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ASSESSMENT OF CURRENT BICYCLE HELMETS FOR THE POTENTIAL TO CAUSE ROTATIONAL INJURY

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EXECUTIVE SUMMARY

The UK Department for Transport believe that it is sensible for cyclists, and especially children, to protect themselves by wearing a bicycle helmet. However concern has been expressed that current bicycle helmets may increase the risk of brain injuries from rotational motion during head impacts. In order to evaluate these concerns, an experimental study has been completed to test a range of current UK bicycle helmets to investigate the potential for head injuries as a result of both linear and oblique impacts.

Although pedal cyclist casualties represent a small proportion of road user casualties in Great Britain forming just over 4% of those killed and just under 7% of those seriously injured in the year 2004[1], of the killed and seriously injured cyclists, 25% are children. This study was therefore focused on helmets suitable for use by children. Consequently, of eight bicycle helmet models chosen to give a representative selection of those available in the UK, four of the chosen helmets were child style models and four were youth / adult helmets.

In real-life bicycle accidents there are many factors that may affect helmet performance and the head injury severity. These include the head impact velocity relative to the object struck and the shape and stiffness of this object. A helmet can absorb impact energy and potentially reduce the injury severity but can also spread the impact force thereby reducing the risk of skull fracture. However, some helmet features, including the increased mass and effective diameter, may increase the potential for head injuries due to rotational motion and acceleration of the head. The literature review, completed as part of this project and summarised in this report, has identified that rotation of the head is also a potential mechanism for, and a significant cause of, brain injury. It has been important therefore to assess both the linear and oblique impact performance to understand this injury potential.

To assess linear impact performance, the most recent bicycle helmet standards specify a drop test onto a rigid anvil using an instrumented headform fitted with the test helmet. Criteria are set to ensure that the injurious linear accelerations are kept below an appropriate safety threshold. The standards do not assess how closely helmets meet these requirements and tests conforming to EN1078, to which all helmets sold in UK must comply, are therefore necessary to assess the real injury potential for such typical linear head impacts.

None of the current bicycle helmet standards include an assessment of oblique impacts or the potential for rotational injury. Neither do the standards stipulate restrictions to helmet design which may eliminate the most hazardous features e.g. by limiting projections to those of a considered safe geometry. However, a European standard for motorcycle helmets, UNECE Regulation 22.05 (Reg 22.05), includes such a test which indirectly assesses and prescribes a performance requirement for rotational acceleration during an oblique impact. The Reg 22.05 oblique impact test method (Method A) was therefore chosen to assess the range of bicycle helmets to evaluate the potential for injurious rotational motion.

The helmet size was an important consideration for helmet selection so helmets were tested with two headform sizes to investigate the range of helmet performance. The headforms were in a range from 50cm to 57cm. Three of the helmets were intentionally tested with headforms up to 1cm outside the size range specified on the helmet. This was necessary to investigate the consequences of this ‘misuse’ since such a helmet and head size combination could occur in the real world where users may rely on comfort and other factors to determine appropriateness of fit.

More than thirty linear impact tests were completed using flat and kerb anvils at impact speeds of 4.57m/s and 5.42m/s respectively. The tests were in accordance with EN1078 but with test headforms and sites perceived to induce the greatest linear acceleration and therefore indicative of the ‘worst case’ injury potential. In five out of eight helmets the peak acceleration was below 200g, a level at which there is only a 2.1% risk of fatality and the risk of an AIS 3+ injury is low.

[1] Road Accidents Great Britain 2004

In a single instance the 250g limit of EN1078, which corresponds to a 10.6% risk of fatal injury, was exceeded. This helmet was tested using a headform over the manufacturer's specified size range by 1cm. This result demonstrates how helmets that meet the Standard may not provide optimal protection in a real-world application. Significantly, this result was some 116g higher than the next worst helmet performance for similar impact conditions. Such differences may equate to the difference between a critical and moderate severity head injury. Given a potential for users to wear helmets outside the manufacturer's specified range where fit allows, the consequence to injury outcome may be significant. The introduction of additional regulatory tests to standards, using a full range of headform sizes that can be fitted, may eliminate optimisation and detrimental helmet performance that may occur as a consequence of a poorly specified size range.

To investigate whether helmets provide improved protection compared to the non-helmeted human head in linear impacts, a comparison was made between helmet test data and post-mortem human surrogate test data. Figure 1 illustrates the extent of helmet performance across a speed range based on linear impact test data for size 54cm helmets. Simple and compound skull fracture tolerance data from post-mortem human surrogate tests are overlaid. Assuming that AIS and linear acceleration can be related using 50g interval steps (Newman, 1986) and that AIS can only be discrete integer values whereas linear acceleration is continuous, it can be seen that the best performing helmets reduce injury severity by at least 1 AIS increment between 2.2 and 3.2m/s. Above 3.2m/s the injury outcome using the best performing helmets is likely to be two or three increments lower than for a bare head and provides potentially life saving benefits.

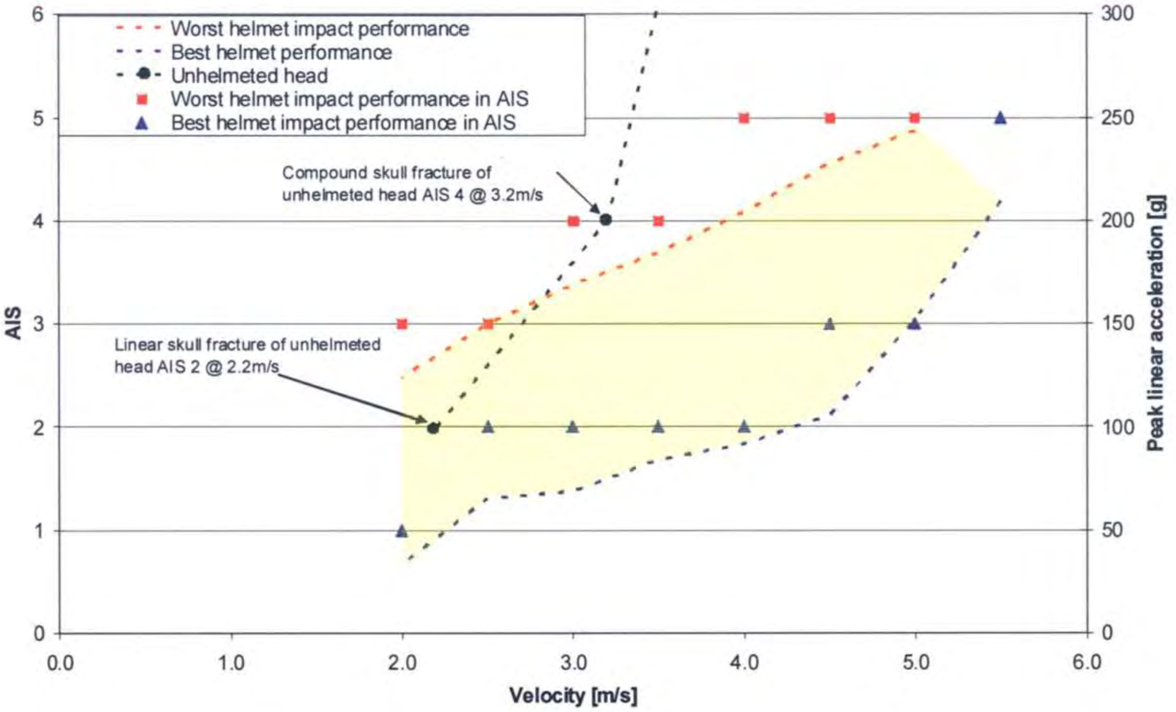

Figure 1

Even for the worst performing helmets, there are significant benefits above 3.2m/s of at least 1 AIS increment, which are, potentially, life saving. Below 2.8m/s the benefit of wearing a poor performing helmet was less definite and a detrimental injury outcome was feasible compared to a bare head. This analysis clearly shows that current helmets are optimised for protection at higher impact speeds and despite the potential for detrimental linear impact performance below 2.8m/s, the helmets can provide life saving protection during typical linear impacts.

Further to linear impact tests, thirty five oblique impact tests have been completed to ascertain the potential for head injury as a result of rotational motion. The impact speed was 8.5m/s onto a 15º anvil of either an abrasive surface to generate high frictional forces or a bar anvil with 6mm high raised edges which engaged with helmet projections. These conditions are likely to be the most severe presented during a bicycle accident and generate the 'worst case' rotational motion.

For all oblique impacts using a size 54cm instrumented headform, the maximum linear acceleration recorded was 98g which equates to a minor head injury of a severity below AIS 2. This was somewhat as expected as the normal component was equivalent to a linear impact at only 2.2m/s. The tangential anvil forces measured were below 3000N and an AIS 2 injury was predicted but with a risk of AIS 3 injury considerably less than 50%. This result indicates that the rotational motion during 15º oblique impacts could have a greater injury potential than the linear motion component.

Although tangential force correlates well with rotational acceleration, it could underestimate the potential for injury and consequently rotational acceleration was also measured and analysed. Using the size 54cm headform, the maximum rotational acceleration measured was 8,509rad/s² and this suggests that a most probable injury outcome of AIS 2 with the probability of AIS 3-6 injury below 35%. This agrees well with the estimates made using tangential force.

Since the response of the human head in an oblique impact is undocumented, the oblique experimental test data was related to post-mortem human surrogate linear impact data to investigate the potential for rotational injury for a helmeted and bare head. For a 15º oblique impact at 8.5m/s a simple skull fracture (AIS 2) would be predicted for a bare head. Given the onset of fracture to occur at a force of 5kN to 6kN, a linear acceleration of 100g and 120g would be required for a 5kg mass head, and 125g to 150g for a 4kg mass head. If the correlation between linear and rotational acceleration for a helmeted headform is assumed, a rotational acceleration of between 7500rad/s² and 12000rad/s² is predicted for a bare head, which is more injurious than the 3000rad/s² to 8500rad/s² measured during oblique tests with a size 54cm headform. However, tests completed with youth helmets and a size 57cm headform showed that in the most severe cases, rotational acceleration typically exceeded 10,000rad/s² and increased to 20,000rad/s², a level at which a 35% - 50% risk of serious AIS3+ injuries would be expected. In such cases, a disbenefit may therefore exist for some helmet wearers.

Linear impact performance, head inertia and helmet fit were all found to be important contributory factors to the level of induced rotational acceleration. Poor helmet fit was shown to reduce rotational accelerations where helmet mass is low but observations such as this were based on a very small number of tests and further studies using a fully instrumented headform would help to resolve the influence of these factors.

Overall, it was concluded that for the majority of cases considered, the helmet can provide life saving protection during typical linear impacts. The level of rotational acceleration observed using a helmeted headform were also found to be no more injurious than expected for a bare human head. However, for both low speed linear impacts and the most severe oblique impacts, linear and rotational accelerations may marginally increase the injury outcome that would be expected for an unhelmeted head. Since the true response of the bare head to oblique, glancing blows is unknown, these observations could not be concluded with certainty, but may be indicative of possible trends.

The design of helmets with a broad size range was also concluded to be detrimental to helmet safety, in terms both of reduced linear and rotational impact performance. Issues relating to helmet fit in regulations should therefore be reviewed. The introduction into EN1078 of a test for tangential force or rotational acceleration during an oblique impact should be considered to ensure that designs do not provide an excessive risk of injury as a consequence of rotation.

ABSTRACT

Concern has been expressed that current bicycle helmets may increase the risk of brain injuries from rotational motion. A range of child and youth bicycle helmets have therefore been tested to evaluate their linear and oblique impact performance This data was used to assess the propensity of the helmet to influence rotational motion and was considered against post-mortem human surrogate data to allow comparison of the risk of injury to that of an unhelmeted head.

Un-helmeted post-mortem human surrogate data indicates that a simple skull fracture for an unhelmeted head (injury rated as AIS 2) may occur at 5kN - 6kN which corresponds to between 100g and 150g for a head mass of between 4kg and 5kg. Assuming that the response of the unhelmeted head is similar to the helmeted head an oblique impact at 8.5m/s at 15º, this may generate between 7500rad/s² and 12000rad/s² of rotational acceleration. This is potentially more severe than the 3000rad/s² to 8500rad/s² measured during abrasive and projection oblique tests with size 54cm (E) helmeted headforms. However, for the most severe cases using a size 57cm (J) headform, rotational acceleration was typically greater than 10,000rad/s² and increased to levels of 20,000rad/s², a level at which a 35% - 50% risk of serious AIS3+ injuries is anticipated.

Overall, it was concluded that for the majority of cases considered, the helmet can provide life saving protection during typical linear impacts and, in addition, the typical level of rotational acceleration observed using a helmeted headform would generally be no more injurious than expected for a bare human head. However, in both low speed linear impacts and the most severe oblique cases, linear and rotational accelerations may increase to levels corresponding to injury severities as high as AIS 2 or 3, at which a marginal increase (up to 1 AIS interval) in injury outcome may be expected for a helmeted head.

The true response of the bare human head to oblique, glancing blows is not known and these observations could not be concluded with certainty, but may be indicative of possible trends. A greater understanding is therefore needed to allow an accurate assessment of injury tolerance in oblique impacts. Linear impact performance, head inertia and helmet fit were identified as important contributory factors to the level of induced rotational motion and injury potential. The design of helmets to include a broad range of sizes was also concluded to be detrimental to helmet safety, in terms of both reduced linear and rotational impact performance. The introduction into EN1078 of an oblique impact test could ensure that helmets do not provide an excessive risk of rotational head injury.

1 INTRODUCTION

TRL is pleased to have been awarded the Department for Transport (DfT) contract to investigate the occurrence of, and potential for, head injuries caused by rotational motion in bicycle accidents, especially to children of 16 years and younger.

Pedal cyclist casualties are a small proportion of road user casualties in Great Britain forming just over 4% of those killed and just under 7% of those seriously injured in the year 2004 (RAGB 2004). However, of the killed and seriously injured cyclists, 25% are children.

Although bicycle helmets can help to prevent or mitigate head injuries in many accidents, it is believed that there may be inadequate knowledge about the effects in certain circumstances. There is particular concern about the effects of helmets when the head is rotated sharply, this can occur for example during a glancing impact with the road surface. This is of concern because the brain may be injured if rotation occurs within the skull. TRL was therefore commissioned by the Department for Transport to clarify the known evidence on bicycle helmets and rotational injuries.

The study was conducted in two phases, with the first phase being a literature review and technical appraisal to investigate what is known about rotational head injuries. This literature review has been reported separately and only a summary of it is included here.

The second phase was a technical appraisal of the issues relating to rotational head injuries resulting from head impact. This appraisal was based on existing knowledge and further knowledge gained from testing a range of current bicycle helmets. This included the use of a headform, within the helmet, fitted with a nine-accelerometer array to assess rotational motion during a glancing impact. The results of this second phase of the study are reported in detail in this report.

2 LITERATURE REVIEW AND LEGISLATION

2.1 Literature review

The TRL literature review has been reported in full by Chinn (2004); the executive summary is reproduced below:

This report is a literature review concerning the effect of rotational motion on the head and brain injuries to pedal cyclists and motorcyclists. The report includes a review of the biomechanics of head injury with particular emphasis on rotational motion and head injury criteria. Injuries resulting from head impacts during motorcycle accidents have been more fully researched than those for pedal cyclists and therefore there is a section on motorcyclist head injuries, based mainly upon the recent COST 327 research, which identifies and quantifies those caused by rotation. Thereafter, a review of data relating specifically to bicycles is included together with a brief section describing the difference between the most common bicycle helmet Standards.

The human head is exposed to loads greatly exceeding the capacity of its natural protection. This explains why, despite the extensive research on head injury during the past 50 years and the continuous improvement of head protection devices, head injury is still by far the most common cause of fatal injury in accidents. The consequences of severe head injuries are often fatal or long lasting and not fully recoverable.

Biomechanic studies frequently conclude that contusions and head injury are synonymous even though observations in patients and in primates have shown that severe and fatal brain injury can occur without visible contusions. Nevertheless, contusions are frequently found in head injured patients and are often referred as "coup" and "contrecoup" which is taken from the French to mean ricochet. Contusions are all caused by contact of the brain with more rigid intracranial (skull) surfaces This single mechanism can cause contusions anywhere inside the skull at an interface between the brain and these rigid surfaces. Therefore, the different contusion classifications bring more confusion than clarification and are better avoided. Contusions are generally believed to be caused by linear acceleration.

Subdural haematoma has been shown to result in a high probability of fatality. The most common type of acute subdural haematoma (ASDH) results from the tearing of veins that bridge the subdural space between the surface of the brain and the various dural sinuses. Gennarelli and Thibault (1982) found that in angular accelerated primates ASDH always overlays the ruptured veins, its size being related to the number of disrupted veins. The subdural haematoma was in these cases always frontally predominant and frequently extended into the hemispheric fissure. It is generally believed that epidural and subdural haematoma head injuries in humans are caused by rotational acceleration; this is supported by work on primates. Damage to the axons affects the motor neuron system that controls all movement and thus, severe damage leads to paralysis and often a fatal injury, particularly when the axonal damage is diffuse. Such damage is believed to be almost entirely the cause of rotational motion and is the reason why many protagonists support the need for helmets to protect against rotational acceleration.

The biomechanics section of this report comments generally on skull fractures and notes that basal skull fractures were common in helmeted and unhelmeted motorcycle riders. Of concern to researchers is the frequency of facial injuries, especially fractures of the facial bones to cyclists. Huelke et al. (1988) have concluded that basilar skull fractures, particularly the hinge type, were produced by facial impact, especially impact to the anterior mandible (jaw bone). Base of the skull fractures frequently lead to serious and fatal brain injury; the jaw bone is not protected by a bicycle helmet. Therefore, caution is needed when assessing brain injuries that may be prevented by bicycle helmets.

For helmets to protect it is important that the values measured in tests correspond to known injury severity. The estimated values for linear acceleration indicate that fatal injuries can occur at 200g and above. This is fairly consistent with Newman's proposed scale (1986) which assigns AIS 4 to a range of peak resultant acceleration of 200g - 250g. This was found to be consistent with HIC (Head Injury Criterion) values of the order of 1000 or more. For rotational acceleration the research shows that concussion, AIS 1-2, can occur at 5,000 rad/s^2 and fatal injury, AIS 5-6, can potentially occur at 10,000 rad/s^2. This correlates with data from the same research that indicates that there is a 35% risk of a brain injury of AIS 3-6 at 10,000 rad/s^2.

Tangential force is measured in motorcycle helmet Standards, Reg 22.05 and BS6658. It is related to rotational acceleration at the centre of a headform but can be a function of helmet geometry. Research has shown that the value of 3500N as used in Reg 22.05 corresponds to AIS 3 and a probability of injury of just less than 50%.

TRL was asked to include the research from COST 327. This study of motorcycle accidents found that the higher the impact speed, the more likely it became that the head injury was critical or fatal. For example, between 61 and 70 km/h, 36% were AIS 6 and between 71 and 80 km/h, 57% were AIS 5-6. It also shows that brain ruptures or vascular separation and subarachnoid bleeding, brain oedema, were caused mostly by an indirect force associated with rapid rotation. Overall, the COST research shows that rotational motion contributed to head injuries of AIS 2 and greater in some 63% of cases and in some 38% of cases rotational motion was judged to be the principal cause. In addition, if motorcycle helmets could be made to absorb 24% more energy then it is postulated that some 20% of the AIS 5-6 casualties would sustain injuries of only AIS 2-4.

Replication of accidents is a very good method for identifying likely protection. McIntosh (1995) reproduced eighteen pedal cyclist accident cases in laboratory tests whereby the accident helmet damage was replicated. In twelve cases there was no head injury and in six cases there was a head injury. The average impact velocity was 20km/h, (12mile/h) for both groups. However, of the cyclists who sustained a serious head injury (>AIS 2) 75% were struck in the temporal region even though only 25% of the impacts were in that region. McIntosh concluded that bicycle helmets offer least protection from an impact in the temporal region.

Overall, there is substantial evidence that bicycle helmets offer protection against brain injury. However, there is ambivalence in the evidence for facial injury. Many reports state that helmets do not protect the middle and lower face but do protect the upper face. One report from Norway states that soft shell helmets increase the risk of facial injury.

An analysis of bicycle accident statistics in New Zealand suggests that in non-motor vehicle crashes, bicycle helmet use may reduce head injuries by between 24% and 32%, depending on age group. A reduction of 20% is reported for motor vehicle crashes. However, further evidence (Curnow, 2003), contests that this reduction is wholly attributable to helmet use as the statistical studies take no account of scientific knowledge of brain injury mechanisms and do not distinguish injuries caused through fracture of the skull and by angular acceleration.

Since most of the studies are based upon conventional accident mechanisms and injury data, it is not possible to identify which, if any, of cyclists' head injuries are caused by rotational motion: this tends to agree with Curnow. Nevertheless, it is known from biomedical research that rotational motion causes brain injury, hence many of the authors believed that it is important and that a test to determine the potential to induce rotational acceleration should be included in bicycle helmet Standards.

A UK study has shown educational schemes to have some success in increasing helmet wearing rates. In Berkshire (UK) an education scheme among 11-15 years olds increased helmet wearing from 11% to 31% after five years. In the campaign area, hospital casualties for bicycle related head injuries in this age group fell from 22% of all bicycle injuries to 12%.

2.2 UK Legislation

Except for some privately organised events, it is not compulsory to wear a crash helmet when riding a bicycle on or off the road in the UK. However, the Department for Transport (DfT) believes it is sensible for cyclists, and especially children, to protect themselves by wearing a bicycle helmet. A review of bicycle helmet effectiveness (Towner et al., 2002) commissioned by the Department concluded that overall there is evidence that bicycle helmets can be effective at reducing the incidence and severity of head, brain and upper facial injuries and that they can be effective in reducing injury for users of all ages, though particularly for children. The report also concludes that there is evidence that compulsory helmet wearing may discourage some people from cycling, leading to decreased bicycle use.

Regular surveys of helmet wearing rates completed by the DfT show that, bicycle helmets were worn by 28% of all cyclists on major roads in built up areas in 2004. This compares to 16% in 1994. The corresponding figures for child cyclists are 14% in 2004 compared to 18% in 1994. Whilst compulsion remains an option that will be reviewed from time to time, the DfT view that at current helmet wearing rates, enforcement would be difficult and, without greater public acceptance, could have an adverse effect on the levels of cycling. The current helmet wearing survey is underway and will report later in 2007.

The DfT believe it would be irresponsible not to promote a product that can reduce injuries and continue to promote helmet wearing on a voluntary basis, especially by children.

3 EXPERIMENTAL STUDY: OBJECTIVES AND HELMET SELECTION

3.1 Objectives and methods

This study was developed to investigate the potential risk of brain injuries which may exist as a result of rotational motion during head impact. A selection of current bicycle helmets has been tested to determine whether they influence the potential for rotational head injuries during an accidental head impact.

Pedal cyclist casualties are a small proportion of road user casualties in Great Britain, forming just over 4% of those killed and just under 7% of those seriously injured (RAGB 2004). However, 25% of those killed and seriously injured are children. An emphasis has therefore been placed on helmets suitable for use by this vulnerable road-user group.

In real life bicycle accidents, the head can be loaded in a number of ways. The impact must be characterised by a number of components which include:

- Head velocity relative to the object struck;
- Shape and stiffness of object struck;
- Friction between the helmet and the object struck;
- Neck forces.

The most obvious benefits of a bicycle helmet are that it:

- Absorbs energy, thereby reducing the severity and hence injury potential of the impact;
- Spreads the impact force thereby reducing the risk of skull fracture(open head injury);
- Greatly reduces the risk of abrasion and other soft tissue injuries.

However, helmet features and an increased effective diameter of the head may increase the potential for rotational acceleration.

To investigate the impact performance of bicycle helmets and identify the likely benefits to wearers in an accident, TRL tested a range of current helmets available in the UK (Table 3.1). The study included tests to measure both linear and rotational acceleration using experimental and standard test methods.

Helmet standards currently provide the main mechanism to ensure that a certain level of protection is provided to the rider. The standards for bicycle helmets are described in the literature review.

To assess linear impact performance, the more recent bicycle helmet standards specify a drop test onto a rigid anvil using an instrumented headform fitted with the test helmet. The test conditions are closely controlled and instrumentation is used to investigate whether the linear impact performance meets controlled criteria / thresholds. Although the results of these tests are normally used to pass or fail a helmet against the set criteria, in this study data from test instrumentation have been used to distinguish performance differences between the helmet makes and types. Since it is not the purpose of this study to evaluate differences in test methods, EN1078 - for which helmets sold in the UK must comply, was chosen for the linear impact test configuration.

Most current bicycle helmet standards include tests for assessment of the linear impact performance of the helmets. The literature review has identified that rotation of the head is also a significant cause of brain injury, yet none of these current standards assess helmet performance in terms of rotational injury potential. Neither do the standards stipulate restrictions to helmet design which may eliminate the most hazardous features e.g. by limiting projections to those of a considered safe geometry.

The standard for motorcycle helmets, Reg 22.05, does include a test which indirectly assesses and prescribes a performance requirement for rotational acceleration. However, it prescribes two methods for assessing helmet performance; Method A and Method B. Method A incorporates an oblique impact using a free motion headform whereas Method B uses a rigid headform and sliding trolley impact. It is the authors' opinion that Method A is the most realistic of the two methods as it generates a combination of linear and rotational loading as would be expected in accidents. This method also takes account of the effect of the helmet geometry and liner stiffness, both of which may affect the potential for rotational brain injury. Therefore, for the TRL study, the Reg 22.05 oblique impact test method (method A) was chosen and adapted for use with bicycle helmets.

The method comprises abrasion tests, whereby the performance is mainly dependent upon friction and the liner stiffness, and projection strength tests whereby the performance is dependent upon the geometry and force required to sheer or bend the projection. The performance criteria of Reg 22.05 were selected to align approximately with the onset of head injury.

Reg 22.05 (Method A) currently prescribes two oblique impact configurations, one to assess abrasion resistance and the other projection strength. In both tests a headform fitted with a helmet is dropped onto an anvil mounted at 15º at 8.5m/s onto either, 1) an abrasive anvil which generates high frictional forces or, 2) a bar anvil with 6mm high raised edges which engage with helmet projections. The anvil load measured must not exceed 3.5kN with a 25Ns impulse during abrasive tests. A 2.5kN anvil load with a 12.5Ns impulse must not be exceeded during the projection strength test. Figure A.1, Appendix A, shows the apparatus and Figure A.5, a close up of the projection anvil.

For further understanding it is useful to compare the test method with an equivalent accident situation. Figure 3.1 shows the accident conditions equivalent to the Reg 22.05 oblique test resolved into vertical and horizontal components. The impact equates to a fall of 0.25m when travelling at approximately 30km/h. Although these accident conditions may be an improbable situation for a junior rider, they represent worse case circumstances (they are based on a motorcycle helmet standard) and are the most appropriate conditions to evaluate the helmet's abrasion and projection strength performance. At less acute angles, which might be more likely for higher falls or slower riding speeds, the helmet performance would be greatly influenced by the linear impact performance.

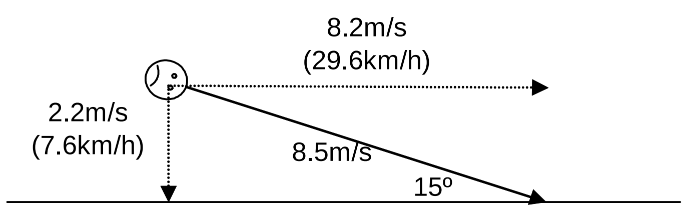

Figure 3.1 Accident situation equivalent to the Regulation 22.05 Method A test

As for the linear impact performance, the number of potential tests were limited but needed to evaluate the wide range of helmet features. For this reason, it was not possible to assess the same features for each test helmet and a direct comparison between helmets was not possible. Instead, a wide range of features representative of those found on current bicycle helmets were evaluated to establish whether any helmet features or designs have a detrimental influence on rotational forces imparted to the head.

For each of 8 helmet models approximately 4 helmet features were evaluated using Reg 22.05 Method A. Generally, sites were chosen which were likely to generate the maximum rotational loading and were judged to be worst case, for example, impacting the extended 'tail' projections.

Approximately half the impacts were onto the abrasive anvil and half onto the bar anvil. In both configurations, the loads generated on the anvil were measured by means of a tri-axial load cell so that the normal and tangential loads measured at the impact surface could be compared directly with Reg 22.05 criteria. Nevertheless, the exact rotational acceleration generated within the headform about its centre of gravity was also important since the relationship between anvil forces and rotational acceleration is unknown for bicycle helmets. Instrumented headforms, fitted with nine-accelerometer arrays, were therefore used for this purpose.

3.2 Helmet selection

3.2.1 Make and model

In order to complete an experimental study which may objectively assess current bicycle helmets, it was necessary to select test helmets which were representative of those currently available on the market and which best illustrated the range of protection offered.

To ensure a representative selection of helmets was made; details of a wide range of helmets available in the UK were obtained through internet searches. The helmet designs and features were then evaluated to establish the suitability of the helmets. The principal considerations were;

- Is the helmet design or the material likely to increase, decrease or have little or no effect on the risk of rotational injury during oblique impacts?
- Are the helmet features or projections likely to increase, decrease or have little or no effect on the risk of rotational injury during projection impacts?
- Does the helmet have any unique features which may be detrimental to linear impact performance?
- Does the helmet include chinguard protection?
- What materials are used in the helmet construction?
- Is the helmet designed and marketed for use by a child adult or both?
- What are the available sizes and the size range for any one helmet?

Helmets were chosen so as to include those which may influence the risk for rotational injury and also to ensure that as many different designs and features could be evaluated during the course of the study. For example, helmets with large, rigid projections were included because they were judged to be most likely to induce rotational motion than helmets with a smooth finish; both types were included for comparison.

In total, eight helmets were chosen for this study as detailed in Table 3.1. Based on the size range, design and helmet graphics available, four helmets were considered to be targeted towards the youngest children and four were suitable for older ‘youth’ or small adult riders.

The sizes of ‘child specific helmets’ ranged from 46cm to 57cm whereas the ‘youth / adult‘ helmets ranged between 51cm and 62cm. Two of the ‘youth / adult’ style helmets were designed for use during specific cycling sports (BMX and downhill mountain biking) but could potentially also be worn when cycling on the road.

The design features incorporated into these helmets included both built-in and detachable peaks, plastic ‘micro-shell’ and rigid polycarbonate shell constructions, exposed polystyrene liner materials, chinguard, and ventilation slots. These features were not specific to these helmets within the range of models reviewed but those helmets chosen allowed at least one of each of these features to be evaluated. All of the chosen helmets conformed to EN1078 which is the appropriate test standard to which helmets must conform for sale in the UK and EC.

It should be noted that helmet sales figures were not used during the helmet selection process for two reasons: i) sales figures do not necessarily represent the exposure rate, i.e. how much each helmet is worn and its likelihood to be involved in accidents, and ii) sales figures are not indicative of the likely protection offered by the helmet. Other factors such as style and cost may greatly influence consumer decision making. Despite this, the helmets selected for this study were, by coincidence, produced by four major UK helmet suppliers; BELL, Specialised, MET and Giro and are therefore likely to have good market penetration and be representative of current and at least short-term future helmet stock.

Table 3.1 Test helmets

Make/Model	Style	Standards	Nomimal mass [g]	Size range [cm]	Photo
Met Buddy	Child	EN1078	231	46-54	
Bell Bellino	Child	EN1078	235	52-56	
Giro Rodeo	Child	EN1078	262	50-55	
Met Super Loopy	Child/Youth	EN1078	243	52-57	

Make/Model	Style	Standards	Nomimal mass [g]	Size range [cm]	Photo
Met Stradivarius	Youth/Adult Road Racing	EN1078	273	54-57	
Met Parachute	Youth/Adult Downhill MTB	EN1078	554	54-57	
Bell Faction	Youth BMX	EN1078	427	51-56	
Specialised Airforce	Youth/Adult Recreational	EN1078	301	54-62	

3.2.2 Size selection

The performance of a bicycle helmet can vary with the head mass. Although in some circumstances the head loading may be favourable, often the injury outcome can be more severe where the head mass is not aligned with that for which the helmet design has been optimised. For example, if mass increases, the energy that must be absorbed by the helmet also increases. Thus, with a heavier head there is an increased probability that a helmet will fail to absorb all the impact energy and 'bottom out'. At this point the helmet effectively becomes very stiff and causes excessive loading to the head capable of greater injury severity. For heads lighter than those to which the helmet is optimised, more severe injury outcomes may be expected as the liner stiffness is increased to provide higher levels of energy absorption. Here the liner may be too stiff and deform inadequately causing the head to stop more quickly than is ideal, and consequently generating excessive head loading.

Although differing headform size can counteract mass influences, the significance of headform mass between sizes is an important factor to investigate when assessing the potential for injury between helmets. This is especially true for helmets which will accommodate a large range of sizes, since there is likely to be a greater variation in wearers' the head mass.

Helmets for which the specified size lies close to the next headform sizing may also be sensitive to head mass. This situation was observed for two of the helmets selected with the size range extending to within 1cm of the next size of test headform. For this reason it was necessary to include a range of headform sizes of differing mass for this study.

Due to the over-representation of children in cycling accident statistics, smaller child and youth helmets have been targeted in this study. The selected helmet sizes (Table 3.2) were therefore intended to be suitable for use by a child or youth with a 54cm head circumference (size E headform). This allowed the performance of the helmet to be evaluated for both user groups by selecting a slightly smaller 50cm (A) headform for child style helmets and a slightly larger 57cm (J) headform for adult style helmets.

The headforms needed to test helmets to EN1078 (the helmets standard for pedal cyclists) conform to geometry and mass constraints defined by EN960. Although there are eight sizes in total ranging from reference A (50cm circumference) to O (62cm circumference) - each with a unique mass, only sizes 50cm (A), 54cm (E), 57cm (J), 60cm (M) and 62cm (O) are used for shock absorption testing. Since the helmets range from 50cm to 57cm only A, E and J size headforms were required.

For the Bell Faction, Bell Bellino and Met Super Loopy helmets it was thought that the helmet sizing had been purposely specified to be less than the range of head sizes that they would fit. This would allow more stringent impact tests using smaller and larger headforms to be avoided. This was of concern because the selection of helmets by the end-user may be influenced by comfort, style, availability and perhaps even the ability to be grown into rather than the specified size range alone. This could potentially result in helmets being worn on heads with a circumference, and possibly mass, outside the manufacturer's optimal design range whilst still being considered a good comfortable fit.

In the case of the Bell Faction, the size is defined as 51-56cm, which is 1cm larger than a 50cm (A) headform and 1cm smaller than a 57cm (J) headform. However, the helmet was found to fit 50cm (A) and 57cm (J) headforms without difficulty and could potentially be worn by cyclist with this head geometry. It is feasible that this size range was specifically chosen to restrict testing to a 54cm headform alone. TRL therefore tested this helmet with a 54cm (E) and slightly oversized 57cm (J) headform, the latter of which has greater mass and may be considered a more severe test due to the increased impact energy.

As for the Bell Faction, the Bell Bellino and MET Super Loopy helmets readily fitted 50cm (A) and 57cm (J) headforms yet the designated size ranges are 51cm to 55cm. The helmet designs and graphics clearly suggest that the helmets are intended for use by children and it is therefore feasible that such helmets may be bought slightly undersize (below the manufacturers rating) by parents wishing their children to grow into it, whilst still being an apparently safe fit. These helmets were therefore tested with 50cm (A) and 54cm (E) headforms.

Table 3.2 Helmet sizes and test headforms

Sample	Typical user	Helmet size [cm]	EN1078 headform sizes	TRL test headform sizes
Met Buddy	Child	46-54	A, E	A
				E
Bell Bellino	Child	52-56	E	A*
				E
Giro Rodeo	Child	50-55	A, E	A
				E
Met Super Loopy	Child/Youth	52-57	E, J	A*
				E
Met Stradivarius	Youth/Adult	54-57	E, J	E
				J
Met Parachute	Youth/Adult	54-57	E, J	E
				J
Bell Faction	Youth/Adult	51-56	E	E
				J*
Specialised Airforce	Youth/Adult	54-62	E, J, M, O	E
				J

*Intentionally outside the designated size range

4 EXPERIMENTAL STUDY: HEADFORMS, TEST METHODS AND RESULTS

4.1 Headforms and instrumentation

Linear impact

The chosen linear impact performance test specification was based on the most severe test configurations within EN1078. The tests for shock absorption are made using rigid metal headforms which conform to BS EN960 - the geometry and mass of which varies depending on their size.

The headforms are available in a range of sizes each with a unique geometry and mass as shown in Table 4.1.

For EN1078 shock absorption tests, only headform sizes 50cm (A), 54cm (E), 57cm (J), 60cm (M) and 62cm (O) apply and helmets should be tested with all headform sizes that fit within the manufacturers' claimed head size range. For example, a helmet specified to fit 52cm – 59cm heads would be tested using 54cm (E) and 57cm (J) headforms.

Table 4.1 Definition of EN960 headform size and mass

EN 960 designated size (Shock absorption tests only)	Internal circumference of helmet [cm]	Headform mass [kg]
A	500	3.1 +/- 0.10kg
C	520	n/a†
E	540	4.2 +/- 0.12kg
G	560	n/a†
J	570	4.8+/- 0.14kg
K	580	n/a†
M	600	5.6 +/- 0.16kg
O	620	6.1 +/- 0.18kg

† mass is only specified for shock absorption tests

This is significant as EN1078 prescribes a particular impact velocity rather than fixed impact energy. Consequently, tests with larger, heavier heads require helmets with a greater energy absorption capacity than helmets tested with smaller headforms of reduced mass. It was therefore important that this study, which was primarily focused on child and youth helmets, used a range of headform sizes including 50cm (A), 54cm (E) and 57cm (J).

During linear impact tests the resulting acceleration on the headform is of the greatest importance. This parameter indicates the rate at which the initial impact speed of the headform is reduced, usually to zero, and is measured in ‘g’ (the acceleration due to gravity equal to $9.81 m/s^2$) using accelerometers. The acceleration may be used to determine HIC (Head Injury Criterion) which is a complex measure of injury potential that includes acceleration and duration. Peak acceleration and HIC can be related to the potential for injury.

Three accelerometers, with mutually orthogonal axes, were mounted at the centre of gravity of each test headform to determine the resultant linear acceleration regardless of headform orientation. The headform was dropped in freefall and data was captured in accordance with SAE J211 using a 100kHz sampling frequency and CFC1000 filter during post processing.

Oblique impact

The oblique impact test is based on the motorcycle helmet standard, Reg 22 05, which prescribes testing with a headform conforming to (BS6489/EN960). Again the geometry and mass are variable with size. However, only one headform size, 57cm (J), is currently required for these oblique tests. Due to the particular interest in children’s helmet performance it was decided that, as for the linear impact tests, the use of smaller headforms was important to this study. Sizes 50cm (A), 54cm (E) and 57cm (J) were utilised.

The headform sizes specified for linear and oblique testing equate to a helmet range between 50cm and 57cm. Head size is of course variable from person to person and is not regimented by age alone but this range is generally representative of a typical child of about 3 years old to a medium sized adult. This size range allowed the study to assess helmet performance for typical children but also make some judgement on the likely factors for adults which may be important.

The oblique impact test method is based on Reg 22.05, Method A for abrasion testing. This method requires the measurement of anvil forces tangential and normal to an oblique impact surface by way of a tri-axial load cell rigidly fixed at 15° to the impact direction.

To determine the rotational acceleration imparted to the headform, TRL fitted a nine-accelerometer array. The nine accelerometer configuration is shown in Figure A3 Appendix A. The accelerometers are configured to allow the measurement of acceleration at 9 unique points, 3 points on each of three orthogonal axes. Bespoke software, designed at TRL was used to compare the acceleration differentials at these points and thereby determine rotational motion of the array in three axes and the resultant rotational acceleration about the centre of gravity.

TRL has specified that three headform sizes should be used in this study; 50cm (A), 54cm (E) and 57cm (J) but only a 57cm (J) headform was available with a nine accelerometer array. The collection of data for smaller (child) headform sizes was considered important, so a 50cm (E) headform was modified to include a nine accelerometer array. The geometry of this headform restricted the space available to accommodate this array and a cantilevered array was therefore deployed. This ensured that the length of the array arms, important for improved accuracy, were maximised. The cantilevered arrangement is shown in figure A4, Appendix A.

Data was captured in accordance with SAE J211 with a 100kHz sampling frequency. A CFC1000 filter was used during post processing. The 54cm (E) headform required the use of a CFC180 filter for post processing. This filter was used to reduce the high frequency resonance that was observed on the test data, particularly for projection tests where the overall signal was small. The noise was a consequence of the small headform size and array configuration which promoted resonance of the array block.

A bench test of the 54cm (E) headform fitted to a rigid base using a hammer strike to excite the system revealed a resonance in the region of 800Hz. The use of CFC180 filter was appropriate for removal of the superfluous resonance data during the post processing phase. This filter is a low pass filter with a 3dB limit frequency of 300Hz. To validate the suitability of this filter, data obtained using the headform which did not show evidence of resonance at this frequency was filtered using both CFC1000 (3dB limit frequency of 1650Hz) and CFC180 filters. The peak linear and rotational accelerations obtained using these two filters were almost identical in magnitude and it was therefore concluded that the CFC180 filter was appropriate for this application.

4.2 Test methods

4.2.1 Linear impact tests

Figure A1, Appendix A, shows the apparatus for drop testing a helmeted headform. Most standards require the use of this type of free-motion headform apparatus or one based on a guided headform configuration. A test velocity or impact energy, headform size, orientation and anvil type are usually specified for each test and a limit is placed on the peak resultant acceleration measured during the impact.

Another typical requirement is headform mass with some standards requiring a fixed mass for all headform sizes whereas others specify variable mass with size. For example, the US ASTM standard F1447 specifies an impact velocity of 6.2m/s, (onto a flat surface) and a headform mass of 5kg for all sizes: the headform acceleration must not exceed 300g. The harmonised European (CEN) Standard EN 1078 specifies an impact velocity of 5.42m/s (onto a flat surface) and a different mass for each headform size; the resultant headform acceleration must not exceed 250g.

Since, it was not the purpose of this study to evaluate different International Standards, the test method chosen for this study was that specified by EN 1078, to which most helmets sold in the UK conform. This standard uses free-motion, variable mass, headform apparatus. Table 4.2 summarises the test configuration and requirements.

Table 4.2. Summary of bicycle helmet standards

	STANDARD						
	ASTM F1447-98	**BS EN 1078:1997**	**Snell B-95**	**CPSC 16 CFR Pt 1203**	**SABS 1542:1991**	**AS/NZS 2063:1996**	**JIS T 8134:1995**
Headform mass (standard size)	5±0.1kg including assembly	4.7±0.14kg	5-6.5kg including assembly	5±0.1kg including assembly	Headform striker mass 5kg	5±0.1kg including assembly	Standard 5kg
Anvil	Flat, hemi and kerbstone	Flat and kerbstone	Flat, hemi and kerbstone	Flat, hemi and kerbstone	Flat and Hemispherical striker	Flat only	Flat only
Impact velocity	Flat 6.2m/s Hemi/kerb 4.8m/s	Flat 5.42m/s Kerb 4.57m/s	*Flat 5.82-6.63m/s Hemi/kerb 4.71-5.37m/s*	Flat 6.2m/s Hemi/kerb 4.8m/s	Flat 4.57m/s Hemispherical 4.57m/s	Flat 5.42m/s	*Flat 4.43m/s*
Impact energy	*Flat 98J Kerb 59J*	*Flat 69J Kerb 49J*	*Flat 110J Hemi/Kerb 72J*	*Flat 98J Hemi/kerb 59J*	*Flat 52J Kerb 52J*	*Flat 73.6J*	*Flat 49J*
Maximum acceleration	300g	250g	300g	300g	300g	300g or 200g for 3ms, or 150g for 6ms	299.8g, or 149.9g for 4ms
Retention system strength	4kg mass, drop height 0.6m, 23.5J	4kg mass, drop height 0.6m, 23.5J	4kg mass, drop height 0.6m, 23.5J	4kg mass, drop height 0.6m, 23.5J	10kg mass, drop height 0.3m, 29.4J	None	38kg mass, drop height 0.02m, 7.5J
Positional stability	4kg mass, drop height 0.6m, 23.5J	10kg mass, drop height 0.175m,17.2J	4kg mass, drop height 0.6m, 23.5J	4kg mass, drop height 0.6m, 23.5J	None	None	None

Values in italics are calculated using values specified in the Standard and assume a standard sized headform is used

To allow a comprehensive yet cost effective evaluation of the protection offered by the helmets TRL proposed two linear impact configurations per helmet model. The impact configurations selected from EN1078 were considered to be the worst case and most likely to exceed the performance limitation of the helmets in ambient conditions.

The tests selected for helmet evaluation were a kerbstone anvil test at 4.57m/s (1m drop) and a flat anvil test at 5.42m/s (1.5m drop). Tests on the two anvil types represented extreme impact conditions where a helmet may generate severe head loading. Typically, during a flat anvil impact the liner may be too stiff and stop the headform rapidly whereas during a kerbstone anvil test the liner may be too soft and the liner will bottom out (crush to the maximum extent).

Test sites were chosen above the test line as prescribed by EN1078. Flat anvil impacts were made to the side, temporal region, of the headform whereas front sites were selected for the kerb anvil to give the worst case results. Impacting the helmet in areas with the least protective liner material or where ventilation holes offered low headform coverage (see figures E1 onwards, Appendix E) will increase the probability of exceeding the helmet's protective capability by causing the material to bottom out.

The impact tests were performed using two different sizes of test headform for each helmet to investigate the effect of headform mass. Although EN1078 currently requires that helmets need only be tested with headform sizes within the manufacturer's specified helmet size range (up to 5 in total), there is potential for manufacturers to restrict the size range below that which a helmet would actually fit thereby avoiding regulation tests with the extreme headform sizes. Since the specified range may not represent real-world use, two helmets were tested with headforms that were outside the prescribed size range to investigate this. The helmets were still judged to have an acceptable fit on the test headform (Table 3.2).

For all tests, comfort padding provided with the helmet remained fitted and the retention system was fastened to the tightest adjustment available, in accordance with the manufacturer's instructions. New undamaged helmets were used for each impact test series as detailed in Table 3.1.

The results from this test programme were not anticipated to identify helmets that failed to comply with the Standard to which it had been approved but to observe whether any particular helmet features or designs perform less well than others and have a detrimental effect on safety. Table 4.3 gives the test matrix for the linear impact study by helmet type and Table 4.4 gives the matrix by helmet make and model.

Table 4.3. Linear impact performance test matrix by helmet type

<table>
<tr><th colspan="3">Helmet</th><th rowspan="2">Impact site</th><th rowspan="2">Test Headform</th><th rowspan="2">Impact Anvil</th><th rowspan="2">Required Velocity [m/s]</th></tr>
<tr><th>Model</th><th>Type</th><th>Part</th></tr>
<tr><td rowspan="4">1,2,3 & 4</td><td rowspan="4">Child</td><td rowspan="2">1</td><td>Front</td><td rowspan="2">A</td><td>Kerb</td><td>4.57</td></tr>
<tr><td>Side-L</td><td>Flat</td><td>5.42</td></tr>
<tr><td rowspan="2">2</td><td>Front</td><td rowspan="2">E</td><td>Kerb</td><td>4.57</td></tr>
<tr><td>Side-L</td><td>Flat</td><td>5.42</td></tr>
<tr><td rowspan="4">5,6,7 & 8</td><td rowspan="4">Youth/ adult</td><td rowspan="2">1</td><td>Front</td><td rowspan="2">E</td><td>Kerb</td><td>4.57</td></tr>
<tr><td>Side-L</td><td>Flat</td><td>5.42</td></tr>
<tr><td rowspan="2">2</td><td>Front</td><td rowspan="2">J</td><td>Kerb</td><td>4.57</td></tr>
<tr><td>Side-L</td><td>Flat</td><td>5.42</td></tr>
</table>

Table 4.4 Linear impact performance test matrix by helmet make and model.

Helmet	Headform (EN960)	Anvil	Impact speed [m/s]	Site Evaluated
1 Met Buddy	E	abrasive	8.5	side - right - face down
				crown - face down
			6.0	crown - face down
		projection	8.5	visor - crown down
				longitudinal ridges - crown down
	A	projection	8.5	visor - crown down
	A	abrasive		crown - face down
2 Bell Bellino	E	abrasive	8.5	side - right - face down
				rear - right side down
		projection		visor - crown down
				longitudinal ridges - crown down
				visor - crown down
	A	projection	8.5	longitudinal ridges - crown down
3 Giro Rodeo	E	abrasive	8.5	side - right - face down
				crown - face down
		projection	8.5	visor - crown down
				longitudinal ridges - crown down
				visor - crown down
	A	abrasive	8.5	side - right - face down
4 Met Super Loopy	E	abrasive	8.5	side - right - face down
				crown - face down
			6.0	side - right - face down
		projection	8.5	visor - crown down
				longitudinal ridges - crown down
5 Met Stradivarius	E	abrasive	8.5	side - right - face down
				crown - face down
			6.0	side - left - face down
		projection	8.5	tail - crown first
				front right chinstrap anchorage tabs
				crown - side down - to catch ridges
				tail (2 contact points) - crown first
6 Met Parachute	E	abrasive	8.5	side - right - face down
				crown - face down
		projection	8.5	side - right - chin anchorage bolt
				chinguard - right side down
7 Bell Faction	E	abrasive	8.5	side - right
				rear - side right down
		projection	8.5	nose down - aiming to catch central vents
	J	projection	8.5	nose down - aiming to catch central vents
		abrasive	8.5	side - right
8 Specialised Airforce	E	abrasive	8.5	front right - on microshell - face down
				rear - on EPS
			6.0	rear - on EPS
		Projection	8.5	crown - side down - to catch ridges
				tail - crown first
	J	Projection	8.5	crown - side down - to catch ridges
		Abrasive		rear - on EPS

4.2.2 Oblique impact tests

Figure 4.1 illustrates the force diagram for an oblique impact to assist in the understanding of the method and results of the abrasive and projection tests.

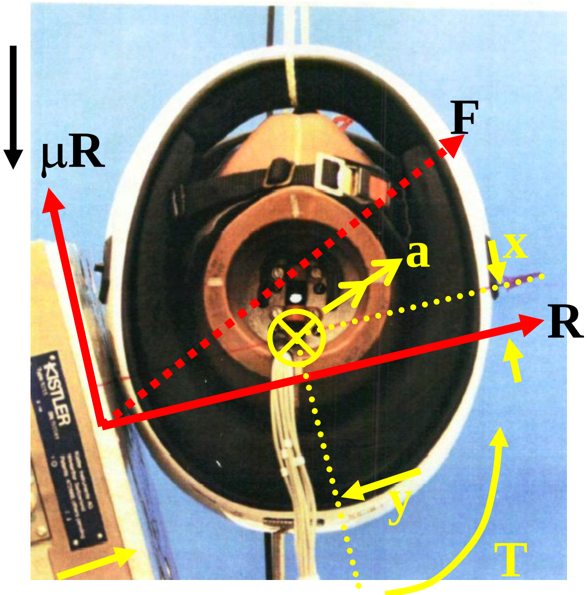

Figure 4.1 Force diagram for an oblique impact.

The diagram illustrates that, during an idealised oblique impact, force components are generated normal and tangential to the impact surface due to the interaction of the helmet shell with the anvil surface. These forces are dependant on many factors including the helmet stiffness and, in the case of the tangential force, helmet friction or projection strength. Anvil forces transmitted through the helmet to the headform cause it to accelerate. In idealised conditions, at the point of maximum acceleration the load transmitted to the headform would theoretically be equal to that exerted on the anvil surface.

It can be seen that the anvil force components need not act directly through the centre of gravity of the headform and consequently a torque may also be generated on the helmet and head, causing them to rotate. The level of rotational acceleration of the head is theoretically a function of the torque exerted on the head and the head’s moment of inertia. However, the impact event is dynamic and there may be many factors which influence the head rotational response. For example, the helmet geometry will influence the offset between the anvil forces and the centre of gravity, This in turn may influence the impulse to the head and the ensuing rotation which may consequently peak at a different time to the maximum anvil force. Such variability makes it difficult to predict or quantify the likely outcome of varying any test parameter by measurement of anvil forces alone. This is primarily why instrumented headform tests are preferred for the analysis of oblique impacts as the true rotational acceleration in the head can be measured.

For this study, instrumentation has therefore been included to measure headform peak linear acceleration (g), peak rotational acceleration (rad/s^2) and anvil peak normal and tangential force (N). Further measures included the maximum rotational acceleration of the headform in each of the three orthogonal axes about the headform as defined by SAE J211 and a calculation of HIC.

The rotational velocity (rad/s) and the anvil tangential impulse were calculated by integration of the rotational acceleration and anvil force with respect to time for the initial 15ms impact duration. The coefficient of friction is the ratio between the tangential and normal anvil forces and was calculated using instantaneous force levels where normal component exceeded 70% of the peak normal force recorded over the entire impact event. This method was more appropriate for this type of dynamic event since the peak normal and peak tangential force are not otherwise aligned.

4.3 Test results and discussion

4.3.1 Linear impact test results

The results of the linear impact tests are given in Tables 4.5 and 4.6 with the graphical test data in Appendix B. The results have been divided into two groups: the child / youth and youth / adult style helmets.

The required impact speed was 4.57m/s for the kerbstone tests and 5.42m/s for the flat anvil tests with the impact energy ranging between 34.5J and 72.6J depending on the headform and impact configuration. The headform energy is calculated by $E = 1/2mv^2$ where m is the mass of the headform, v is the impact speed and E is the headform energy. The mass of the headforms was 3.2kg, 4.2kg and 4.9kg for sizes 50cm (A), 54cm (E) and 57cm (J) respectively.

Helmet performance was assessed with two parameters: peak linear acceleration (in ‘g’) and HIC. These parameters are suitable for making direct comparisons between helmets and for relating helmet performance to injury threshold and injury probability.

Research has shown that both the magnitude (peak value) and exposure (time) of head acceleration may affect injury outcome. The AS/NSZ 2063 and JIS T8134 bicycle helmet standards, discussed in the literature review, prescribe acceleration-time exceedence criteria which are similar for both Standards. These criteria have a limited application to events where a particular acceleration value is exceeded. HIC however, is a more sophisticated parameter and a function of both acceleration time history and the peak acceleration value. This parameter can therefore be used to assess the impact severity of any head impact event and is widely used in the automotive industry where HIC 1000 is typically defined as the value that must not be exceeded.

Most current bicycle helmet standards prescribe only a value of peak acceleration, typically 250g or 300g, which must not be exceeded. The EN1078 Standard which was used for this test work states that 250g must not be exceeded. AS/NSZ and JIS standards are the only bicycle helmet standards to prescribe maximum acceleration exceedence values. AS/NSZ requires that 200g is not exceeded for more than 3ms and 150g for not more than 6ms whereas JIS requires that 149.9g must not be exceeded for more than for 4ms. None of the bicycle helmet standards prescribe values for HIC and this may indicate a lack of support for this parameter in the industry.

Table 4.5 Results for the Child / Youth helmets

Helmet			Impact Site	Headform	Impact Anvil	Test ref	Impact speed [m/s]	Headform Energy [J]	**Peak linear acceleration [g]**	HIC	Comments
1	Met Buddy	a	Front	A	Kerb	h28fy	4.57	34.7	171	890	front right vent
		a	Side-L	A	Flat	a28fy	5.42	48.6	241	1556	side left, >200g 1.6ms
		b	Front	E	Kerb	j02cy	4.57	45.2	151	682	
		b	Side-L	E	Flat	a02cy	5.42	64.0	211	1117	>200g 0.7ms
2	Bell Bellino	a	Front	A	Kerb	g28fy	4.57	34.7	167	909	front central vent
		a	Side-L	A	Flat	b28fy	5.42	48.4	200	1207	side left
		b	Front	E	Kerb	k02cy	4.57	45.7	149	746	
		b	Side-L	E	Flat	b02cy	5.42	63.5	180	888	
3	Giro Rodeo	a	Front	A	Kerb	f28fy	4.57	35.0	178	1004	front central vent
		a	Side-L	A	Flat	c28fy	5.42	48.5	203	1179	side left, >200g 0.7ms
		b	Front	E	Kerb	l02cy	4.57	44.6	154	734	
		b	Side-L	E	Flat	c02cy	5.42	63.4	208	1138	>200g 0.5ms
4	Met Super Loopy	a	Front	A	Kerb	e28fy	4.57	34.9	133	675	front central vent
		a	Side-L	A	Flat	d28fy	5.42	48.3	153	721	side left
		b	Front	E	Kerb	m02cy	4.57	45.5	120	531	
		b	Side-L	E	Flat	e02cy	5.42	62.6	164	743	

Table 4.6 Results for the Youth / adult helmets

Helmet			Impact Site	Headform	Impact Anvil	Test ref	Impact speed [m/s]	Head form Energy [J]	**Peak linear acceleration [g]**	HIC	Comments
5	Met Stradivarius	a	Front	E	Kerb	n02cy	4.57	45.1	113	437	
		a	Side-L	E	Flat	f02cy	5.42	63.7	165	759	
		b	Front	J	Kerb	f01cy	4.57	50.9	131	577	
		b	Side-L	J	Flat	b01cy	5.42	71.4	158	596	
6	Met Parachute	a	Front	E	Kerb	o02cy	4.57	45.4	126	559	
		a	Side-L	E	Flat	g02cy	5.42	63.4	192	830	
		b	Front	J	Kerb	g01cy	4.57	51.3	112	441	
		b	Side-L	J	Flat	c01cy	5.42	71.6	169	667	
7	Bell Faction	a	Front	E	Kerb	p02cy	4.57	45.3	137	658	
		a	Side-L	E	Flat	h02cy	5.42	63.8	241	1618	>200g 1.6ms
		b	Front	J	Kerb	h01cy	4.57	51.9	132	604	
		b	Side-L	J	Flat	d01cy	5.42	72.3	**285**	1481	failed EN1078, >200g 1.4ms
8	Specialised Airforce	a	Front	E	Kerb	q02cy	4.57	45.1	140	607	
		a	Side-L	E	Flat	i02cy	5.42	63.6	178	859	
		b	Front	J	Kerb	e01cy	4.57	52.1	123	521	
		b	Side-L	J	Flat	a01cy	5.42	71.9	163	622	

4.3.2 Linear impact test results

Figure 4.2 shows the peak linear acceleration measured for all of the helmets during the linear impacts onto the kerbstone anvil at 4.57m/s and the flat anvil at 5.42m/s. Table 4.7 gives the probability of injury for various values of peak linear acceleration (ref).

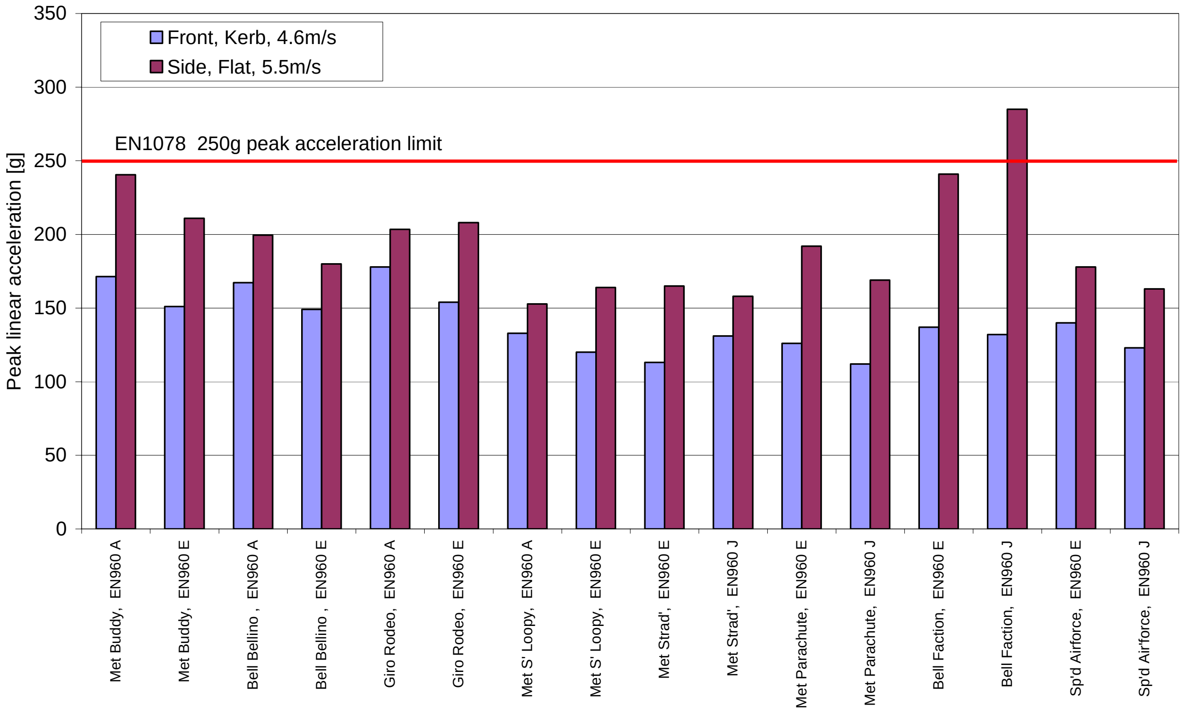

Figure 4.2 Linear acceleration performance

These results are encouraging because only one helmet exceeded the EN1078 value of 250g and only three helmets exceeded 200g for both impact types. Table 4.7 shows that at 250g a severe injury is likely and the probability of fatality may be as high as 10.6%. Below 200g, an injury of AIS 3 or less is likely and the risk of fatality is below 2.1%.

Five out of eight helmets achieved 200g or less in this study. Potentially, a reduction of 50g of the peak acceleration permitted, from 250g to 200g would result in some of the poorest performing helmets failing the requirement. If such a requirement were applied to EN1078 then an 8% reduction in the risk of fatal injury may be achieved (Table 4.7)

What was also encouraging was that, despite anvils being aligned to provide worst case results, for example aiming the helmet so that it struck the kerbstone between air vents, the helmets remained in one piece during these impacts and the damage from each impact was localised. In a bicycle accident the helmet could therefore continue to provide protection during subsequent impacts.

In the single case where the peak acceleration exceeded 250g the helmet, the Bell Faction was tested using a 57cm (J) headform. This headform exceeds the manufacturer's designated size range of 51cm – 56cm by 1cm. The 285g result is a consequence of the helmet's inability to absorb some, or any, of the 8.4J of additional impact energy associated with the larger headform size (72.3J for size 57cm (J) compared with 63.8J for size 54cm (E)). Consequently the probability of a fatal injury is approximately 55% and a critical injury would be highly likely.

Table 4.7 Injury outcome for a range of peak linear acceleration.

Peak linear acceleration	AIS head injury severity	Injury interpretation	Approximate probability of fatality (ref)
< 50g	0	No injury	0.0%
50 - 100g	1	'Minor' injury	0.0%
100 - 150g	2	'Moderate' injury	0.1 - 0.4%
150 - 200g	3	'Serious' injury	0.8 - 2.1%
200 - 250g	4	'Severe' injury	7.9 - 10.6%
250 - 300g	5	'Critical'	53.1 – 58.4%
> 300g	6	'Unsurvivable' (Maximum)	>58.4%

Although this result may be expected because the helmet was tested outside the test specification, this shows how a helmet may be optimised to the Standard and how a defined size range may restrict the application of the most severe impact tests. In this example, the impact performance was some 116g higher than the next worse helmet performance in similar impact conditions. Table 4.7 shows that such differences may equate to the difference between a critical and moderate severity head injury.

It is important to realise that such impacts may be quite feasible in accidents, because helmets such as the Bell Faction may comfortably fit a head outside the size (and mass) range designated by the manufacturer. Helmet size is especially important to riders under the age of 16 years where growth rate is high and an increased tendency to wear helmets outside the manufacturer's specified range may be expected, particularly because many people are unaware of their head size and may simply rely upon comfort when purchased. It is also possible that a helmet may be bought with the idea that it can be 'grown into' or that the helmet should be used until it will not fit to gain extra value. In either instance it appears that it would be difficult to prevent this 'misuse'.

It can be seen that, for EN1078 impact conditions and helmets such as the Bell Faction, the consequence of fit relative to stated size may be very significant. Although these Standard impact conditions may not be fully representative of the real life conditions, additional tests with all headform sizes which can comfortably be achieved within the test helmet may eliminate some doubt over the consequence of size to impact performance.

Despite this exceptional result for the Bell Faction, it is significant that when testing within the manufacturer's designated size range (with 54cm (E) headform), this helmet was still a poor performer with a peak linear acceleration of 241g, only 9g (3.6%) below the EN1078 requirement of 250g. Closer inspection of the test data (see Appendix A) reveals that this helmet is initially much stiffer than other helmet designs when tested onto the flat anvil and this may be a consequence of the stiff outer shell. If the combined results for both kerb and flat anvils are considered (using headforms within the manufacturer's size range) the Met Buddy has a less favourable result, averaging 193g compared to the Bell Faction of 189g. This is only 24g above the average of all the helmets tested (165g) which utilise thin plastic outer shell constructions. Poor helmet performance cannot therefore be attributed to helmets with a stiff outer shell.

Further to the above observations, the correlation between linear acceleration and HIC has also been investigated. Figure 4.3 shows a good correlation between these parameters for all the helmet linear impacts with a correlation coefficient of 0.94 (r^2=0.88). This correlation is possibly what may be expected given that they all satisfy the requirements of EN1078.

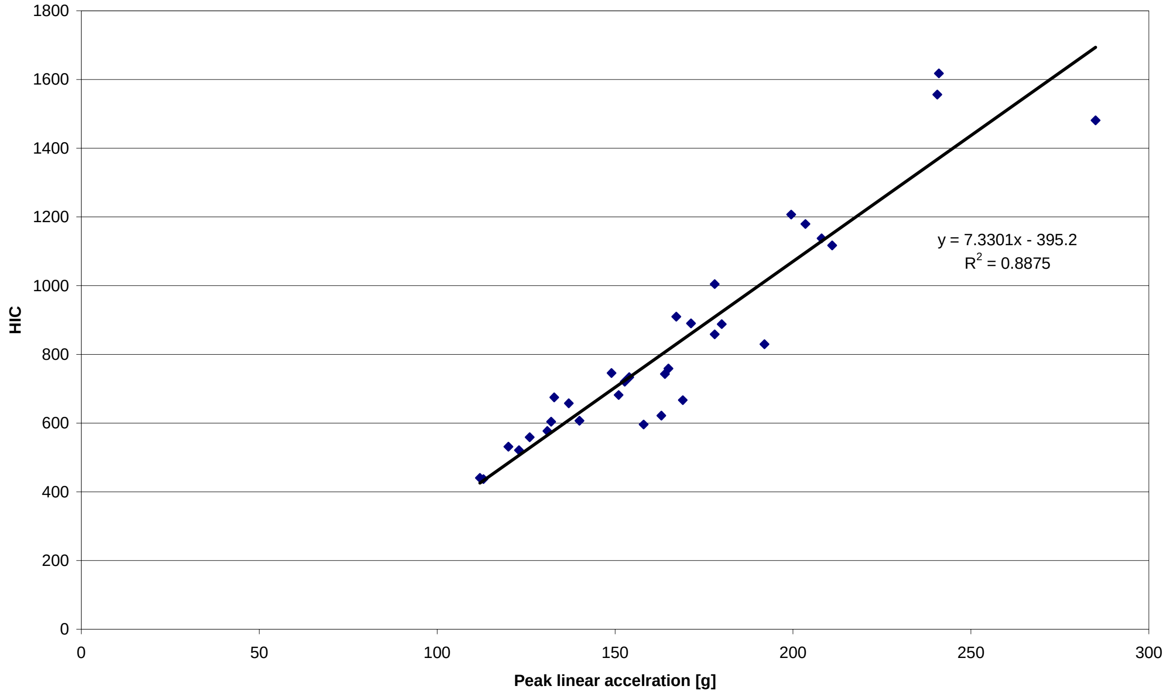

Figure 4.3 Peak linear acceleration versus HIC.

HIC consequently reflects the same trends observed for peak linear acceleration and Figure 4.4 illustrates that the MET Buddy and Bell Faction helmets again have the poorest combined results for the kerbstone and flat anvil tests. These higher HIC levels were not exceptional and in total, four helmets (seven impacts) exceeded HIC 1000, a threshold which is widely considered in the automotive industry as unsafe. Three of these helmets were of the smallest child designs.

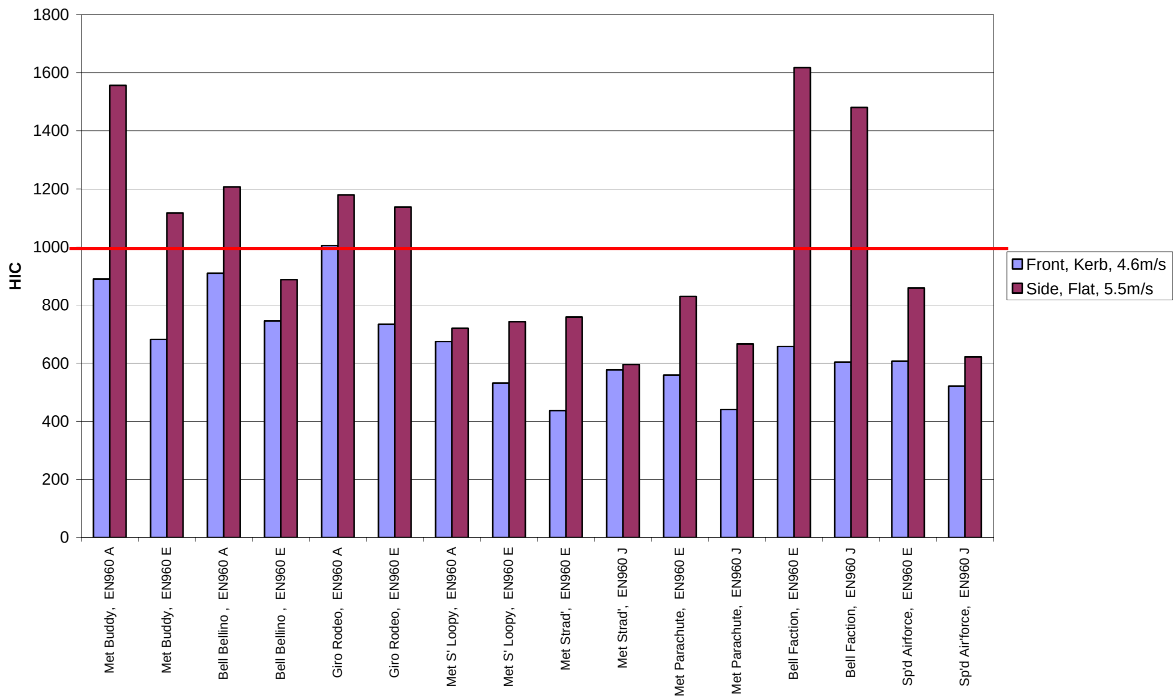

Figure 4.4 HIC for flat and kerbstone anvil tests.

It is interesting to note that HIC 1000 was exceeded during the same impacts in which the peak linear acceleration was 200g or more. This confirms the correlation trend that shows 200g corresponding approximately to HIC 1000. For the cases where 200g was exceeded, the exceedence duration was calculated and found to be less than 1.7ms. It can be concluded that HIC 1000 is probably a more stringent requirement than the AS/NSZ - 200g for 3ms.

<u>Other observations</u>

The kerbstone anvil impacts gave on average 55g less than the flat anvil tests and that youth / adult style helmets gave on average 24g lower peak accelerations than child style helmets (excludes helmets tested outside manufacturer's size range). These observations are somewhat as expected since the kerb anvil tests have a lower impact energy but are more aggressive than the flat. Consequently, a lower acceleration due to greater helmet penetration would be expected provided the helmet liner does not bottom out. The youth / adult helmets were usually larger and therefore have greater energy absorbing potential. This may partly explain a slightly better impact performance because the likelihood of bottoming out is reduced.

In all tests a rigid anvil was used. This anvil is representative of the worst case impact onto a rigid surface such as the road. In real life it is of course likely that helmet impacts will occur onto other surfaces which may be deformable. In such cases some energy can be absorbed by the impact surface and this allows the helmet to provide a level of protection at much higher impact speeds than tested here. Nevertheless, in some cases the impact surface may deform sufficiently to spread the load over a greater area of the helmet. In such instances the effective stiffness of the helmet increases and the injury potential may also increase. This has not been investigated here as some benefit would likely be offered to the rider than if no helmet was worn (see below).

In summary, analysis of the linear impact test data has shown differences in the impact performance of the helmet range such that the worst helmet performance gave a probability of fatality of approximately 53.1% whereas the best performing helmets were below 0.4% for the same impact conditions. It would appear that the helmets with a hard shell gave a worse performance than the others. However, there were insufficient tests to conclude that a hard shelled helmet would necessarily perform differently or afford less protection than other helmet types. Indeed, for some impact conditions that are not assessed here e.g. penetration of the shell, a stiffer helmet shell may provide additional levels of protection. A more detailed investigation would be necessary to evaluate this fully.

<u>Comparison with no helmet</u>

Research using post-mortem human surrogates carried out at Wayne State University (Wayne State, 1971) has demonstrated the tolerance of the non-helmeted heads to impact injury. The research has identified that linear skull fractures of AIS 2 severity may occur at speeds as low as 2.2m/s when an un-helmeted head strikes a flat rigid surface. At impact velocities above 3.2m/s, compound skull fractures are likely and may also result in tearing of the dura layer. An injury of this type equates to a severity of AIS 4, but would rate as an AIS 5 injury if there were any soft tissue damage to the brain. An increase in impact energy above the threshold for compound skull fracture is likely to cause damage to the brain tissue and cause life-threatening injuries.

Given that an AIS 4 injury approximates to 200-250g (Newman, 1986), as per Table 4.7 an AIS 4 or greater injury would only be expected for 6 out of the 32 helmet impacts in an impact onto a kerb and flat anvil in this study. The impact energy was between 2.1 and 2.9 times as great during these impacts than that required to cause similar injury on an un-helmeted head (at 3.2m/s). This demonstrates that helmets provide a likely injury benefit at higher impact speeds.

Test data from low speed impacts using a 54cm (E) headform on the kerbstone and flat anvil were used to calculate the likely distribution of AIS head injury against impact velocity. The analysis assumed that the helmet performance was not rate dependant. In practice helmets tend to be rate dependent between 0 m/s and 1m/s impact velocity and thereafter approximately linear.

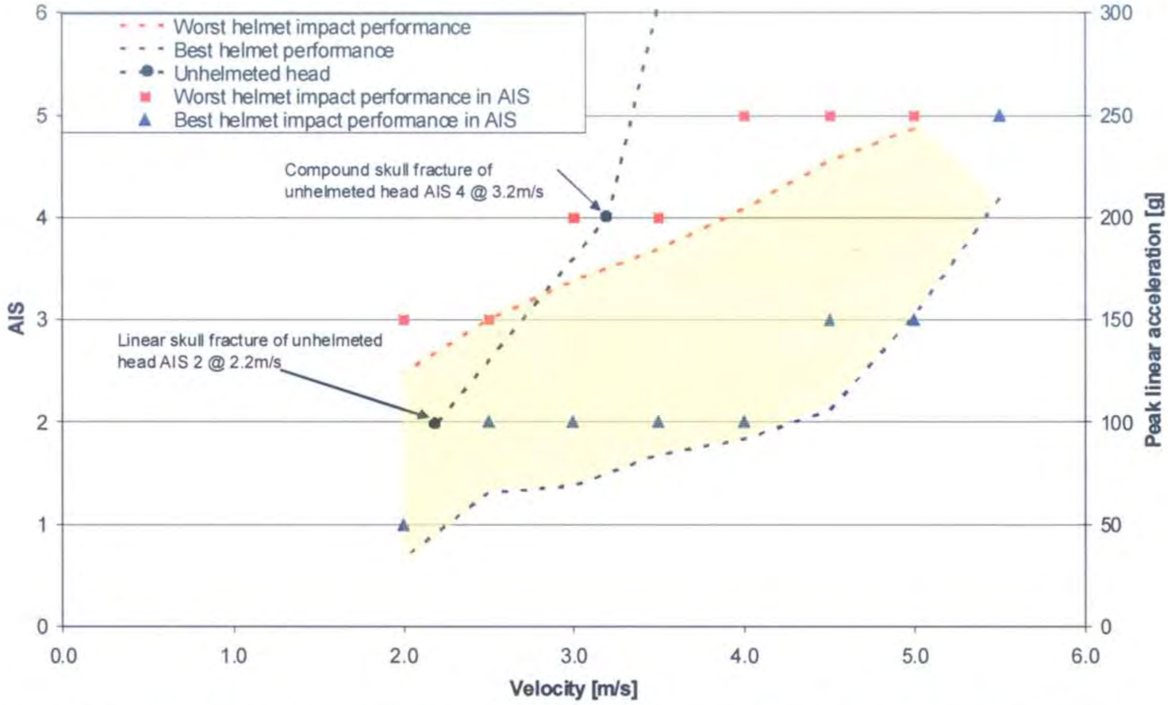

Figure 4.5 AIS and linear acceleration versus impact test velocity (based on size 54 helmet performance)

Figure 4.5 illustrates the helmet performance related to injury potential generated using the test data and the values of linear acceleration that correspond to simple and compound skull fractures identified from post-mortem human surrogate tests. The right vertical scale shows the linear acceleration in g (input) and the left vertical scale shows the injury potential as AIS values (output) for a range of impact velocities. It should be noted that AIS can only be discrete integer values whereas linear acceleration is continuous.

Above 3.2m/s the response of an un-helmeted head is not fully documented but is likely to be similar to that represented by the dashed line since the onset of brain soft tissue damage is likely for small increases in impact energy once skull fracture has occurred. Such injuries have AIS 5 or AIS 6 severities. The yellow area indicates the range of helmet performance observed for the helmets tested and is bounded by the upper (red) line and lower (green) lines which show the least and most severe injury outcomes for these helmets.

Except for the worst performing helmets, the injury potential is less for a helmeted head than an un-helmeted head throughout the range of test velocity. Between 2.2m/s and 3.2m/s it can be seen that, potentially, the injury outcome for a helmeted head may be more than 1 AIS increment lower than the un-helmeted headform. However, it is also significant that for the worst helmet performance observed there is potentially a detrimental injury outcome for helmeted heads below 2.8m/s.

Between 3.2 m/s and 5.0 m/s the injury outcome is probably two to three increments lower for a helmeted than an un-helmeted head. This is consistent with the fact that helmets are designed to provide optimal protection at 4.4 m/s (kerbstone) and 5.4 m/s (flat anvil). These curves show that above 3.2m/s the potential benefits, even for the worst performing helmets, are significant and potentially life saving. It is potentially justified to demand a low speed (less than 2.8m/s) test with a lower maximum permitted acceleration because the injury outcome at this speed equates approximately to AIS 3 (severe) even though the risk of fatality is low and below 2.1%.

4.3.3 Oblique impact test results

The results for the oblique impact tests are given in Tables 4.8 and 4.9 and in graphical form in Appendices C through to D. For oblique abrasion tests, the graphical data (Appendix C) illustrates the dynamic coefficient of friction, calculated as the average tangential anvil force divided by the average normal anvil force for the period where the normal force exceeds 70% of the maximum recorded during the entire impact duration.

The test data have been split into groups relating to the following sub-divisions;

1) Abrasive anvil tests with standard Reg 22.05 test configuration;
2) Abrasive anvil test with varied head sizes (abrasive and projection);
3) Abrasive impacts with reduced impact speed;
4) Projection strength tests with standard Reg 22.05 test configuration.

The parameters allow the comparison of oblique impact performance for helmets when similar impact conditions have been used and can be related to the injury probability and threshold limits defined by historical research and regulations.

Table 4.8 Results for oblique impact onto 15º abrasive anvil using EN960 54cm (E) headform at a nominal speed of 8.5m/s

	Helmet	Impact Site	Test ref	Peak linear acc [g]	Peak rotation acc. [rad/s²]	Peak rotation velocity [rad/s]	Peak Normal force [N]	Peak Tangential force [N]	μ
1	Met Buddy	side - right - face down	x06dy	48	5307	46	2468	1092	0.5
		crown - face down	w06dy	81	5931	37	3503	1664	0.6
2	Bell Bellino	side - right - face down	v06dy	52	3972	39	3027	1572	0.6
		rear - side right down	**z07dy**	15	3453	44	687	366	0.7
		rear - right side down	**b20gy**	26	3378	43	1024	554	0.7
3	Giro Rodeo	side - right - face down	y07dy	83	6820	41	4849	2476	0.6
		crown - face down	**a20gy**	42	3174	30	1955	707	0.4
4	Met Super Loopy	crown - face down	v07gy	59	4321	31	3201	1408	0.5
		side - right - face down	w07dy	43	4512	38	2275	1121	0.5
5	Met Stradivarius	side - right - face down	a24cy	61	3077	34	3443	1748	0.6
		side - right - face down	b24cy	35	2810	34	1628	646	0.5
		crown - face down	c24cy	56	4148	28	2906	1419	0.5
6	Met Parachute	side - right - face down	z06dy	40	4502	36	1937	864	0.5
		crown - face down	y06dy	56	4267	32	2907	1656	0.6
7	Bell Faction	side - right	c05dy	67	5821	48	3579	1773	0.6
		rear – side right down	e05dy	67	8509	52	3302	1658	0.6
8	Specialised Airforce	front right - on microshell - face down	d24cy	80	3461	30	3981	2079	0.6
		rear – on EPS	a05dy	**98**	5341	37	4600	2997	0.8

Table 4.9 Results for oblique impact onto 15º abrasive anvil using variable headform sizes at a nominal speed of 8.5m/s

	Helmet	Impact Site	Head size (EN960)	Test ref	Peak linear acc [g]	Peak rotation acc. [rad/s²]	Peak rotation velocity rad/s]	Peak Normal force [N]	Peak Tangential force [N]	μ
1	Met Buddy	crown - face down	E	w06dy	81	5931	37	3503	1664	0.6
		crown - face down	A	c25gy	94	0	0	3188	1394	0.6
3	Giro Rodeo	side - right - face down	E	y07dy	83	6820	41	4849	2476	0.6
		side - right - face down	A	d25gy	94	0	0	3188	1394	0.6
7	Bell Faction	side – right	E	c05dy	67	5821	48	3579	1773	0.6
		side – right	J	a21gy	73	16224	89	2887	1517	0.6
8	Specialised Airforce	rear - on EPS	E	a05dy	98	5341	37	4600	2997	0.8
		rear - on EPS	J	b21gy	43	10737	88	2494	1916	0.8
1	Met Buddy	visor - crown down	E	a21dy	39	5138	39	1843	1089	0.8
		visor - crown down	A	a25gy	49	0	0	1237	733	0.8
2	Bell Bellino	longitudinal ridges - crown down	E	d21dy	88	6738	37	4526	2521	0.7
		longitudinal ridges - crown down	A	b25gy	106	0	0	3133	1992	0.9
		visor - crown down	E	a19gy	34	4039	30	1068	716	1.0
7	Bell Faction	nose down - aiming to catch central vents	E	n21dy	68	4668	31	3798	1851	0.6
		nose down - aiming to catch central vents	J	e21gy	134	20642	118	5817	2365	0.5
8	Specialised Airforce	crown - side down - to catch ridges	E	o21dy	53	3507	29	2939	1502	0.6
		crown - side down - to catch ridges	J	c21gy	76	6420	77	2785	1276	0.6

Table 4.10 Results for oblique impact onto 15º abrasive anvil using 54cm (E) headform at variable impact speeds

Helmet		Impact Site	Test ref	m/s Velocity [m/s]	Headform Energy [J]	Peak linear acc [g]	Peak rotation acc. [rad/s²]	Peak rotation velocity [rad/s]	Peak Normal force [N]	Peak Tangentail force [N]	μ
1	Met Buddy	crown - face down	w06dy	8.65	157.8	81	**5931**	37	3503	1664	0.6
		crown - face down	b27gy	6.09	78.2	**68**	**4704**	26	2916	1286	0.6
4	Met Super Loopy	side - right - face down	w07dy	8.63	157.0	43	4512	38	2275	1121	0.5
		side - right - face down	d27gy	6.08	78.0	36	3908	33	1494	638	0.6
5	Met Stradivarius	side - right - face down	b24cy	8.61	156.4	35	2810	34	1628	646	0.5
		side - left - face down	a27gy	6.09	78.1	42	2502	17	1793	986	0.8
8	Specialised Airforce	rear - on EPS	a05dy	8.63	157.1	**98**	5341	37	4600	2997	0.8
		rear - on EPS	c27gy	6.04	76.9	36	4442	38	1508	900	0.8

Table 4.11 Results for projection strength tests onto 15º bar anvil using EN960 54cm (E) headform at a nominal speed of 8.5m/s.

	Helmet	Impact Site	Test ref	Peak linear acc [g]	Peak rotation acc. [rad/s²]	Peak rotation velocity [rad/s]	Peak Normal force [N]	Peak Tangential force [N]	μ
1	Met Buddy	visor - crown down	a21dy	39	5138	39	1843	1089	0.8
		longitudinal ridges - crown down	b21dy	62	3686	32	3306	1587	0.5
2	Bell Bellino	visor - crown down	c21dy	42	2803	30	1553	983	0.6
		longitudinal ridges - crown down	d21dy	88	6738	37	4526	2521	0.7
		visor - crown down	a19gy	34	4039	30	1068	716	1.0
3	Giro Rodeo	visor - crown down	e21dy	53	4098	41	2493	1611	0.8
		longitudinal ridges - crown down	f21dy	88	3867	31	4516	2776	0.7
		visor - crown down	b19gy	41	5296	38	1616	1083	0.9
4	Met Super Loopy	visor - crown down	g21dy	44	2654	20	1314	815	0.7
		longitudinal ridges - crown down	h21dy	58	3330	25	3034	1508	0.6
5	Met Stradivarius	tail - crown first	i21dy	37	1361	15	1400	469	0.4
		front right chinstrap anchorage tabs	j21dy	32	1980	22	1166	556	0.6
		crown - side down - to catch ridges	k21dy	37	2347	28	1849	822	0.6
		tail (2 contact points) - crown first	c19gy	24	1155	10	611	331	0.7
6	Met Parachute	side - right - chin anchorage bolt	m21dy	57	2908	39	2759	1509	0.7
		chinguard - right side down	d19gy	35	2596	25	454	481	1.3
7	Bell Faction	nose down - aiming to catch central vents	n21dy	68	4668	31	3798	1851	0.6
8	Specialised Airforce	crown - side down - to catch ridges	o21dy	53	3507	29	2939	1502	0.6
		tail - crown first	p21dy	27	1701	23	966	572	0.4

4.3.3.1 Abrasive test results

Linear acceleration, HIC and normal force

Despite an increase in impact speed from 5.42m/s for the linear impact tests to 8.5m/s for oblique tests, the peak linear accelerations recorded were on average 136g (192g - 56g) lower for oblique impacts than the linear impact counterparts (flat anvil tests using 54cm (E) headform). This is significant as it illustrates that the overall severity of an impact is not simply a function of the impact speed but is also dependent on the impact angle. In fact the normal impact component of a 15º oblique impact is equivalent to a 90º impact with 26% of the impact speed. Hence, at 8.5m/s (and 15º) an equivalent linear test would be at just 2.2m/s.

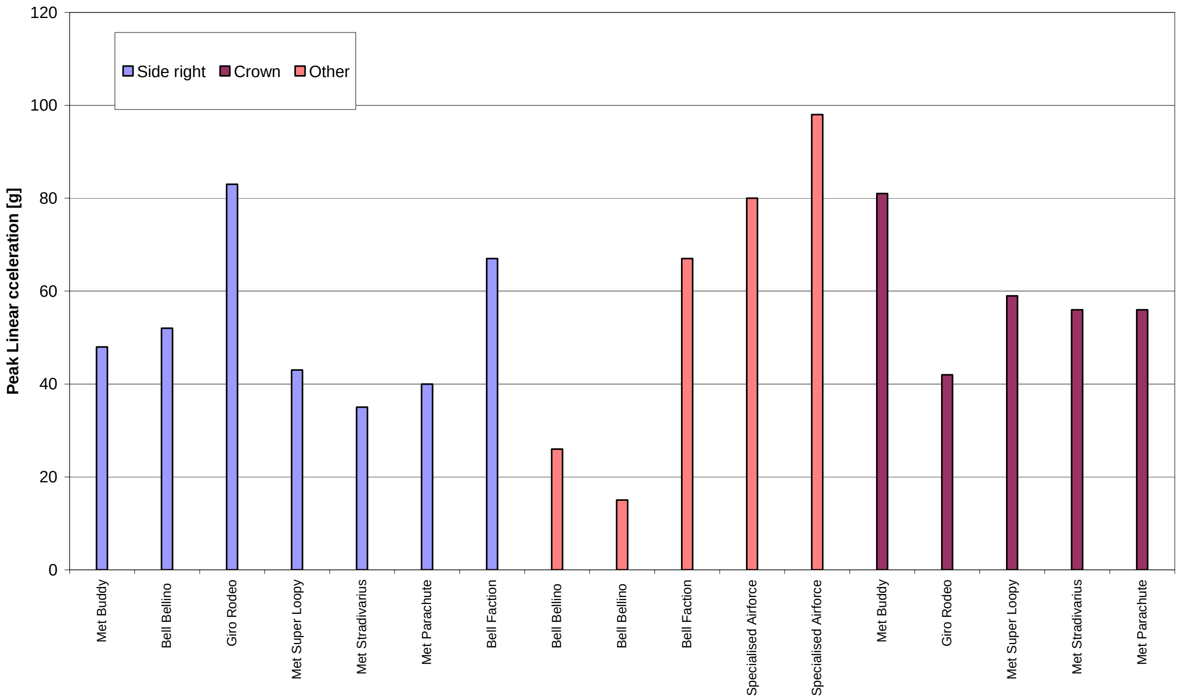

Figure 4.6 Linear acceleration performance of test helmets during oblique abrasive impacts

Predictably, the levels of peak linear acceleration recorded during the oblique impacts are well below the thresholds set by current bicycle helmet standards. Figure 4.6 shows that the maximum acceleration recorded for all tests was just 98g (Specialised Airforce, rear) which is 143g (241g - 98g) lower than the maximum result for a 90º flat anvil at 5.42m/s (using 54cm (E) headform) and 152g below the EN1078 limit of 250g.

Based on a relationship between peak linear acceleration and AIS (Table 4.12), the linear acceleration component of these oblique impacts is judged to have the potential to cause relatively minor head injury of typically AIS 1. This may include headache or dizziness. The lowest peak linear acceleration was just 15g (Bell Bellino, rear) and unlikely to cause any head injury. The average linear acceleration measured for all tests was 56g.

Table 4.12 Injury outcome to peak linear acceleration

Peak linear acceleration	AIS head injury severity	Injury interpretation	Approximate probability of fatality
< 50g	0	No injury	0.0%
50 - 100g	1	'Minor' injury	0.0%
100 - 150g	2	'Moderate' injury	0.1 - 0.4%
150 - 200g	3	'Serious' injury	0.8 - 2.1%
200 - 250g	4	'Severe' injury	7.9 - 10.6%
250 - 300g	5	'Critical'	53.1 – 58.4%
> 300g	6	'Unsurvivable' (Maximum)	>58.4%

A similar trend is noted for HIC levels which were very low for oblique tests when compared with the linear impact test results. The average HIC recorded for the 12 oblique tests was HIC 97 and less than 10% of the HIC 1000 level which is generally considered unsafe. This is as expected since a good correlation between linear acceleration and HIC is known to exist (Figure 4.3) for linear impacts.

The contribution of the linear acceleration during oblique impacts has been shown to have low injury potential,, nominally less than AIS 2 severity. However, a 15º anvil test is intended to exert high torque loads on the headform causing significant rotation. It is therefore important that parameters such as tangential force, as considered by Reg 22.05, and rotational acceleration induced in the headform are used to evaluate protective capabilities of the helmet against the rotational injuries likely during these impacts.

Tangential force and friction coefficient

The Reg 22.05 oblique impact test adopted for this investigation is optimised for motorcycle helmets and the impact conditions are considered somewhat worst case for bicycle helmets. Despite this, the tangential force was consistently less than the 3500N maximum prescribed by Reg 22.05. The maximum tangential force recorded was just less than 3000N with an average result of 1415N.

Table 4.13 summarises a correlation between tangential force and injury outcome which was verified by COST 327. At 3500N the probability of an AIS 3 injury outcome was estimated to be below 50%. At 3000N, as observed for the bicycle helmets tested, the likely injury would be below this level but greater than AIS 2 severity for this particular oblique impact. The average value of 1415N, observed during these helmet tests would likely result in an AIS 2 injuries or lower.

Table 4.13 Injury outcomes related to peak tangential force on a motorcycle helmet (COST 327)

Peak tangential force	**AIS**	**Probability of injury**
1000 N	1	-
2000 N	2	-
3500 N (Reg 22.05)	3	<50%
4000 N	3	50%
7000 N	4	-

AIS 2 or less is the injury outcome predicted if the linear acceleration component only is considered. If the rotational component is included then an injury potential of up to AIS 3 (probability <50%) is predicted. These results suggest that the rotational component of these oblique impacts would therefore have the greatest injury potential. This is an opinion which is generally gaining acceptance in the scientific community with increased levels of research into rotational injury mechanisms. It should however be noted that the permitted tangential impulse of 12.5Ns was not exceeded in any of the tests.

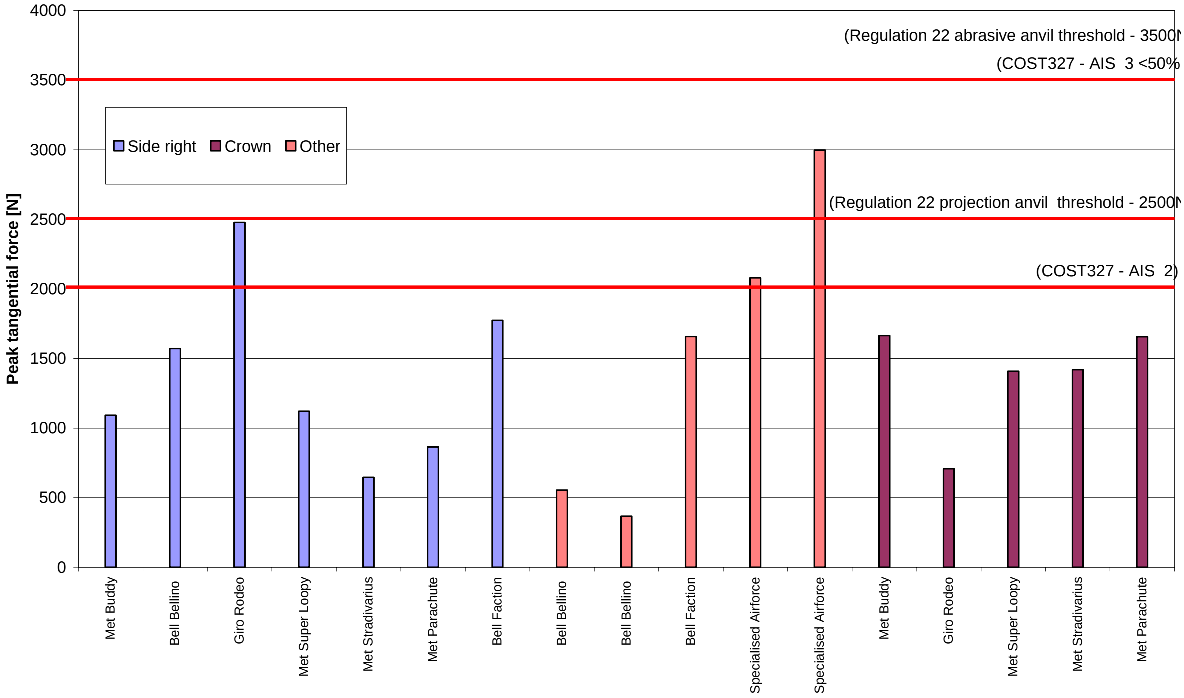

Figure 4.7 Peak tangential force for oblique abrasive impacts.

Theoretically, during abrasive impacts, the peak tangential anvil force is a direct function of the coefficient of friction between the helmet and the impact surface. However, a poor correlation between tangential force and the coefficient of friction was observed. This may be explained by the geometrical differences between the helmets and the complex dynamics of the impact event, during which the helmet may rotate relative to the headform prior to the peak loading conditions. The direction and extent of helmet rotation may influence the load transmitted to the head due to local variations in the helmet performance. This behaviour varies between helmets and cannot be adequately controlled to allow a fair comparison of tangential force between helmets. This is important as it shows that tangential force and coefficient of friction may not be good measures for comparison of helmet performance but may still be indicative of the impact severity and therefore injury risk.

Although the measured coefficient of friction is sensitive to impact site and other parameters, it was found to be approximately constant between helmets of the same type. For all helmets tested, the coefficient of friction ranged from $\mu = 0.42$ to $\mu = 0.78$ (average $\mu = 0.57$) and was comparable to levels typically observed for motorcycle helmets. The differences between helmets were attributed to variations in the helmet geometries and material properties, because friction is dependent on the mechanical interaction which varies due to helmet deformation. Despite this view, no clear trends were observed to suggest that there was a particular impact site or material which performed consistently badly. In fact, impacts onto the microshell plastic materials gave both the highest and lowest friction values observed.

Table 4.14 Average friction coefficient for oblique impact helmet tests

Helmet	Average friction coefficient (μ)
Bell Bellino	0.66
Bell Faction	0.58
Giro Rodeo	0.52
Met Buddy	0.55
Met Parachute	0.54
Met Stradivarius	0.51
Met Super Loopy	0.51
Specialised Airforce	0.68

Although the friction and tangential anvil forces varied among the helmets tested here, tangential force correlated well with peak headform acceleration (r^2=0.95) and peak normal force (r^2 = 0.97). This illustrates the likely importance of linear acceleration in the resulting head impact severity. This suggests that a reduction of linear acceleration in a given impact would result in reduced tangential force and a reduction in head injury potential due to rotation.

The tangential force generated on the helmet has been shown to relate to injuries of AIS 3 severity or less. However, tangential anvil force varied on tests with similar helmets and hence may not be the most reliable measure for comparing the ability of helmets to protect against injury caused by rotational motion. Helmet geometry, the location of impact sites and helmet fit are factors which may each affect the relationship between tangential force and rotational motion and the injury potential. Although tangential force may be indicative of the likely head injury, rotational acceleration measured in the headform is more directly related to brain injury caused by rotation and may allow the best comparison between helmets.

Rotational acceleration

An experimental study to characterise the relationship between anvil loads and rotational acceleration for bicycle helmets would need to be very large and is beyond the scope of this study. However, TRL has completed all 54cm (E) abrasion tests using a nine accelerometer array to measure directly the rotational acceleration in the head during oblique impacts. Analysis of these results, shown in Figure 4.8, shows that for the 15º oblique abrasion tests (at 8.5m/s), the relationship between anvil forces and headform rotational acceleration is poor (r^2 = 0.29).

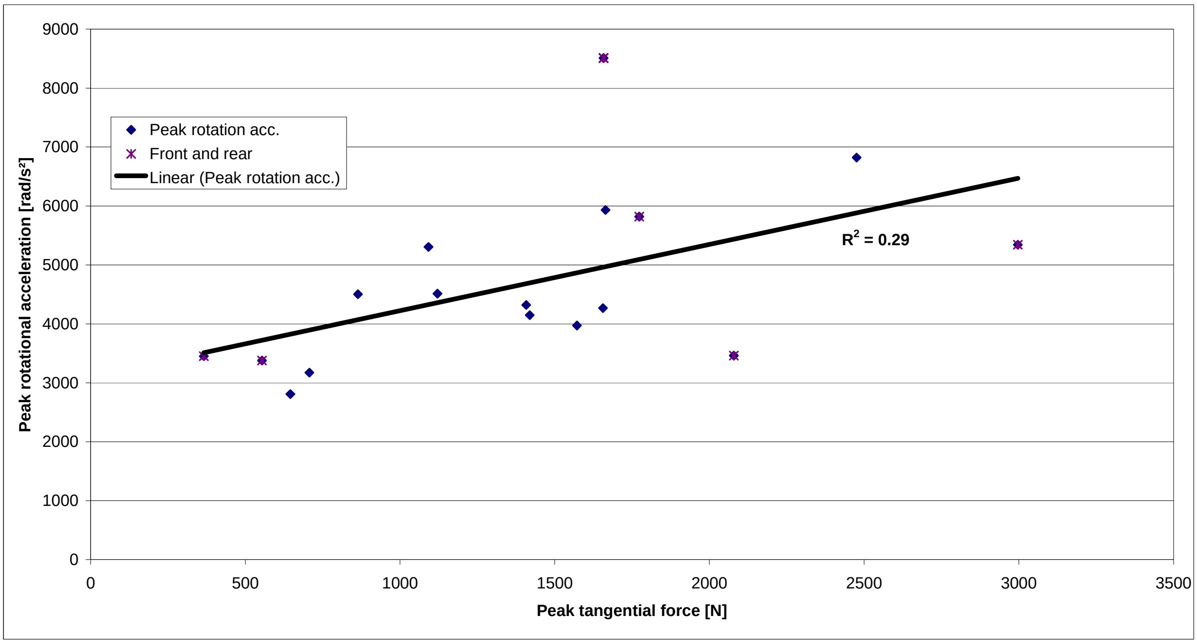

Figure 4.8 Variation of rotational acceleration with tangential force for abrasive impacts

Further analysis using data from helmet tests with near identical impact conditions, i.e. onto the same impact site with similar head orientation, show improved correlations (r^2 = 0.61 and 0.67 for crown and right side impacts respectively) as illustrated by Figure 4.9. This result indicates that rotational acceleration may correlate well with tangential force when similar test configurations are used (e.g. head orientation, impact site). However, this result may also imply that linear impact performance is a particularly dominant factor. Figure 4.10 below, supports this view with improved correlations calculated between linear acceleration and rotational acceleration with the r^2 ranging from 0.69 to a near perfect correlation of 0.996.

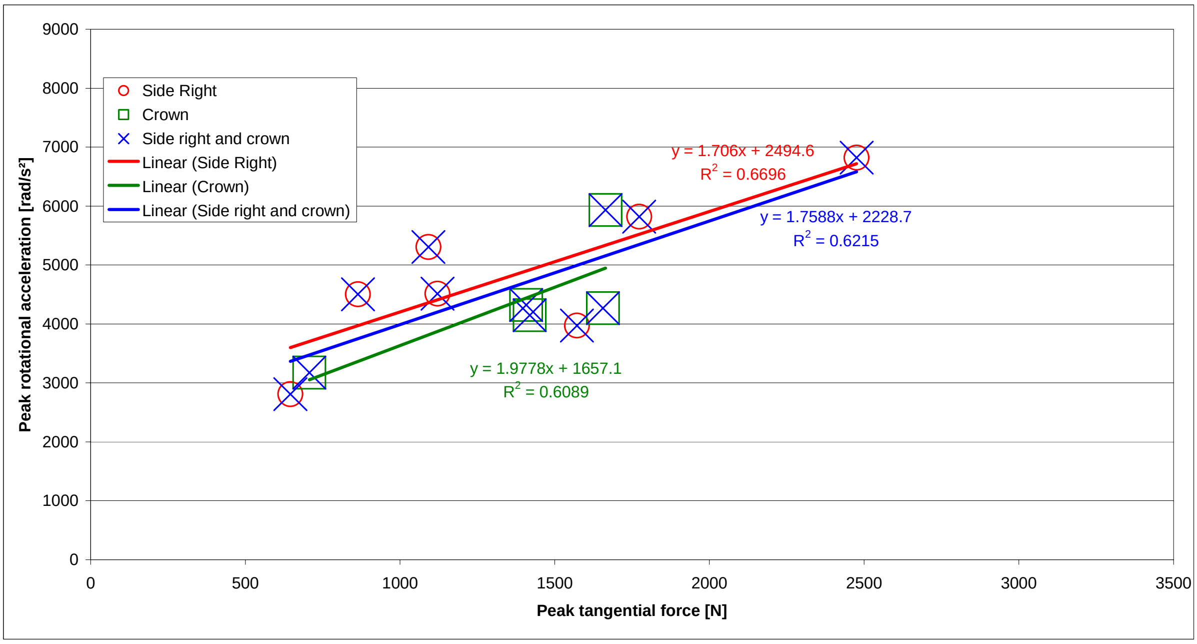

Figure 4.9 Variation of rotational acceleration with tangential force for abrasive impacts with similar impact configurations

Based on right-side impact data (where correlation between tangential force and rotational acceleration is r^2=0.67) presented in Figure 10, the rotational acceleration at the maximum recorded tangential load (2997N) equates to just over 7600rad/s². This value is almost 11% below the 8,500rad/s² measured in the headform using the nine-accelerometer array. This indicates that, even with a relationship between tangential force and rotational acceleration, the force measurement may predict too low an injury potential.

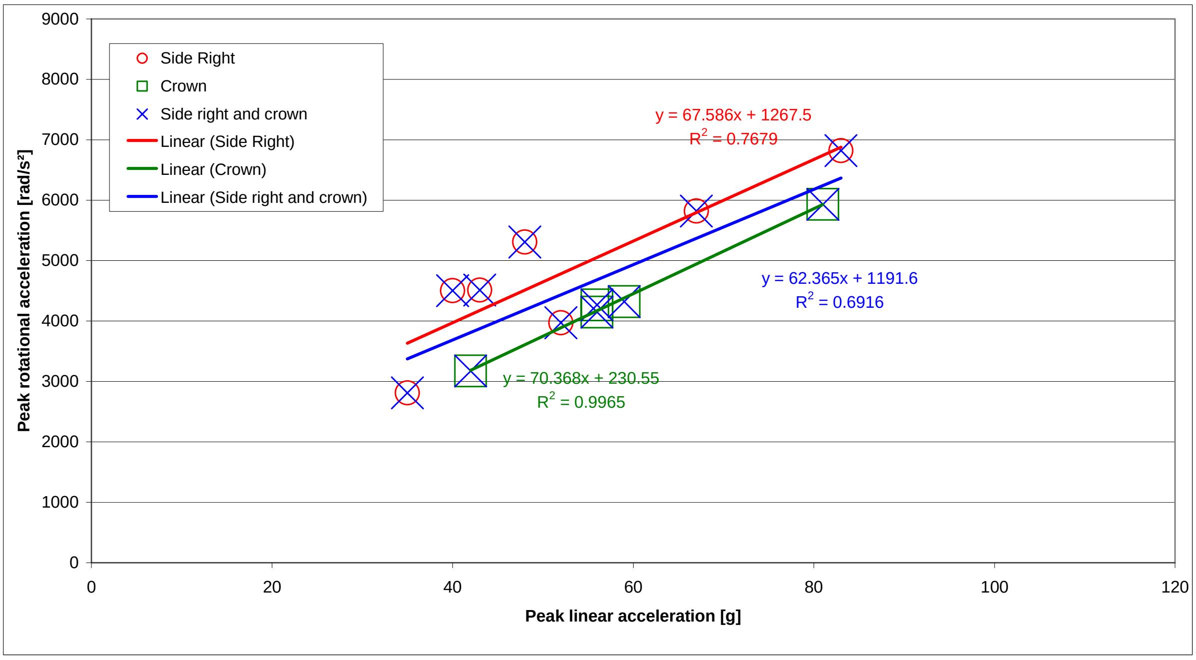

Figure 4.10 Variation of rotational acceleration with linear acceleration for abrasive impacts with similar impact configurations

It must, therefore, be concluded that the direct measurement of the headform rotational acceleration is the most appropriate measure for an experimental assessment to evaluate potential risk of injury, but tangential force may be appropriate to compare helmet performance for controlled impacts with similar impact conditions e.g. helmet impact site.

Fortunately, the literature review has demonstrated that head rotational acceleration is also well researched and human tolerance limits have been identified which suggest a 35% probability of an AIS 3-6 head injury at 10,000rad/s² and at 5000rad/s² an AIS 1-2 is likely. COST 327 has accurately predicted injury outcome in 25% for a series of detailed motorcycle accident replications. Here, AIS 2 injuries are predicted @ 8,000rad/s² and AIS 3 at 19,000rad/s². A 50% risk of AIS ≥3 injuries was predicted at 30,000rad/s² although fatal injuries may be as low as 30,000rad/s². However, alternative data from the replication of a single accident involving an unhelmeted child (Lowenhielm) has reported a fatal brain injury may occur at a rotational acceleration as low as 4,500rad/s². Lowenhielm also reported that the onset of severe brain injury occurs at a rotational velocity of 60rad/s.

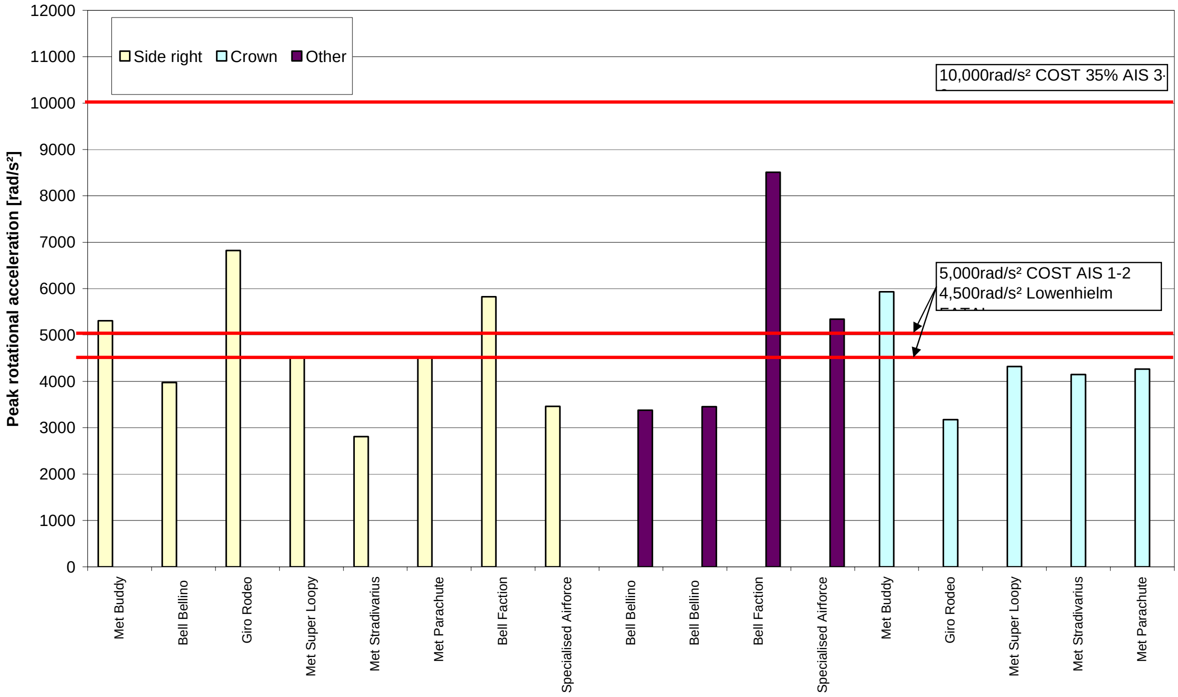

Figure 4.11 Rotational acceleration for oblique abrasive impacts

Test data obtained from the oblique impact tests using a 54cm (E) instrumented headform are given in Figure 4.11. The figure shows that for all impacts the peak rotational acceleration was below 10,000rad/s² at which a 35% probability of AIS 3 - 6 injuries is predicted and in the majority of cases is less than the reported 8,000rad/s² threshold for AIS 2. This is reassuring as these impacts were of a severity intended for motorcycle helmets. Furthermore, only 6 of the 17 impacts resulted in rotational accelerations above 5000rad/s², a level at which AIS 1 - 2 injuries is estimated. However, this accounted for half of the helmet models tested and only four models achieved test results consistently below 5000rad/s².

Generally, those helmet models which achieved a poor rotational acceleration performance also gave a poor performance during linear impact tests. This reiterates the earlier view that linear impact performance may be a dominant parameter controlling the rotational acceleration response during oblique impacts. The Bell Faction was the worst performing helmet overall, achieving a peak rotational acceleration of 8509rad/s² during an impact to the rear of the shell. A more respectable result of 5821rad/s² was achieved during a right side impact. Both impacts produced 67g linear impact component and the coefficient of friction was 0.55 and 0.60 respectively.

To establish whether the construction of the helmets has a significant effect on the rotational acceleration, impact test data obtained during right side impacts alone were scrutinized. Here the Bell Faction, of a stiff outer shell construction, did not generate the highest rotational acceleration and was instead 23% lower than the Giro Rodeo which gave the highest result of 6820rad/s². Furthermore, the Giro Rodeo, which has a microshell construction, gave results between 30% (Met Buddy) and 70% higher (Bell Bellino) than helmets of similar construction. The MET Stradivarius, which used an in-mould microshell construction, gave the lowest results of 2810rad/s², yet the MET Super Loopy, which utilizes a similar in-mould construction, was almost 61% higher.

There is insufficient data to show whether the helmet shell material has a direct influence on the rotational accelerations imparted to the headform or whether any particular materials are particularly more hazardous. Although the limited data may suggests that no relationship exists, it should however be remembered that the rotational acceleration performance appears to be dependent on the linear impact performance. It is thought that the linear impact performance may be construction related but could be improved by optimized helmet design.

In this study, the maximum peak rotational acceleration measured corresponded to an injury severity of less than AIS 3 based upon motorcycle accident impact conditions. Half of the helmets recorded a value of greater than 5000rad/^{2} at which AIS 1 - 2 is predicted.

<u>No helmet</u>

Generally, the range of helmets evaluated gave an oblique impact performance that would satisfy the requirements of current motorcycle helmet standards. The risk of AIS 3 - 6 injuries was low and a head injury of AIS $\leq$2 was most likely, it is important to consider the outcome for an oblique impact event when a helmet is not worn to see whether helmet wearing has a positive or detrimental affect is likely.

Unfortunately, there is little data to predict the human head response to an oblique impact and instead the likely injury outcome must be related to post-mortem human surrogate data for linear impacts (see 4.3.2). For the impact conditions considered (8.5m/s at 15º), the normal impact component is equivalent to a linear impact at 2.2m/s. This impact speed equates to a simple skull fracture for an unhelmeted head, an injury rated as AIS 2 (moderate) which occurs at a force of typically 5kN to 6kN. This corresponds to between approximately 100g and 120g for a 5kg mass head, and 125g to 150g for a 4kg mass head.

This study has shown that there is a reasonable correlation between linear impact performance and rotational impact performance and thus it may be concluded that the greater the linear acceleration in an oblique impact the greater the rotational acceleration. If the graphs shown in figure 4.10 are extrapolated then it can be shown that 100g to 150g correlates with about 7500rad/s^2 to about 12000rad/s^2. Such accelerations would be potentially more severe than the 3000rad/^{s} to 7000rad/s^2 obtained for the tests with helmets.

However, this analysis assumes that the response of the unhelmeted head is similar to the helmeted head with a similar correlation between linear and rotational acceleration and does not account for other factors, the most important of which are diameter and mass. A helmeted head has increased mass and diameter which would tend to increase head loading whereas the smaller diameter of an unhelmeted head would tend to reduce the torque about the head’s centre of gravity. Although this may suggest that the head would have lowered rotational accelerations, the helmeted head has an increased moment of inertia which would also tend to reduce the rotational acceleration induced. Furthermore, a helmet may also allow greater slippage of the head relative to the impact surface thereby reducing rotation. Since the response of the scalp and skull during oblique impacts is not documented it is assumed that these factors will approximately balance one another and the above observations, based on linear impact performance, give a reasonable estimate.

It is beyond the scope of this study to perform experimental test work to characterise the unhelmeted human head response but it should be remembered that the levels of rotational acceleration reported in the previous section, equate to head injuries of around AIS 2 with a low risk of more severe injuries. If the sole contribution of linear acceleration in oblique impacts is considered, values of less than 100g were measured (average 56g) which corresponds to AIS 1 or less (less than 50g is AIS 0). For an unhelmeted head the range of 100g to 150g corresponds to AIS 2. Thus, it seems unlikely that a helmet would increase the risk of injury in an oblique impact onto a flat surface assuming that a projection of the helmet did not strike the surface; this type of impact is analysed in the next section.

It should be noted that the fracture values quoted were obtained with adult post-mortem human surrogates and consequently these values may differ to those for a small child for which the headform mass is about 3kg and skull compliance may differ. A lower tolerance to injury would suggest a potential benefit is more likely when a helmet is worn during abrasive oblique impacts.

4.3.3.2 Projection test results

Linear acceleration, HIC and normal force

For all of the projection evaluations completed, the average linear acceleration was 48g and close to the 56g average from the abrasive tests. The maximum and minimum values: 88g and 24g respectively, were around 10g lower than those recorded for the oblique abrasive tests. The 48g average corresponds to AIS 0, uninjured.

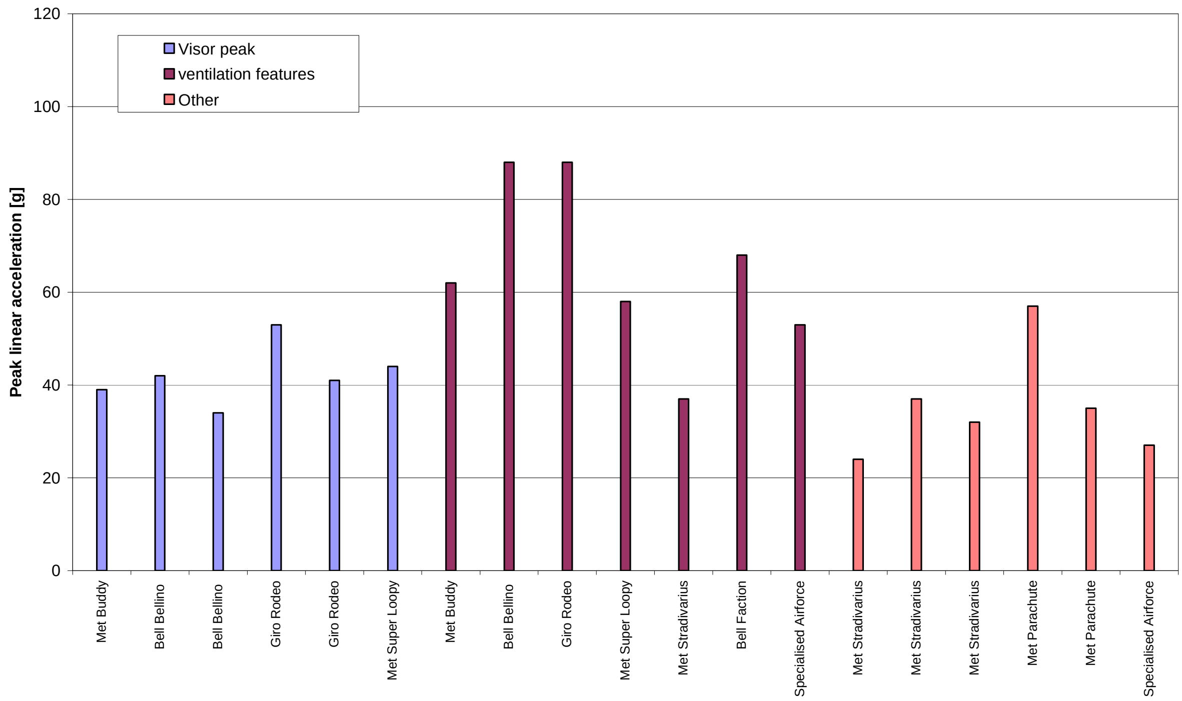

Figure 4.13 Peak linear acceleration measured during the oblique projection impacts

Linear acceleration was found to correlate well with HIC for the projection tests (r^2=0.93) and an average HIC of 75 was recorded. The HIC was well below HIC 1000 which is generally considered to equate to a 15% probability of fatality. Normal force also gave a good correlation with linear acceleration. Generally, the level of linear acceleration, normal force and HIC measured during projection tests were similar to those observed during the abrasion tests.

Tangential force

For motorcycle helmet testing, Reg 22.05 requires that all projections or irregularities in the outer shell surface greater than 2mm shall be tested for shear strength. Motorcycle helmets are predominantly smooth and such features are easy to define. Bicycle helmets are often irregular in shape because of styling and ventilation features. For this study, more detailed rules were defined so that the shell styling and ventilation features were evaluated. It was judged that such features may interlock with the bar anvil and impart significant rotational motions. The features tested included:

- Visor peaks;
- Protruding ridges due to ventilation geometry;

- Chinguard;
- Chinstrap and chinguard fasteners on the shell;
- Styling projections on the rear of helmets.

It was considered that these features of the helmet could interact with objects struck in an accident, the edge of car bodywork for example, impart rotational motion and increase the injury potential. Consequently, it was decided appropriate to assess the helmets against the 2500N maximum permitted tangential force as specified in Reg 22.05.

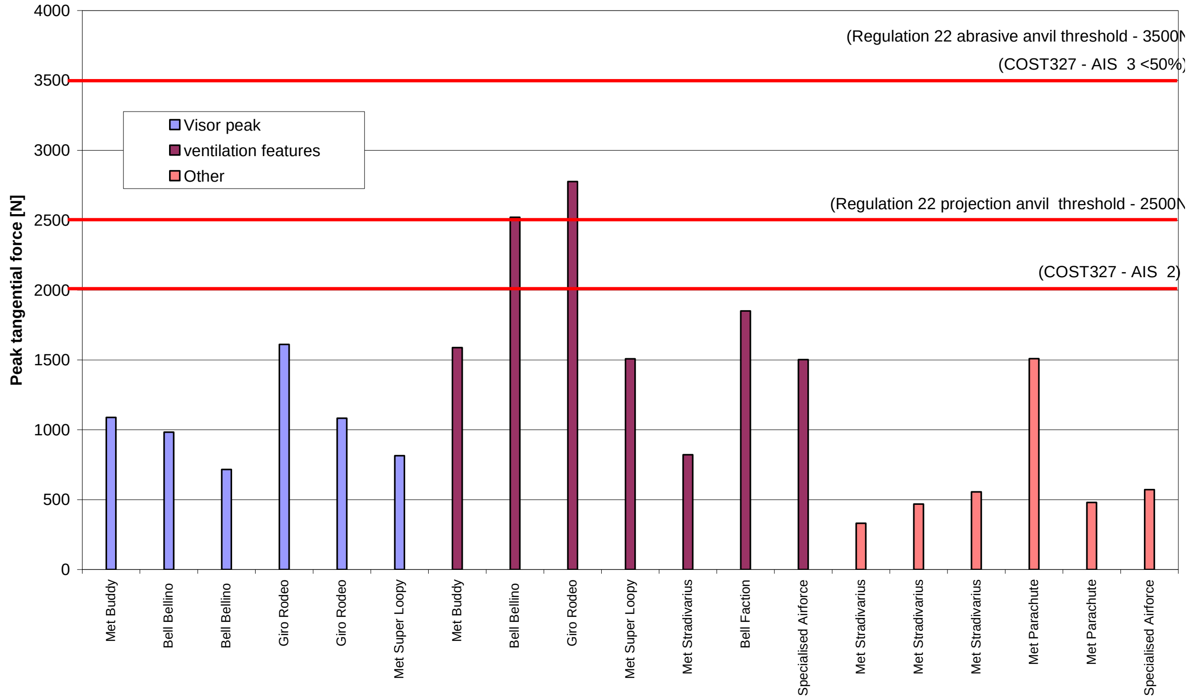

Figure 4.14 Peak tangential force during oblique projection impacts

Figure 4.14 illustrates that only two impacts onto the irregular helmet features exceeded the Reg 22.05 requirement for peak anvil force. These results of 2776N and 2521N were obtained for the Giro Rodeo and Bell Bellino respectively, when impacted transversely onto the longitudinal ridges on the crown area. These results were more than double the 1199N average value recorded for all helmet projection tests. However, it is clear that impacts to ventilation features generate the highest tangential loads with an average result of 1795N onto these features with all but one above the 1199N average recorded for all projection tests. The tangential forces generated by the two helmets that exceeded 2500N equate to an injury potential of AIS 2. For all other impacts the values equate to an AIS 0 -1.

Closer inspection of the test data revealed that helmets which generated the highest tangential loads also generated the highest normal forces with a good correlation between these parameters of r^2=0.93. For motorcycle helmets the projection normally sheers to ensure that 2500N is not exceeded. Appendix E illustrates the appearance of damage recorded for the bicycle helmet projection features tested including a typical:

a) visor peak;

b) longitudinal helmet profile;

c) chinguard and fastener;

d) ‘tail’ styling projection.

It was apparent from inspection of this damage that the lowest forces were generated when projections were weak or sheered off. For example, the Bell Bellino visor peak (Figure E.1) which broke off, generating about 716N force. Generally the forces were observed to be a function of the projection size and the method of attachment to the shell. For example, the MET Parachute helmet, featuring a rigidly attached chinguard, generated an above average 1509N force whilst breaking away a significant section of the helmet (Figure E.2).

Features tested, which did not break away, generated higher anvil forces. These forces also appeared to increase with the area of interaction. This explains the higher than average results obtained for tests onto longitudinal profiles where large rigid sections of the helmet were in contact with the bar anvil. In three such instances (MET Buddy / Giro Rodeo / Bell Bellino) the impact caused sufficient deformation for the bar anvil to penetrate into the helmet liner, thereby generating increased mechanical interaction (Figure E.3). Consequently much greater tangential forces, in some cases exceeding 2500N, were observed. These forces were typically 1500N higher than for other helmet features tested.

Although liner penetration, which caused mechanical interlocking with the Bell Bellino helmet liner could be reduced by the use of stiffer helmet shells, such shells can also generate high anvil loads where shell features can be impacted directly as they do not fail so readily. In fact, the Bell Faction gave the third highest tangential force of 1851N when impacted onto an edge of the shell ventilation hole (Figure E.4). It is expected that helmets with a stiff shell construction and complex helmet geometry, as for the Bell Bellino, would have further increased levels beyond those observed for the microshell type helmet.

These observations are significant in that they illustrate the need for helmet projections to be weak so that they may break away during impact. Although, soft shell helmets may be prone to greater levels of penetration thereby increasing mechanical interaction, stiff helmets may have the highest injury potential where shell features allow direct mechanical interaction. For this reason, the helmet geometry should ideally be smooth and plain to reduce the incidence of interlocking with irregular impact surfaces.

Despite these observations, the values of tangential force recorded were generally similar to those for the oblique abrasion tests, as summarised in Table 4.15. With the exception of two helmet impacts, the forces generated equate to an injury potential of AIS 0-1. It can therefore be generally concluded that projections on bicycle helmets are unlikely to induce rotational motion sufficient to cause serious brain injury. However, this was not true for two of the helmets tested and this suggests that a test in the standard would be useful and would certainly need to be considered if bicycle helmet wearing were to become compulsory.

Table 4.15 Average peak tangential force for projection and abrasion tests

Helmet	Average peak tangential force [N]	
	Projection tests	Abrasion tests
Bell Bellino	1407	831
Bell Faction	1851	1716
Giro Rodeo	1823	1592
Met Buddy	1338	1378
Met Parachute	995	1260
Met Stradivarius	545	1033
Met Super Loopy	1162	1265
Specialised Airforce	1037	2538

4.3.3.3 <u>Rotational acceleration</u>

The tests show that some helmet features generate forces in excess of 2500N. This is above what is specified by Reg 22.05 and may have the potential to cause brain injury. However, tangential force is an indirect measure and rotational acceleration measured from a nine accelerometer array in the head; a more direct way of assessing brain injury potential, was also considered for assessment.

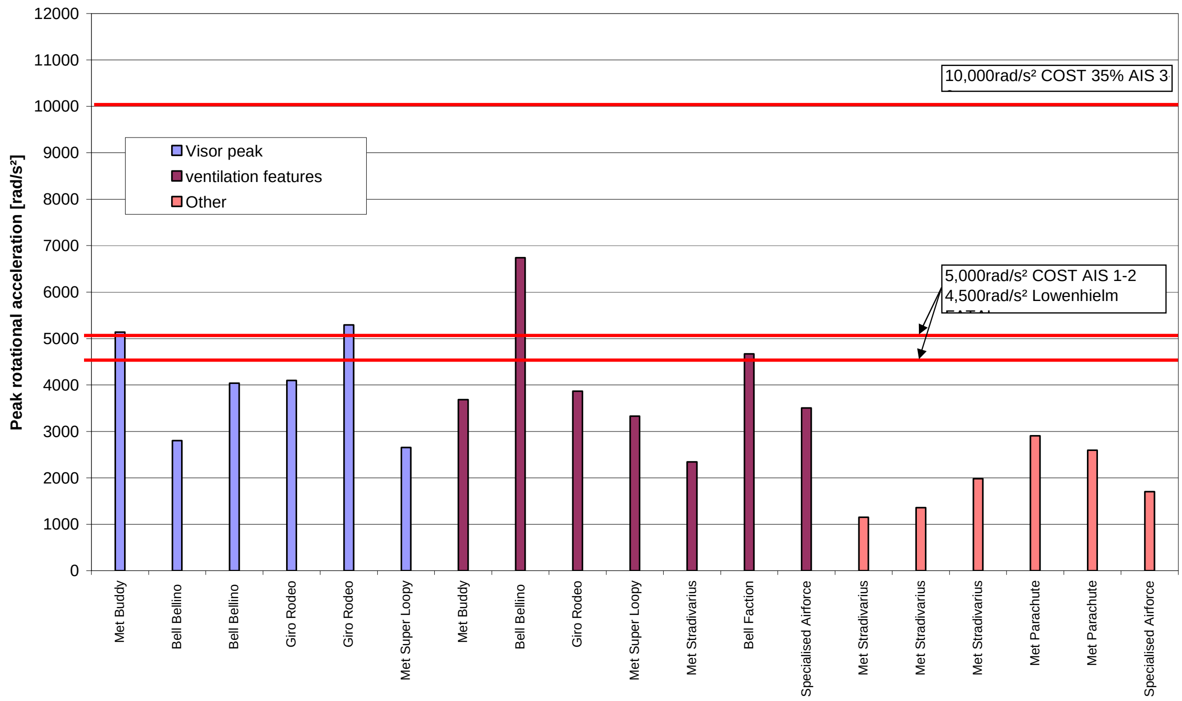

Figure 4.15. Peak rotational acceleration during projection tests

The results show that the maximum rotational acceleration of 6738rad/s² was achieved during a ventilation feature test (Bell Bellino, longitudinal profile) and that the lowest values were recorded for ‘other’ types of projections. This is consistent with the tangential force observations made for the same tests. A slightly different trend was observed for impacts onto the visor features which were found to generate values of rotational acceleration very similar to those for the ventilation features: visors - 4005rad/s² average and ventilation features - 4020rad/s² average, a difference of only 15rad/s². This differs from the tangential force measurements where visor tests were approximately 42% lower than for ventilation features. The correlation between rotational acceleration and tangential force was consequently poor ($r^2=0.48$).

The difference between the relative levels of rotational acceleration and tangential force for the visor and ventilation features is best explained by the different axis through which the headform rotates during these tests. During visor peak impacts the helmet was impacted so that the head was rotated about an axis, referred to as Y, which passes approximately through the ear positions resulting in the forehead moving towards the chin or vice versa. For most ventilation features, the helmet was struck so that the headform would typically rotate around a longitudinal axis, known as X, so that the crown would rotate towards one ear. The moment of inertia of the head varies between these axis and consequently the resultant rotation would also vary for similar impact forces. Although this may not fully explain the 42% variation between the tangential force levels observed at differing sites, it should be remembered that the variation in normal forces between the impact sites will also have a significant contribution to this difference.

Importantly, these results using a size 54cm (E) headform, are no more severe than those recorded during similar oblique impacts onto an abrasive anvil. For most cases lower levels of rotational acceleration were observed for the projection tests. In fact, the peak rotational acceleration recorded was below 8,000rad/s² at which AIS 2 injuries may be expected. This suggests that projections on bicycle helmets are unlikely to induce rotational motion sufficient to cause serious brain injury.

<u>No helmet</u>

Based on the above projection test results, it has been generally concluded that projections on bicycle helmets are unlikely to induce rotational motion sufficient to cause brain injury. Peak rotational accelerations were below the 8,000rad/s² threshold at which AIS 2 injuries would be predicted. However, in this series of tests, two projections produced tangential force levels above 2500N which equate to an injury potential of AIS 2. One of the helmets (Bell Bellino) generated a peak rotational acceleration of 6738rad/s², the maximum measured in the projection tests. This also equates to an injury potential of approximately AIS 2. The peak linear acceleration generated by the Bell Bellino and Giro Rodeo was the maximum measured at about 90g, which corresponds to AIS 1. The injury potential as a result of rotational acceleration during oblique impacts onto projections is potentially higher than would be expected from the linear acceleration component alone.

Despite this, and given the same considerations as the abrasive anvil tests where the injury outcome for an unhelmeted head is expected to be AIS 2 in similar impact conditions (based on a linear acceleration of 100g to 150g), it seems unlikely that even the worst performing bicycle helmet projections would increase the risk of injury in an oblique impact when compared to no helmet. Nevertheless, the fact that in two projection tests the Reg 22.05 limit was exceeded gives cause for concern and consideration should be given to introducing a test into the bicycle helmet standard to ensure that the projection strength and consequential risk of injury is assessed and minimised.

4.3.3.4 Variable impact speed test results

To investigate the variation in performance for variable impact tests speeds, repeat tests were completed at 6.0m/s (as opposed to 8.5m/s) using 54cm (E) headforms. The results are summarised in Table 4.10.

Generally the tests performed on the abrasive anvil at a lower speed of 6.0m/s generated reduced linear and rotational accelerations as well as lower anvil forces. In all cases the linear acceleration was no more than 68g for which an AIS 1 injury would be expected, and HIC 107, which is well below a value at which a head injury would be predicted. Similarly, the rotational accelerations were on average 760rad/s² lower than those observed at 8.5m/s and below 5000rad/s² at which AIS 2 injuries would be expected. The maximum peak tangential force measurement was 1286N with a 953N average and nearly half the threshold for AIS 2 injuries as identified by COST. Overall, it is seen that, at this speed, which may be more realistic for bicycle accidents, a probable injury outcome when wearing a helmet would be below AIS 2 and possibly AIS 1. This would suggest an injury benefit over non-helmet use for such impacts.

4.3.3.5 Variable headform size

An investigation into the effects on helmet performance and injury outcome for differing test headform mass and size during oblique impacts is summarised in Table 4.9; these results include the tests onto the abrasive anvil and for projections.

Child helmets

An increase in linear acceleration and a reduction in peak tangential force were observed for child style helmets when a 50cm (A) headform was used in place of a 54cm (E) headform. An increase in linear acceleration is expected when headform mass is reduced because the stiffness of the helmet energy absorbing liner will provide greater headform acceleration for the same impact force. A reduction in tangential force is not expected but may be a consequence of reduced headform mass and inertia, or reduced helmet fit which would all tend to reduce the force transferred to the anvil. This reduction in tangential force may indicate a less severe injury as a result of rotational motion but the relationship between tangential force and rotational acceleration for these tests is known to be poor. Although, it is feasible that some features of child style helmets may have contributed to a reduction in tangential force and injury potential for smaller heads, it cannot be verified without measurement of rotational acceleration directly at the centre of the smallest headform size i.e. 50cm (A).

Youth / Adult helmets

A range of adult helmets were tested on 57cm (J) and 54cm (E) headforms using both the abrasive and projection anvils. These helmets did not follow the trend seen for children's helmets. Instead, the linear acceleration was found to decrease for the smaller headform size whilst tangential forces were seen to increase in the majority of cases.

Rotational acceleration was measured on these headforms and, surprisingly, this was generally high for the 57cm (J) headform and then reduced when tested with the 54cm (E) size. Thus, a reduction in rotational acceleration and an increase in tangential force occurred when the headform size was reduced. For example, during one impact onto the abrasive anvil the rotational acceleration reduced from 16,224rad/s^2 (J) to 5,821rad/s^2 (E) whereas the tangential force increased from 1517N (J) to 1773N (E). In another test onto a projection the rotational acceleration reduced from 20,642rad/s^2 (J) to 4,668rad/s^2 (E) but the tangential force reduced from 2365N (J) to 1851N (E). The differences in rotational acceleration were substantial and are significant as they indicate that the injury outcome may be significantly higher for a larger headform. Indeed, the levels of rotational acceleration observed were at levels at which serious, AIS 3+ injuries are anticipated. COST 327 suggests the onset of AIS 3 injuries at around 19,000rad/s^2. A 35% risk of AIS3+ injuries are predicted at 10,000rad/s^2, increasing to 50% at 30,000rad/s^2.

This is significant as previous observations, using the size 54 (E) headform suggest that the potential for injury is AIS 2 or below, and approximates to the injury severity expected for an unhelmeted head due to the linear component of the oblique impact. The increased level of rotational acceleration indicates that more severe injuries may be possible for a helmeted head than an unhelmeted head as head size increases.

Of course, the true response of the unhelmeted head to oblique, glancing blows is unknown and greater understanding of this response is needed to allow an accurate assessment of injury risk based on tolerance to rotational acceleration. Further research, possibly using post-mortem human surrogates, must focus on characterising the human head response to oblique impacts to give a greater understanding of the probability of head injury in such events.

For these youth / adult helmets, the trend for increasing rotational acceleration with increased headform size was consistent and occurred in every configuration, whereas the trend for decreasing tangential force occurred only in three out of four tests. These trends align with previous findings that there is little correlation between tangential force and rotational acceleration for these helmets. Similarly, there was no consistent increase in linear acceleration which would explain the significant increase in rotational acceleration.

The large reduction in rotational acceleration when using a smaller, 54cm (E), headform may in part be attributed to reduced linear acceleration which limits rotation inducing torque on the headform. This would counteract the effect of reduced headform mass and moment of inertia which would tend to increase rotational acceleration for smaller headforms. Although a reduction in linear acceleration would not normally be expected for a lighter head because helmet stiffness would tend to increase linear acceleration with reduced headform mass. However, the smaller headform geometry may explain this trend due to a reduced radius and surface area which would tend to increase pressure on the helmet's energy absorbing padding and in turn reduce the effective stiffness of the helmet. This would also contribute to the reduced anvil forces.

A further contributory factor to reduce rotational accelerations for smaller headforms would be the worsening helmet fit. A poor fitting helmet would encourage slippage and limit the transfer of tangential anvil forces to the headform. Although this appears to conflict with research by TRL (Chinn et al) which shows that the worsening helmet fit increases rotational acceleration, this research relates to motorcycle helmets which have a much a greater mass than bicycle helmets. During an impact, poorly fitted helmets will rotate relative to the head when they collide or fit improves suddenly, the helmet momentum can be transferred to the head thereby increasing rotational acceleration. This slide-hammer effect increases with greater helmet mass thus explaining the apparent discrepancy difference between bicycle and motorcycle helmets.

Although poor fit due to reduced headform size may contribute towards reduced rotational acceleration, it would be inappropriate to suggest that this is always the case based on the limited number of helmets tested. Furthermore, this may be very sensitive to the quality of the helmet restraint system, the impact site and other contributory factors such as linear impact performance. Further testing using the nine-accelerometer array would be recommended to investigate the consequences of helmet fit further.

Final observations

The test work completed demonstrates that for a majority of cases, the levels of rotational acceleration observed using a helmeted headform would be no more injurious than expected for a bare human head. However, in the most severe cases the rotational acceleration may increase with headform size to levels corresponding to an injury severity AIS 3. Based on the AIS 2 skull fracture injury, which is anticipated due to the linear component of a 15º oblique impact at 8.5m/s, a marginal disbenefit may therefore exist for some helmet and user combinations. Nevertheless, the unhelmeted head’s response to rotational acceleration during oblique impact events is not well documented and this marginal difference in performance cannot be concluded with certainty. Further research, focused on characterising the unhelmeted human head’s response to oblique impacts would give a greater understanding of the head injury tolerance. Such research may however be unfeasible on ethical grounds.

Since optimisation of helmets to include a broad range of sizes may contribute to reduced oblique impact performance due to issues relating to helmet fit and linear impact performance, this is an area of the standards that should be reviewed. In particular, the current requirements of EN 1078, which includes helmet retention and stability tests, may be inadequate for current designs of bicycle helmet. Furthermore, the introduction into EN 1078 of a test for tangential force or rotational acceleration during an oblique impact could ensure that helmet designs do not provide an excessive risk of injury as a consequence of rotation. Further consideration must be made as to what test would be appropriate, but it should be noted that many of the helmets tested here met the requirements of the Reg22.05 Method A oblique impact test, a test which is intended for the assessment of motorcycle helmets

5 CONCLUSIONS

1. Examination of the linear acceleration results on the flat and kerbstone anvils was encouraging. Only one helmet exceeded the EN 1078 value of 250g and only three helmets exceeded 200g for both impact types. The probability of fatality (from the literature review) showed that at 250g a severe injury is likely and the probability of fatality may be as high as 10.6%. Below 200g, an injury of AIS 3 or less is likely and the risk of fatality is below 2.1%.
2. Five out of eight helmets achieved 200g or less in these tests. Potentially, a reduction of 50g of the permitted peak acceleration, from 250g to 200g would result in some of the poorest performing helmets failing the requirement. If such a requirement were applied to EN 1078 then an 8% reduction in the risk of fatal injury may be achieved.
3. In the one case (Bell Faction) where the peak acceleration exceeded 250g the helmet was tested using a 57cm (J) headform which is larger than the manufacturer's designated size range (51cm - 56cm) by just 1cm. Nevertheless, the helmet fitted this headform comfortably and could be worn by someone with that head size. The 285g result is a consequence of the inability of the helmet to absorb some or any of the 8.4J of additional impact energy associated with the larger headform size. At 285g the probability of a fatal injury is approximately 55% and a critical injury would be highly likely. Even when testing within the manufacturers size range this helmet gave protection inferior to most of the other helmets.
4. Performance in the linear impact tests was examined as a function of velocity and the linear acceleration was compared with the corresponding injury probability as a function of AIS. Two values for skull fracture for an un-helmeted head were included for comparison. The results showed that over the range of 2.2m/s to 5.0m/s the best performing helmets were between one and three increments of AIS below the predicted values for an un-helmeted head. However, between 2.8m/s and 3.2m/s the worst performing helmets may be an increment of AIS higher (worse) than for the un-helmeted head, thereby raising the severity of, or potential for, injury. Thus, there may be justification for introducing a low speed impact into the Standard (EN 1078) to ensure that improved protection is offered at this low speed.
5. Coefficient of friction, based upon the ratio between the normal and tangential force, is a factor that may affect rotational motion and may be dependent upon the helmet shell materials and geometry. The coefficient of friction (μ) was found to vary between helmet types from $\mu = 0.42$ to $\mu = 0.78$ (average $\mu = 0.57$). However, there were no clear trends to correlate geometry or materials with the range of values or the variation in headform loading.
6. The correlation between anvil tangential force and rotational acceleration for the oblique tests onto the abrasive anvil was poor ($r^2 = 0.29$) across the whole range of tests. However, when near identical impact conditions were assessed (same impact site) the correlation improved ($r^2 = 0.69$).
7. In all of the oblique impact tests using 54cm (E) instrumented headforms the peak rotational acceleration was below 10,000rad/s² at which a 35% probability of AIS 3 - 6 injuries were predicted. This is reassuring as these impacts were of a severity intended for motorcycle helmets. Furthermore, only 6 of 17 impacts resulted in rotational accelerations above 5000rad/s², a level at which AIS 1 - 2 injuries are predicted. However, this accounted for half of the helmet models tested and only four helmets achieved test results consistently below 5000rad/s².
8. Generally, those helmet models which achieved a poor rotational acceleration performance also gave a poor performance during linear impact tests. This indicates that linear impact performance may be a dominant factor influencing the rotational acceleration during oblique impacts. The Bell Faction was the worst performing helmet overall, achieving a peak rotational acceleration of 8509rad/s² during an impact to the rear of the shell (for 54cm (E) headform). This corresponds to an injury potential above AIS 2.

9. During right-side oblique impacts, the Giro Rodeo (microshell construction) gave the highest rotational acceleration (6820 rad/s^2) which was 30% higher than the Met Buddy and 70% higher than the Bell Bellino, also of the same microshell construction. The Bell Faction (stiff outer shell) was lower than the Giro Rodeo on this site. However, the sample size was not sufficient to comment on whether the helmet construction has a direct influence on head loading or whether any helmets of certain material constructions are particularly hazardous.
10. Reasonable correlation was found between linear impact performance and rotational acceleration and, given the lack of knowledge of the response of the human head to oblique impact, the linear acceleration during the oblique test was compared with the skull's fracture tolerance. For un-helmeted cadaver tests, the fracture load corresponded to between 100g to 150g and was correlated to a rotational acceleration of between 7500 rad/s^2 and 12,000 rad/s^2 .This is potentially more injurious than the 3000 rad/s^2 to 7000 rad/s^2 measured in the equivalent test with size 54cm (E) helmeted headforms. It is therefore, unlikely that any of these helmets would increase the risk of injury from rotational motion in oblique impacts onto a flat surface.
11. A similar analysis was attempted for the oblique projection tests in which two projections gave values of tangential force above 2500N, which equates to an injury potential of AIS 2. One of the helmets (Bell Bellino) generated a peak rotational acceleration of 6738rad/s^2, the maximum measured in the projection tests. This also equates to an injury potential of AIS 2. The peak linear acceleration generated by the Bell Bellino and the Giro Rodeo was the maximum measured at about 90g, corresponding to AIS 1. Given the same considerations as for the abrasive anvil tests, it is unlikely that any of these helmets would increase the risk of head injury in an oblique impact, compared to an un-helmeted head. Nevertheless, the fact that in two projection tests the Reg 22.05 limit was exceeded gives cause for concern and consideration should be given to introducing a test into the bicycle helmet standard to ensure that the projection strength and consequential risk of injury is assessed and minimised.
12. When youth / adult helmets were tested on a 57cm (J) headform the rotational acceleration was generally very high and typically somewhat in excess of 10,000 rad/s^2 (maximum over 20,000 rad/s^2). These levels exceed those measured using the 54cm (E) headform and correspond approximately with injury severity AIS 3. This injury severity exceeds that predicted for a bare human head for similar impact conditions. However, the predicted injury severity of an unhelmeted head was based on the linear impact response of post-mortem human surrogates. These observations cannot be concluded with certainty and indicate possible trends. The reduced level of injurious rotational acceleration for smaller headforms was judged to be the result of a combination of factors including; reduced linear acceleration, reduced headform size and increased slippage due to poor helmet fit.
13. Research using motorcycle helmets has shown that, the worse the fit the greater the rotational acceleration. The results of the test work completed here contradict this observation but the lower mass of bicycle helmets may explain this trend. However, it is feasible that manufacturing one helmet model to fit such a wide range of sizes may have a disadvantageous effect on fit and linear impact performance. These, in turn may have a detrimental effect on rotational acceleration induced into the head and therefore injury outcome. Thus, although there was limited evidence to suggest that the levels of rotational acceleration observed using a helmeted headform would be any more injurious than those expected for an un-helmeted head, the introduction into EN 1078 of a test for tangential force in an oblique impact and a revision of the helmet retention test should be considered to ensure optimal protection is provided by bicycle helmets.

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