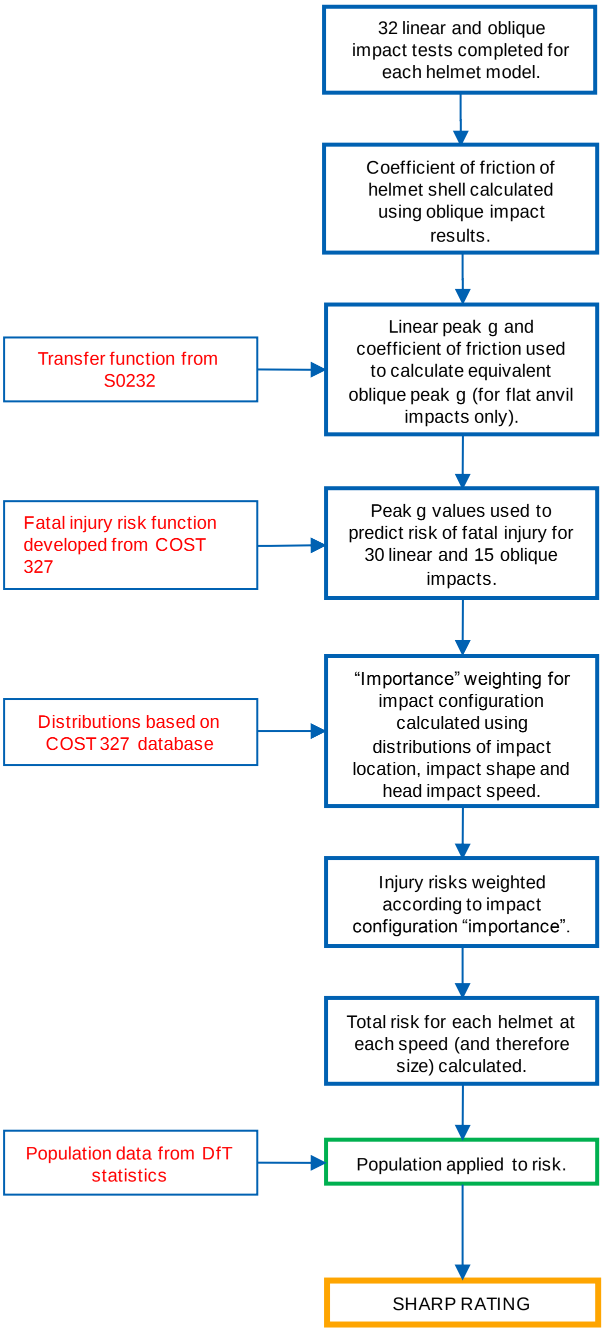


Figure 6.1: SHARP rating procedure flow chart

6.3.1 *Injury Risk Function*

SHARP uses a head injury risk function based on accident replications (see Section 5.3) undertaken by TRL as part of the COST 327 [Chinn *et al.*, 2001] and S0232 [Mellor *et al.*, 2007] projects. This injury risk function estimates the likelihood of fatality, serious injury and slight injury according to a maximum acceleration sustained by the headform in an impact and is shown in Figure 6.2.

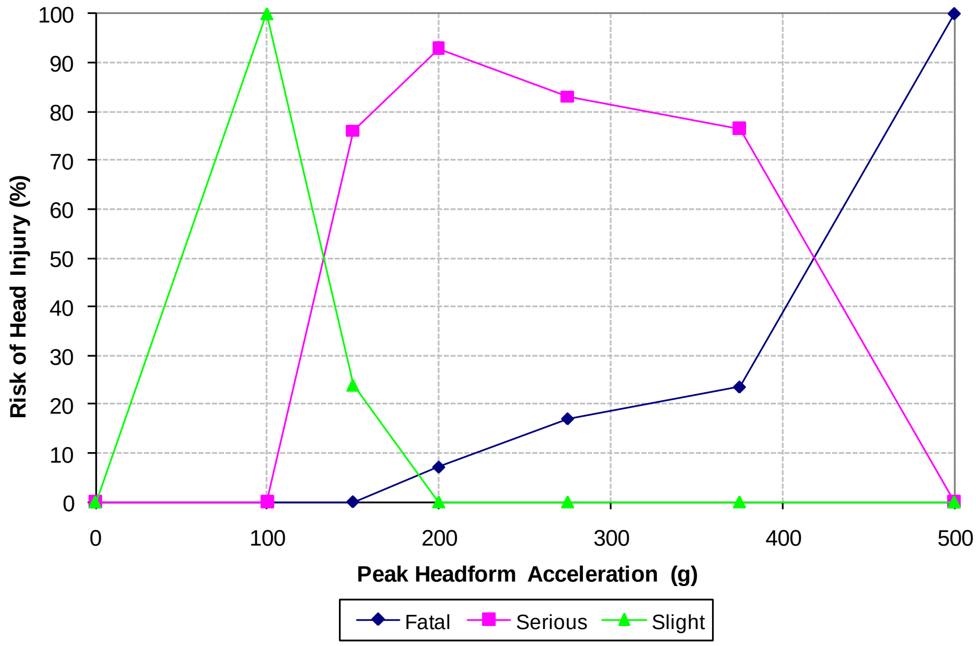


Figure 6.2: Risk of injury severity according to peak *g* (based on COST 327 and TRL S0232)

The risk functions above show that for peak headform accelerations of less than 100 *g*, only slight or no injuries are expected. Above a peak headform acceleration of 100 *g*, a slight, serious or fatal injury is predicted, depending on the peak headform acceleration. It should be remembered that these risk functions are derived from a sample of more than 20 motorcyclist accident reconstructions. A relatively small sample such as this is unlikely to be perfectly representative of the whole population of motorcyclists, but is currently the most comprehensive guide to the risk of serious and fatal head injury to helmeted motorcyclists. It should also be noted that people have a wide range of tolerance to impact loading due to factors such as age, sex, stature, fitness. For example, a young, fit male would be expected to have a greater tolerance to impact forces than an elderly, sedentary female This has been shown through biomechanical tests on, for example, the lower leg as shown in Figure 6.3 from [Funk *et al.*, 2001].

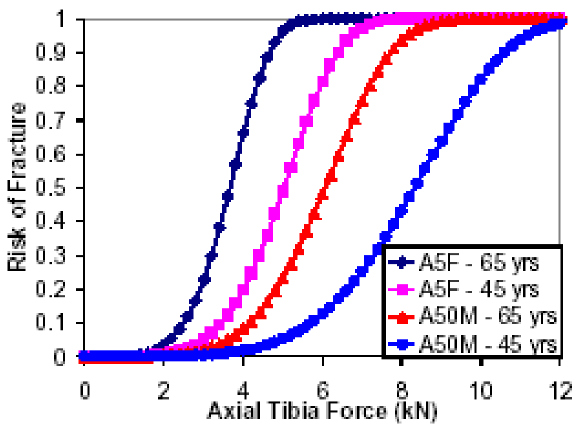


Figure 6.3: Injury risk functions for the American 5th percentile female and American 50th percentile male at two different ages assuming no Achilles tension [Funk *et al.*, 2001]

This natural variation in injury tolerance across the population leads to a (small) risk of fatal head injury at relatively low headform accelerations (e.g. the 7.1% risk of fatal injury at 200 *g* shown in Figure 6.2). Similar considerations apply throughout the risk range. This variation in injury tolerance within the population may explain why studies (such as those quoted by Dr Mills) on groups of young, healthy males exhibit higher tolerances to injury.

6.3.2 Helmet Fit

Dr Mills correctly notes that helmet fit is not assessed in the SHARP evaluation protocol. SHARP uses industry-standard head forms, defined in published international standards, when evaluating the helmets. These head forms are in general use for both scientific and regulatory purposes.

SHARP acknowledges the importance of a helmet's fit in order for it to afford good protection. It is also recognised that fit is particular to the individual. SHARP use their literature and web site to make it clear that the first step in choosing a helmet is to find one with a good fit. SHARP provides guidance on how to choose a comfortable and well fitting helmet; it is understood that this guidance, including the web-based video, was created in partnership with the helmet manufacturers and the supply industry and has been approved by them.

6.3.3 Rotational Impact Evaluation

The rotational impact model has been described in Section 4.2. This model is used to calculate an equivalent oblique impact acceleration for each of the flat anvil tests, and the risk of fatal head injury for each of these results is determined using the injury risk function in Section 6.3.1. Finally, a weighting for the risk of sustaining a head impact at this velocity is determined (see Section 6.3.4).

Rotational acceleration will vary with helmet size, shape and mass, but the strategy for improving the performance is the same - to lower the friction in the impact and to lower the normal reaction force. As noted in Section 6.2.2, these are the main parameters used in the SHARP evaluation, and SHARP strongly encourages a reduction in both.

Furthermore, the direct measurement of head rotational acceleration is normally made using a nine-axis accelerometer array. This equipment is not generally available in motorcycle helmet test or development laboratories. It was considered preferable to use standard equipment that is available in all helmet test and development laboratories, and used in motorcycle helmet standards and Regulations that apply in the UK, whilst ensuring that this standard equipment encourages the appropriate improvements in helmet design.

6.3.4 Velocity Weighting for the Oblique Equivalent Accelerations

Section 8 of Dr Mills paper suggests that the oblique impact tests in SHARP are undertaken at an impact angle of 37.5°. This is not the case - the oblique impacts use a 15° anvil, which is the test condition used in Regulation 22.05 and BS 6658:1985. The value of 37.5° is used simply to determine the *velocity weighting* for the oblique equivalent tests; that is, the risk of a motorcyclist having an accident that resulted in an oblique impact with a normal velocity of 6, 7.5 or 8.5 $m.s^{-1}$ (and therefore a resultant velocity of 9.9, 12.3 and 14.0 $m.s^{-1}$, as shown in Figure 6.4).

Based on COST 327 data, 37.5° was found to be a typical angle between the head and the impacted surface for oblique impacts. (It should be noted that the tables in COST 327 do not give this value directly, so it is derived from the combination of body angle to the horizontal, and the head impact angle relative to the body.) It is worth noting that the cumulative velocity distribution is reasonably linear between about 5 and 17 $m.s^{-1}$. A sensitivity analysis was undertaken during the development of the SHARP evaluation protocol that showed that the rank order of the helmets was barely affected if a completely linear cumulative velocity distribution was used instead of the distribution determined by COST 327, as shown in Figure 6.5.

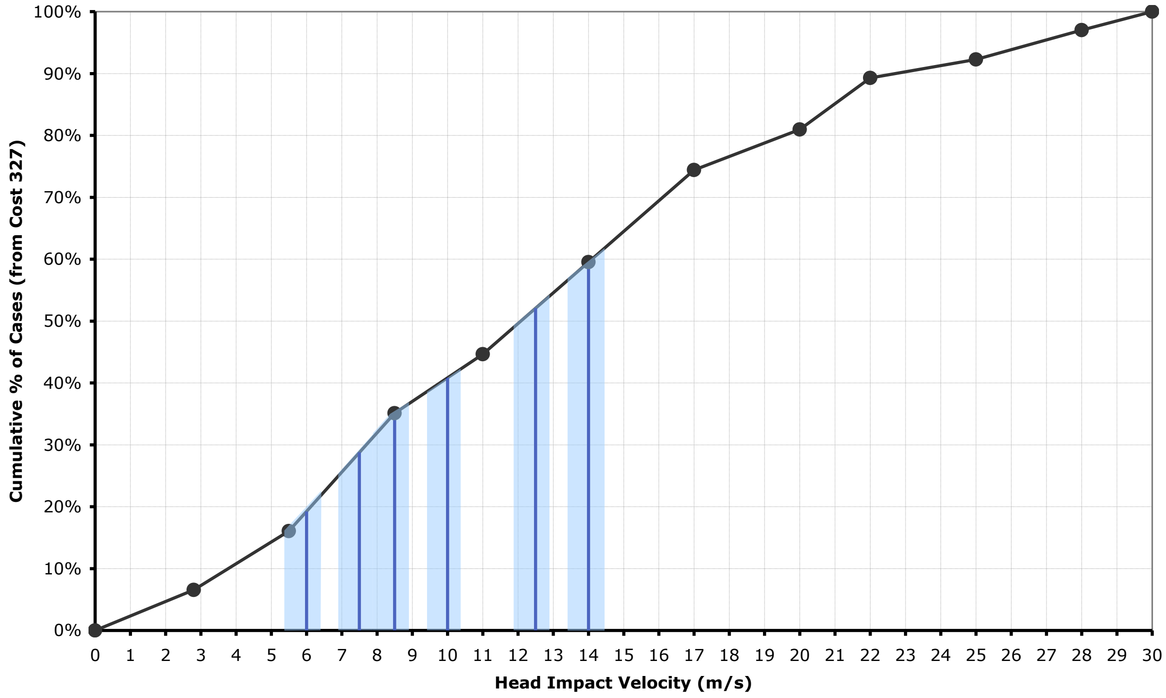


Figure 6.4: Cumulative head impact velocity distribution (from COST 327)

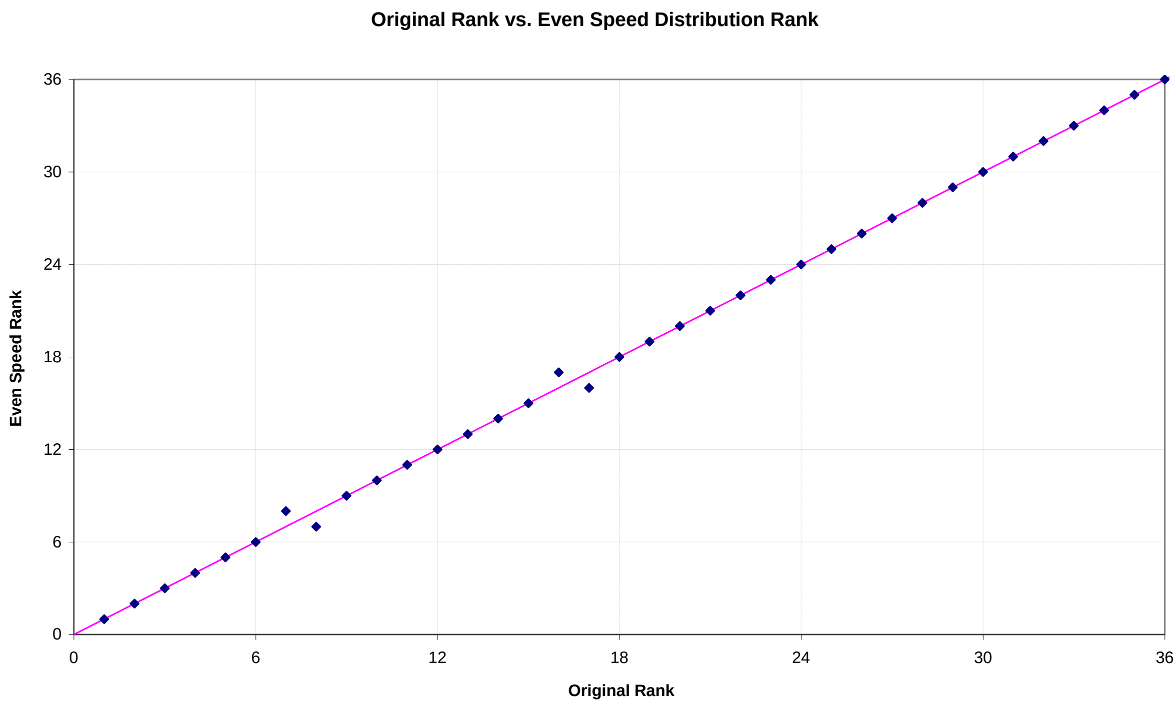


Figure 6.5: Linear head impact velocity distribution helmet ranking vs. COST 327 accident data head impact velocity distribution helmet ranking

7 Discussion

Based on the current understanding of head injury mechanisms (see Section 2), there are three main factors that, for a given person and head impact site, affect the risk of serious head injury:

- Distribution of impact forces;
- Linear head acceleration; and
- Rotational head acceleration.

A motorcycle helmet can affect these factors by dissipating energy through the shell and liner, and by reducing the coefficient of friction between the helmet and the impact surface. The shell and liner help to distribute the load over wider area, which reduces the risk of fracture, and reduces the linear head acceleration, which lowers the risk of focal brain injury. The shell and liner also reduces both of the components of force that produce rotational head acceleration, which reduces the risk of diffuse brain injury such as DAI. Reducing the coefficient of friction lowers the second component of force that produces rotational acceleration, which further reduces the risk of diffuse brain injury.

The SHARP motorcycle helmet assessment and rating programmes encourages and rewards improvements in each of these facets of helmet design using a combination of 32 flat plate, kerb-type and oblique impact tests per helmet model. No single test type dominates the assessment, so that a rounded approach to motorcycle helmet safety is encouraged.

Furthermore, the tests are weighted according to the best available motorcycle accident data to ensure that improvements are targeted to where they will make the greatest difference to motorcyclist safety. In addition, a capping system is used to ensure that a very good rating (five stars) is not achievable if a helmet has a poor result in a test with a low weighting. This ensures that five star helmets must have good all-round performance.

Dr Mills has recently reported that the four main parameters affecting rotational acceleration in a motorcycle helmet/head impact are: linear head acceleration, impact velocity normal to the road, the friction coefficient between the shell and road, and the impact site/direction [Mills *et al.*, 2009]. SHARP encourages improvements in linear head acceleration at a range of normal velocities and impact sites, including tests at higher velocities than used in current European motorcycle helmet regulation. Furthermore, this is combined with assessment of the coefficient of friction between the shell and an idealised road surface (alumina paper, which is used in the British Standard test developed by the committee that Dr Mills Chairs, and which is used in the Method A oblique test in UNECE Regulation 22.05).

The strong linear relationship between headform rotational acceleration and peak tangential force in oblique motorcycle helmet drop tests was clearly demonstrated in the COST 327 project (see Figure 5.4) for a range of helmet types (with different geometries, linear impact performance and coefficients of friction), tested at velocities of 6.0, 7.5, 8.5, 10.0 and 12.0 $m.s^{-1}$. These tests included both sliding and rolling of the helmet during the impact, as all practicable helmet impacts must. The tests were very similar to the tests conducted in SHARP, and the tests in COST 327 covered a much wider range of impact speeds than SHARP. Such wide-ranging data is very important for the validation of the SHARP evaluation protocol. It should be noted that these data were from more than 80 real tests on real production motorcycle helmets, not estimates from a finite element model of a single helmet type.

Dr Mills reports that Glaister [1996] and Mills *et al.* [2009] both argued that the simplest method to reduce the risk of head injury due to rotational acceleration would be to reduce the peak linear head acceleration in direct impact tests. The larger proportion of the SHARP rating is attributable to the linear impact performance; this may be expected given that 40% of the impact type distribution is purely linear and 60% is oblique, in which potentially injurious head accelerations are caused by a combination of linear force and the coefficient of friction with the impact surface. Nevertheless, reducing the coefficient of friction can yield considerable additional benefits in reducing the risk of rotational brain injury, such as DAI, and is therefore also strongly encouraged by SHARP.

In his discussion, Dr Mills contends that the probability of impact to the side site X of Regulation 22.05 is too high because this site is rarely hit in practise. Dr Mills provides no evidence for this contention, and it is clear from a correct interpretation of both the impact site and injury statistics (as discussed in Section 5) that impact and injury at the sides is common and is appropriately proportioned in SHARP. This is further supported by the helmet damage descriptions and sketches in the Accident Reconstruction reports from COST 327.

Dr Mills notes that if a helmet has been designed to pass the 7.5 $m.s^{-1}$ linear impact tests by a small margin (e.g. 260 *g* when the maximum allowed headform acceleration is 275 *g*), then the liner may bottom-out when tested at 8.5 $m.s^{-1}$ and give a peak linear acceleration greater than 300 *g*. He also notes that such a result could strongly influence the SHARP rating; this is apparently intended as a criticism of the scheme.

In fact, previous research at TRL has shown that this is exactly what happens with many motorcycle helmets when tested at 8.5 $m.s^{-1}$ [Mellor *et al.*, 2007], with some helmets greatly exceeding 300 *g* and therefore exposing the wearer to a high risk of fatal injury. The same research also demonstrated that it was perfectly feasible to design a helmet that performed well at 8.5 $m.s^{-1}$ whilst maintaining good performance at 6 and 7.5 $m.s^{-1}$ (see Figure 7.1 and Figure 7.2). It is this desirable - and achievable - increased range of protection that is encouraged by the SHARP rating scheme.

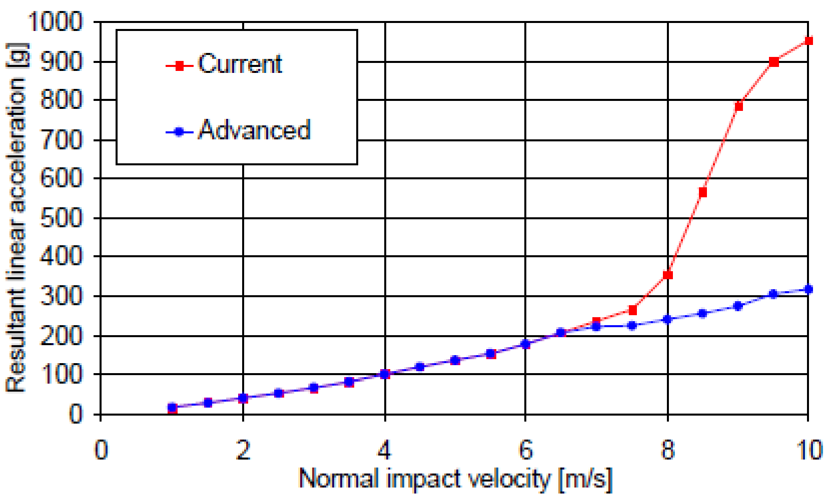


Figure 7.1: Linear impact performance for a typical current motorcycle helmet (red line) compared with the TRL advanced helmet design (blue line) [Mellor *et al.*, 2007]

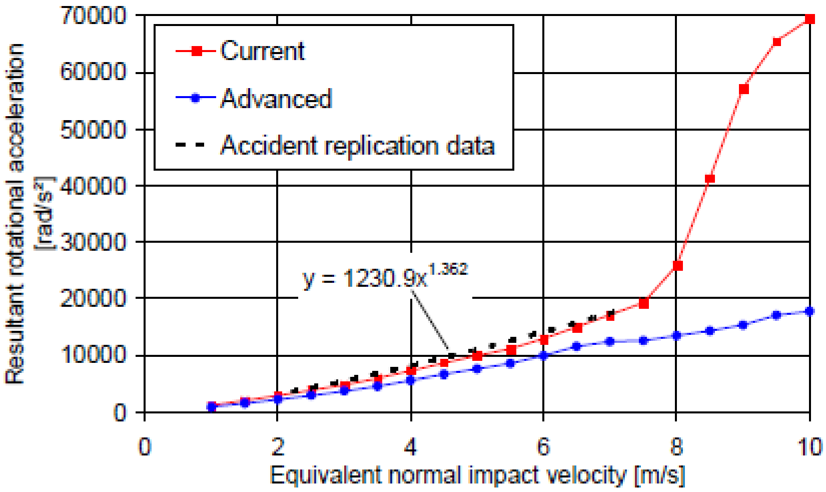


Figure 7.2: Rotational impact performance for a typical current motorcycle helmet (red line) compared with the TRL advanced helmet design (blue line) [Mellor *et al.*, 2007]

8 Conclusions

1. The primary evidence sources for the SHARP protocols are the 'European Co-operation in the Field of Scientific and Technical Research, Action 327' (COST 327), 'New Helmet Designs: Performance Assessment and Cost Benefit Analysis' (S100/L) and 'Motorcyclists' Helmets and Visors – Test Methods and New Technologies' (S0232). COST 327 was a Europe-wide Action on motorcycle accidents, with a focus on head and neck injuries.
2. Research reported in COST 327 and summarised in this report has shown that rapid motions in a head impacts can cause severe and fatal head injury, including cranium fracture, focal brain injury, and diffuse brain injury (often referred to as diffuse axonal injury, or DAI).
3. SHARP has drawn upon the extensive research from the COST 327 Action to investigate the link between head and brain injury and the parameters used to measure helmet performance in the laboratory. It also used research from COST 327 to devise its test methods and protocols, using test equipment defined by current national and international Standards.
4. Reconstructions of over 20 motorcyclist head impact accident cases were undertaken in COST 327. Each head impact was reconstructed several times until the damage to the test helmet matched the damage to the original helmet worn in the accident. The headform accelerations and other parameters measured in these reconstructions were correlated with head injury information from histological examinations or CT scans performed by a Consultant Neuropathologist. Head injury risk functions for different severities of injury were then determined.
5. Extensive theoretical studies to investigate the effect of, for example, friction between the helmet shell and the surface struck, offset of the centre of gravity, and moment of inertia of the head/helmet, have been reported. This data were compared with practical results from oblique helmet tests similar to those undertaken in SHARP, to demonstrate that the approach implemented by SHARP is consistent with helmet test results.
6. The SHARP motorcycle helmet safety evaluation protocol strongly encourages helmet designs that reduce the linear acceleration in a head impact. This reduces the risk of cranium fracture and localised brain injury. It also reduces the risk of diffuse brain injury due to rotational acceleration. SHARP also strongly encourages helmets with a lower coefficient of friction, which additionally reduces the risk of rotational brain injuries.
7. Accident analysis from COST 327 was used to weight the SHARP test results to ensure that improvements in helmet design are targeted at reducing the risk of fatal head injury in the most common accident scenarios. Weightings for impact location on the helmet (i.e. front, side, rear and crown), impact type (i.e. oblique, flat and kerb-type impacts), and the risk of having a head impact at a given speed are used.
8. In order to determine impact location distribution, COST 327 divided the helmet in to eight zones radially when viewed from above. This enabled the impact location distribution used in SHARP to be determined very accurately. These data showed that the location of impacts around the helmet was fairly even with 26.9% lateral right, 26.3% lateral left, 23.6% frontal and 21.0% to the rear. The crown, received only 2.2% of the impacts.
9. The impact type distributions detailed in COST 327 were used to define the proportion of motorcyclist head impacts that involve oblique loading, and the proportion of the remaining impacts that are best represented by standard flat and kerb-type tests used in Regulations and Standards.

10. Good helmet fit is also an important component of motorcycle helmet safety. SHARP provides guidance on how to choose a comfortable and well fitting helmet; it is understood that this guidance, including the web-based video, was created in partnership with the helmet manufacturers and the supply industry and has been approved by them.

Acknowledgements

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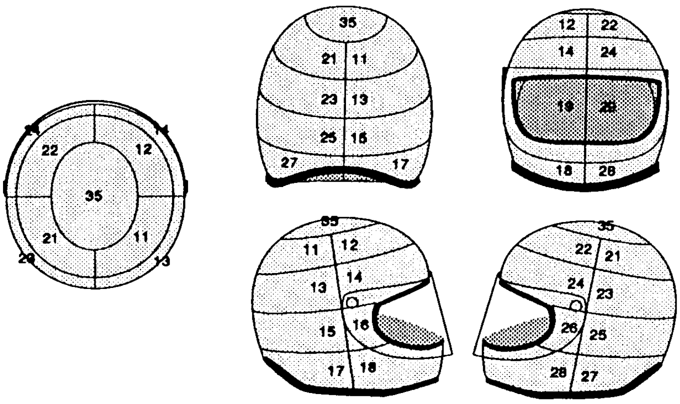
Appendix A Helmet Deformation and Head Injuries

Table A.1 shows the detailed categorisation of helmet impact location that was used in COST 327. This table is from the interim report of the COST 327 Task Group on Accident Data [Otte *et al.*, 1998], and shows less than half of the total number of cases presented in the COST 327 final report, but clearly shows that the helmet is divided in to eight zones radially when viewed from above. For instance, Section 16 on the helmet is part of the right front quadrant of the helmet when viewed from above. It is further divided in to 'Section 16 lateral' and 'Section 16 frontal', which gives a much more precise definition of the impact locations.

Table A.1: Helmet deformation and head injuries: outside defects of the helmet



location of helmet	total n	total %	deformation n	deformation %	laceration n	laceration %	crack n	crack %	other n	other %
Sec. 35	8	2,4	-	-	7	87,5	1	12,5	-	-
lateral right										
Sec. 11 lateral	10	3,0	-	-	8	80,0	1	10,0	1	10,0
Sec. 12 lateral	9	2,7	-	-	8	88,9	1	11,1	-	-
Sec. 13 lateral	14	4,2	-	-	12	85,8	1	7,1	1	7,1
Sec. 14 lateral	19	5,6	2	10,5	15	79,0	2	10,5	-	-
Sec. 15 lateral	12	3,5	3	25,0	4	33,3	5	41,7	-	-
Sec. 16 lateral	10	3,0	2	20,0	4	40,0	4	40,0	-	-
Sec. 17 lateral	5	1,5	1	20,0	3	60,0	1	20,0	-	-
Sec. 18 lateral	11	3,3	3	27,3	7	63,6	1	9,0	-	-
lateral left										
Sec. 19 lateral	9	2,7	3	33,3	3	33,3	3	33,3	-	-
Sec. 21 lateral	6	1,8	1	16,7	4	66,6	1	16,7	-	-
Sec. 22 lateral	6	1,8	-	-	4	66,7	2	33,3	-	-
Sec. 23 lateral	12	3,5	3	25,0	8	66,7	1	8,3	-	-
Sec. 24 lateral	14	4,2	2	14,3	10	71,4	2	14,3	-	-
Sec. 25 lateral	10	3,0	2	20,0	8	80,0	-	-	-	-
Sec. 26 lateral	10	3,0	1	10,0	7	70,0	2	20,0	-	-
Sec. 27 lateral	5	1,5	1	20,0	4	80,0	-	-	-	-
Sec. 28 lateral	7	2,1	1	14,3	4	57,1	2	28,6	-	-
Sec. 29 lateral	7	2,1	-	-	3	42,9	4	57,1	-	-
frontal										
Sec. 12 frontal	6	1,8	-	-	5	83,3	1	16,7	-	-
Sec. 14 frontal	6	1,8	2	33,3	4	66,7	-	-	-	-
Sec. 16 frontal	9	2,7	-	-	4	44,4	5	55,6	-	-
Sec. 18 frontal	15	4,4	2	13,3	11	77,4	2	13,3	-	-
Sec. 19 frontal	8	2,4	1	12,5	3	37,5	4	50,0	-	-
Sec. 22 frontal	4	1,2	-	-	3	75,0	1	25,0	-	-
Sec. 24 frontal	9	2,7	4	44,4	4	44,4	1	11,2	-	-
Sec. 26 frontal	2	0,6	-	-	2	100,0	-	-	-	-
Sec. 28 frontal	8	2,4	1	12,5	6	75,0	1	12,5	-	-
Sec. 29 frontal	8	2,4	2	25,0	4	50,0	2	25,0	-	-
rear										
Sec. 11 rear	12	3,5	-	-	10	83,3	1	8,3	1	8,3
Sec. 13 rear	13	3,9	-	-	10	76,9	2	15,4	1	7,7
Sec. 15 rear	13	3,9	1	7,7	10	76,9	2	15,4	-	-
Sec. 17 rear	4	1,2	1	25,0	3	75,0	-	-	-	-
Sec. 21 rear	5	1,5	-	-	5	100,0	-	-	-	-
Sec. 23 rear	13	3,9	-	-	12	92,3	1	7,7	-	-
Sec. 25 rear	14	4,2	1	7,1	13	92,8	1	7,1	-	-
Sec. 27 rear	2	0,6	-	-	2	100,0	-	-	-	-
total	**335**	**100,0**	**40**	**11,9**	**233**	**69,6**	**58**	**17,3**	**4**	**1,2**

Appendix B Summary Table of AIS Scale with Head Injury

Table B.2 shows some examples of the key types of cranium and brain injury and their related AIS value. The full AIS head injury coding for the head (not including the face) contains nearly 200 detailed physical head injury codes and nearly 40 additional codes relating to loss of consciousness and neurological deficit. It provides a very comprehensive and detailed record of the head injuries sustained in an accident.

Table B.2: Overview of some key head injuries and how they relate to the AIS 1990 scale

	AIS 0 Uninjured	AIS 1 Minor	AIS2 Moderate	AIS 3 Serious	AIS 4 Severe	AIS 5 Critical	AIS 6 Maximum
Scalp							
: superficial abrasions, contusions, lacerations		X					
: major laceration or minor blood loss			X				
: blood loss >20% or total scalp loss				X			
Intracranial vessels (arteries)							
: laceration					X	X	
Cranial nerves							
: contusion, laceration, loss of function			X				
Brain							
: swelling, contusions, haemorrhage				X			
: haematoma, large >30cc contusion					X		
: massive >50cc contusions, diffuse axonal injury, large haematoma						X	
: brain stem crush, penetrating injury							X
Loss of consciousness							
: < 1 hour			X				
: 1 - 6 hours or < 1 hour with neurological deficit				X			
: 6 – 24 hours, or 1-6 hours with neurological deficit					X		
: > 24 hours, or 6-24 hours with neurological deficit						X	
Skull (base or vault of cranium) fracture							
: simple vault fracture			X				
: compound vault or simple base of skull fracture				X			
: complex, open, loss of brain tissue (vault and base)					X		

Technical Response to the Unpublished Paper 'Critical Evaluation of the SHARP Motorcycle Helmet Rating' by NJ Mills

The SHARP motorcycle helmet safety rating scheme was launched by the Department for Transport (DfT) in June 2008 to provide motorcyclists with objective information on the impact protection offered by motorcycle helmets in the event of an accident. The rating scheme is based on considerable previous research regarding head injury mechanisms, motorcycle accident investigations, motorcyclist head impact accident reconstructions, and the development of an advanced helmet that demonstrated the potential for considerable improvement in the protection offered by helmets.

Recently, an unpublished paper (Critical Evaluation of the SHARP Motorcycle Helmet Rating, by NJ Mills) has criticised the approach taken by SHARP to the rating of helmet impact performance. The authors have been asked by the DfT to provide a technical response to this unpublished paper, which is contained in the present report.

The primary evidence sources for the SHARP protocols are the 'European Co-operation in the Field of Scientific and Technical Research, Action 327' (COST 327), 'New Helmet Designs: Performance Assessment and Cost Benefit Analysis' (S100/L) and 'Motorcyclists' Helmets and Visors – Test Methods and New Technologies' (S0232).

It is understood that the DfT will publish a separate paper that describes the details of the SHARP test and evaluation protocols. It is not the purpose of this report to duplicate this exercise, but instead to explain the technical foundations of SHARP and how they relate to real-world motorcycle accidents.

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