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MOTORCYCLE AND BICYCLE PROTECTIVE HELMETS REQUIREMENTS RESULTING FROM A POST CRASH STUDY AND EXPERIMENTAL RESEARCH

PREPARED FOR

FEDERAL OFFICE OF ROAD SAFETY

CANBERRA

SCHOOL OF CIVIL ENGINEERING

QUEENSLAND INSTITUTE OF TECHNOLOGY

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Royal Brisbane Children's Hospital

Princess Alexandra Hospital

Mater Hospital

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SUMMARY

The lack of recent crash data for motorcyclist and bicyclist crashes involving head injury and little ongoing research in Australia into protective helmet improvements resulted in the funding of this study by the Federal Government. The overall aim was to provide a better basis for the development of improved motorcycle and bicycle protective helmet standards and to initiate ongoing research.

The emphasis in this project *has* been to research the sufficiency and effectiveness of the shell and liner properties of both motorcycle and bicycle protective helmets so as to identify any deficiencies in design. A reduction of these defective elements would, in many cases, lead to an overall improvement in injury outcome for these road users. The three main activities followed in this project were: a review of previous head protection research, a post crash study of both motorcyclist and bicyclist crashes, and a program of experimental research focussing on the structural properties of helmets. The following sections summarise the work carried out and the results and findings for motorcycle and bicycle helmets.

MOTORCYCLE HELMETS

The introduction of the compulsory wearing of motorcycle helmets in Australia in the 1960s resulted in a substantial decline in serious head injury in motorcyclist crash victims. There is ample evidence in the United States of America of the effectiveness of the motorcycle helmet where, in some States, the compulsory helmet laws have been repealed leading to a dramatic rise in head injury trauma. A simple fall from a motorcycle onto a hard surface can result in serious head injury where there is no protective helmet used. The more complete coverage of the head provided by the full face helmet *has* resulted in substantial savings in facial laceration injury.

It is evident, both from previous post crash studies and the post crash study undertaken in this project, that there is a high level of impact to the lower face and sides of the head in motorcycle crashes - refer Figure 6 in the report - and very little impact to the top area of the head. Impacts to the jaw area (even where full face helmets are used) can result in life threatening base-of-skull injury caused by a transmittal of the impact force through the jaw to the base of the skull.

Experimental results have indicated that for an impact to the jaw area, the facebar of a full face helmet reduces intracranial pressure pulses and rotational acceleration of the head. The more rigid fibreglass helmet facebars are effective in reducing rotational acceleration. The post crash study also provided some evidence of injury reduction where the crash victim was wearing a full face fibreglass helmet and sustained an impact to the facebar of the helmet. Figure 8 in the report illustrates the variation deduced from injury patterns.

There has been considerable recent research interest in the effect of rotational acceleration on head injury. Severe rotational acceleration produced by an impact to the head, directed away from the centre of gravity to the head so that rotation is produced, often causes diffuse brain injury and brain stem damage. There is some evidence in previous research that heavy helmets nay result in increased rotational acceleration in a crash impact.

Work carried out in this project on impacts to the lower facial area indicates that rotational acceleration does increase with mass, particularly where an open face helmet is used. Previous research has demonstrated that the full face helmet tends to reduce rotational acceleration because of partial transmission of the impact through the lower part of the helmet into the torso. Also crash simulation experiments carried out in this study have demonstrated quite marked differences in resultant rotational acceleration for the two main types of motorcycle helmet shell, polymer and fibreglass. The experiments also indicate that rotational accelerations are very high in typical low speed impacts where there is a first impact with the helmet.

Where there is forward momentum and an impact occurs with the pavement, an additional reactive impact pulse is produced by the pavement horizontally because of friction between the helmet and the pavement. Fibreglass helmet shells can have down to one third the sliding resistance of polymer shell helmets during impact with the pavement. Unfortunately, the crash simulation experiments also showed that a fibreglass full face helmet can have 1.5 to 2 times more rotational acceleration during impact than the full face polymer helmet. It is considered that this difference *may* be the result of a 20 percent higher mass for the fibreglass helmet, although further experimentation is needed to verify the differences found.

Rotational acceleration is reduced if the direct impact is reduced. The polystyrene padding of liners in motorcycle helmets available in Australia are virtually all of the same density, i.e. 50 kg/m^{-3}. In the past crash evaluatlon of motorcycle helmets involved in crashes, very little crushing of the liner foam was usually evident, indicating that the liner density could be reduced. The experimentation found that the standard impact attenuation test (AS1698) using a solid magnesium headform produced more severe damage to a helmet than would be the case for a real head in a similar crash impact. What in fact happens in a real crash impact is that the human head deforms elastically on impact. The standard impact attenuation test making use of a solid headform does not consider the effect of human head deformation with the result that all acceleration attenuation occurs in compression of the liner. Since the solid headform is more capable of crushing helmet padding, manufacturers have had to provide relatively stiff foam in the helmet so that it would pass the impact attenuation test.

As significant elastic deformation of the head can result in brain damage it would be preferable to have a softer liner material in the helmet so that less deformation of the head occurred. A softer liner, probably about 30 kg/m^{-3} rather than the current 50 kg/m^{-3}, should also reduce rotational acceleration. Further work using humanoid headforms with human deformation characteristics is needed to verify the above findings.

In summary, the recommendations of this project for motorcycle helmets are that:

(1) The solid headform required by Australian Standard AS1698-1980 be replaced by a humanoid headform. This should lead to a substantial reduction in helmet liner stiffness

(2) The current Australian Standard AS1698 is only concerned with the upper part of the helmet where little impact occurs and should be changed to include the vulnerable facial and side of the head areas.

(3) AS1698 should specify full face helmets with stiff face bars and, if possible, phase out open face helmets

(4) Refinements should be made to the shell of motorcycle helmets to stiffen the shell, improve its sliding properties but also to reduce the shell mass.

Because of the complexity of issues involved, more research is needed. Brain damage produced by impact is still an active research area and the effects of rotational acceleration are only starting to be understood. The transmission of impacts to the intracranial space is complex and the conflicting requirements of helmet materials is perplexing. This project has identified a number of factors which will lead to improvements in motorcycle helmets. While further research is required, it is apparent from the findings of this project that there is scope for motorcycle helmet improvements that will result in a significant reduction in head region injury caused by crash impacts.

BICYCLE HELMETS

A significant finding of the post crash work undertaken was the protective effect of bicycle helmets, particularly where the crash involved another vehicle. Figure 14 in the report illustrates the significant difference in severe head injury between helmeted and unhelmeted bicycle crash victims. In this study, over 50 percent of bicyclist crashes were found to involve children under 12 years of age. There is about a 30 percent over-representation of bicyclists with head region injury for bicyclists not wearing a bicycle helmet, compared with those who were.

Of the unhelmeted cases involving severe head injury, over 40 percent would definitely have had an improved outcome if a substantial bicycle helmet had been worn.

Previous research has indicated that a child's skull is more deformable than an adult's. Further experimentation carried out in this project has indicated there is considerable flexibility in the child skull. These differences are illustrated in Figure 15 of the report. Fracture deformation is considered to be between 1.7 and 5 times greater than the adult skull, depending on the zone struck. Bicycle helmets used by children are the Same as those used by adults and are all tested for impact attenuation using a solid magnesium headform. The considerations of head deformation on impact discussed in the motorcycle section above are even more crucial where children's heads are involved.

As the results in Figure 15 illustrate, the child skull is far from being solid and will deform readily on impact. This fact is well known in the medical field and is largely why a child who has had a rather modest impact to the head is usually admitted to hospital for observation. The substantial elastic deformation of the child head that can occur during impact can result in quite extensive diffuse brain damage.

It is quite apparent that the liner material in children's bicycle helmets is far too stiff and should be reduced to less than 30 kg/m^{-3} rather than the 50 kg/m^{-3} material currently used. It is crucial that further experimentation be carried out in this area and a headform be produced for the experiments with deformation properties similar to a typical child head.

Bicycle helmet crash simulation experiments carried out as part of this project indicated very high rotational accelerations for a fall over the handlebars at 45 km/hr^{-1}. The rotational accelerations were found to be 30 percent higher than those found in similar tests using a full face polymer motorcycle helmet. More work needs to be done in this area as there would seem to be a deficiency in rotational acceleration attenuation and may be the result of insufficient shell stiffness.

The post crash study indicated that a high proportion of impacts were to the lover facial and side of face areas and it is imperative that the temporal area be more fully protected than it is by current bicycle helmet designs.

The main recommendations of this project for bicycle helmets are :

(1) Actively pursue the complete definition of more appropriate requirements for childrens' bicycle helmets including changes in the way in which they are tested.

(2) Use of a more realistic headform in impact attenuation tests.

(3) Extend the bicycle helmet to more effectively protect the vulnerable temporal areas and to stiffen the shell but also to, if possible, reduce the mass of the helmet (particularly for childrens' helmets).

Wile children do have substantial protection from impacts to the head when wearing a bicycle helmet, it is likely that substantial head deformation occurs in a major impact due to the stiffness of the bicycle helmet liner in the Australian Standard bicycle helmet. The extent of this problem should be quickly defined by further research and changes to bicycle helmet standards should be addressed with a view to incorporating requirements for children.

MOTORCYCLE AND BICYCLE PROTECTIVE HELMETS REQUIREMENTS RESULTING FROM A POST CRASH STUDY AND EXPERIMENTAL RESEARCH

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INTRODUCTION

BACKGROUND

The introduction of compulsory motorcycle protective helmet usage in Australia through the 1960s was a first in the world indicating a high level of road safety consciousness on the part of Australian governments. Given the steady rise in motorcycle and bicycle injuries and fatalities, the effectiveness of protective helmets has been closely scrutinised by the House of Representatives Standing Committee on Transport Safety and its predecessors. There was found to be a lack of recent post crash data and ongoing research into protective helmet improvements. This project addresses these issues.

A further issue in need of resolution is the effectiveness of bicycle helmets and an appropriate standard, as governments ponder the possibility of introducing a mandatory helmet usage law for this type of road user.

The Australian Standards for both Motorcycle and Bicycle Helmets are very much adaptations of British and US Standards and it *has* only been in comparatively recent times that certain issues relating to protective helmets have been considered, supported by laboratory investigation. The helmet standard acceptance tests of impact and penetration resistance have been cemented into many helmet acceptance standards since the tests were first developed in the 1950s. This report questions the validity of these tests.

SCOPE OF WORK

The overall objective of the project *has* been to investigate current motorcycle and bicycle protective helmets to identify any deficiencies that, if Overcome, would result in an overall improvement in injury outcome for these road users. Quite obviously there are a range of issues including ventilation, carbon dioxide retention, external and internal projections, *sun* peaks, visibility, conspicuity and helmet retention, all of which may have effects on head injury outcome. While these issues are, where possible given some attention in this report, the main issues that have been investigated are the structure and stiffness of the protective helmet as it is believed that this is where improved understanding will lead to better head protection design.

There have been three main activities in this project: a review of previous head protection research, a post crash study of both motorcycle and bicycle crashes, and a program of experimental research which has focused on important issues identified by the other two main activities. The post crash study concentrated on crashes involving head injury with hospitalisation or death as an outcome. The aim was to try to identify protective helmet factors which may alter the outcome in serious crashes. It is well accepted that protective helmets improve the outcome in minor crashes (Cannell, King, 1982) and there is evidence from the United States, for motorcycle helmets, of quite high reductions in morbidity where helmets are worn . There may be scope to further improve outcome for the more serious crashes by refinements to helmet design, although some researchers question the head injury protective properties of existing motorcycle helmets.

After completion of a thorough review of previous research, and the collection and analysis of an initial sample of motorcycle and bicycle crashes, four areas of experimental work were defined and are as follows:

Head hard tissue stiffness
Protective helmet shell and padding stiffness
Jaw protection
Crash simulation

The aim of the first two areas of experimentation was to gain an improved understanding of the transmission of impact forces during collision involving the head, as the current standard protective helmet acceptance testing does not consider the whole system but only the acceleration attenuation of the helmet. (A metal representation of the head is used and it is like testing the protection with a large metal hammer representing the head rather than a more realistic deformable egg-like headform).

Jaw protection was identified as a crucial area for experimentation because recent research (Harms 1984) has indicated that force transmission via the lower jaw to the base of the skull can be lethal. It was further recoqnised from the post crash study that a significant proportion of head injuries involve both impacts to the lower jaw and fracture of the base of the skull. Development of a means of measuring the force transmission consequences of a blow to the jaw area, either protected or unprotected, was one of the most significant contributions of this project. The other was in the area of crash simulation where a dummy was impacted either with the pavement or a vehicle in a series of simulated crash situations. Crash simulation afforded an opportunity to measure rotational acceleration, again a phenomenon which has attracted considerable recent research interest.

Protective helmets have been thought of as giving protection from a direct blow. The standard acceptance tests for both motorcycle and bicycle helmets reflects this concept of direct impact in that the tests require the helmet to be simply dropped onto a hard surface and to be able to prevent a sharp pointed object, when dropped from a considerable height, from penetrating to the head.

These tests do not reflect the actual crash situation which usually involves considerable horizontal acceleration of the protected head as well as vertical acceleration, resulting in high levels of rotational acceleration of the head on impact. Viano (1985) found that substantial rotational acceleration causes vein rupture within the intracranial space and brain stem injury both with highly morbid consequences. The crash simulations, while hard to accomplish, have enabled the quantification, at least in relative terms, of rotational acceleration for various types of helmet.

ORGANISATION OF THE REPORT

The report has been structured into two parts - a main report and a more detailed report. The main report outlines the work done, presents the findings from the three main areas of work (review of previous work, post crash study and experimental research) and discusses the various findings identifying the scope for improvements to protective helmet standards.

STUDY APPROACH

LITERATURE REVIEW

The review was carried out in two main areas: Previous research related to post crash investigation, and previous research related to helmet testing and experimentation. Several quite extensive post crash surveys have been carried out previously, as well as a number of prospective studies based on hospital records and coroner's reports. Head injury and its longer term effects is an area of medical science which is still not well understood. The review also covered areas involving the assessment and severity ranking of head injury.

In the area of experimentation the review of previous work covered a wide range of areas, including these where detailed experimentation was carried out as part of the project. A small number of crash simulation studies have been carried out previously, and these are reviewed.

One of the most complex problems of human biomechanics is the impact response of the head and neck system. Different researchers have tackled the problem in various ways but in all cases with only partial success. Mathematical models, physical head/neck models and cadaver experiments are reviewed. A number of studies have been carried out involving various forms of helmet testing, including some quite good recent work from the United Kingdom relating to protective helmet liner stiffness and glancing impacts to protective helmets. A literature survey was also required to assist experimentation in the areas of human tissue stiffness properties and cadaver preparation.

A full description of both reviews can be found in the detailed report, and the main findings are discussed and reviewed in the following two chapters.

POST CRASH SURVEY

The survey inputs were obtained from a hospital review of Crash victims and advice from the police for very serious and fatal crashes. A medical research fellow from the Royal Brisbane Hospital reviewed admissions into neurosurgical, facial and spinal units at Royal Brisbane, Princess Alexandra and Mater Hospitals. The survey team was in close contact with the Police Traffic Accident Investigation Squad (TAIS) and, where possible, attended crashes soon after the crash while TAIS were in attendance. TAIS is a specialist police group which becomes involved when there has been a serious life threatening accident. The survey team also kept in close contact with the Gold Coast and Townsville police TAIS units.

The Police also provided information on fatal crashes outside Brisbane and these have been included in the sample. In these cases, and for Brisbane fatal crashes, the post mortem report was obtained. For most of the Brisbane fatal crashes the post mortem was attended.

In summary then, the survey procedure had three variations as follows:

1. Hospital admissions
   - Review victim's medical sequelae. If at a later date the victim dies, attend the post mortem and obtain the post mortem report.
   - Follow up the crash with the Police to obtain a description of the crash. (For many of the bicyclist cases this was not possible because the Police were not called).
   - Obtain the helmet and record properties and damage.

2. TAIS attended crashes
   - Visit the crash site, where possible while TAIS are in attendance and obtain the details of the crash.
   - Follow up the victim at hospital and obtain the medical sequelae. Alternatively, in the case of a fatality, attend the post mortem and obtain the post mortem report.
   - Obtain the helmet and record helmet properties and damage.

3. Fatal crashes outside Brisbane
   - Advice from Police of a fatal crash.
   - Follow up with Police to obtain a description of the crash and the helmet.
   - Obtain the post mortem report.
   - Assess helmet properties and damage.

The survey data collection procedures were modelled on the Transport Road Research Laboratory in the United Kingdom motorcycle crash survey procedure. In the survey three survey forms were used, viz:

- Accident Site *form*
- Medical Report form
- Helmet Assessment form

Copies of these survey forms can be found in Appendix A. The survey data was computerised into a database with the fields as set out in Appendix A.

SAMPLE COLLECTED

The survey was conducted over a 12 month period *from* June 1985 to July 1986 and data on a total of 329 crashes were collected and processed. Of that total there were 171 bicycle crashes and 158 motorcycle crashes. A total of about 70 crashes were identified via Brisbane TAIS and about 80 percent of these were attended at the time TAIS were in attendance.

As the minimum level of head injury in the crashes surveyed required hospitalisation, the sample of crashes is biased towards severe crashes. Of the bicyclist cases surveyed, 8.8% were fatalities compared with 3.3% fatalities for all of Queensland in 1985 (comparing fatal bicyclist crashes with bicyclist casualty crashes). Similarly, for motorcyclist cases surveyed, 45% were fatalities compared with fatalities of 5.4% for all of Queensland in 1985.

The emphasis in the post crash survey was to collect detailed information about each crash. Even with detailed information it is often difficult to determine events in a bicyclist or motorcyclist crash. Given the range of crash situations, objects and vehicle types involved in the crash, the variation in impact speed, variation in directions of victims before, during and after impact and their level of protection and the ability to avoid or minimise injury, it is very difficult to make comparisons between crashes.

For instance, just prior to impact a motorcyclist may try to lay his motorcycle down and in so doing may only have a glancing blow with the road surface, and any vehicles present, resulting in a relatively good outcome. Again, in another crash the motorcyclist may attempt the same manoeuvre and end up sliding on his aide into the underneath area of the vehicle with a very severe outcome. Often motorcyclists who crash directly into a vehicle and fly over it seem to have "better" outcomes. They may still sustain serious extremity injuries but survive with a relatively good long term outcome. Where a motorcyclist has a direct impact with a vehicle which is at a large angle, and into an area of the vehicle which will tend to catch the motorcyclist's head - especially possible with commercial vehicles - truck trays, bullbars and other structures, then the outcome is usually severe.

While in some cases a motorcyclist or bicyclist seems to have a charmed life, the rider is killed in other crashes which are relatively low speed and apparently survivable. Part of the explanation may lie in variations in the physical characteristics of riders. For instance, the severity of intracranial injury may be dependent on the sharpness of bony ridges and variation in strength and condition of veins in the intracranial space. Riders accustomed to falls seem to be more capable of handling the crash situation. For instance, some have avoided gross injury by standing up on impact with the side of a vehicle and flying over it, or by bunching up and tumble turning and protecting their head.

The above discussion is simply an illustration of but some of the many possible variations indicating that each crash should be exhaustively investigated. It is recommended that each crash be extensively analysed and reviewed in future bicyclist and motorcyclist post crash investigations. It may, in fact, take considerable forensic investigation, very detailed examination of the victim and extensive interviews with witnesses to reasonably define an individual crash. This is the approach now being taken by the Transport Road Research Laboratory in the United Kingdom where exhaustive studies of a very limited number of selected motorcyclist crashes is now being undertaken.

An issue that at times confounds a post crash survey of helmet use is the non availability of the helmet. In some crashes the helmet is destroyed (usually burnt in a fire resulting from the crash). Helmets are sometimes lost at the site, buried with the victim, kept by the family or friends as a momento or withheld because of suspicion or mistrust. Usually about 20 percent of helmets are unavailable for one of these reasons.

The post crash survey has collected together quite detailed data about a number of crashes, as indicated by Appendix B. The data file is about 1 Megabyte and very detailed information is available for some crash situations. It is recommended that further postcrash data collected be targetted at specific crash types in the human tolerance range, and that the crashes be completely reconstructed and possibly simulated to fully define the impact forces and energies involved.

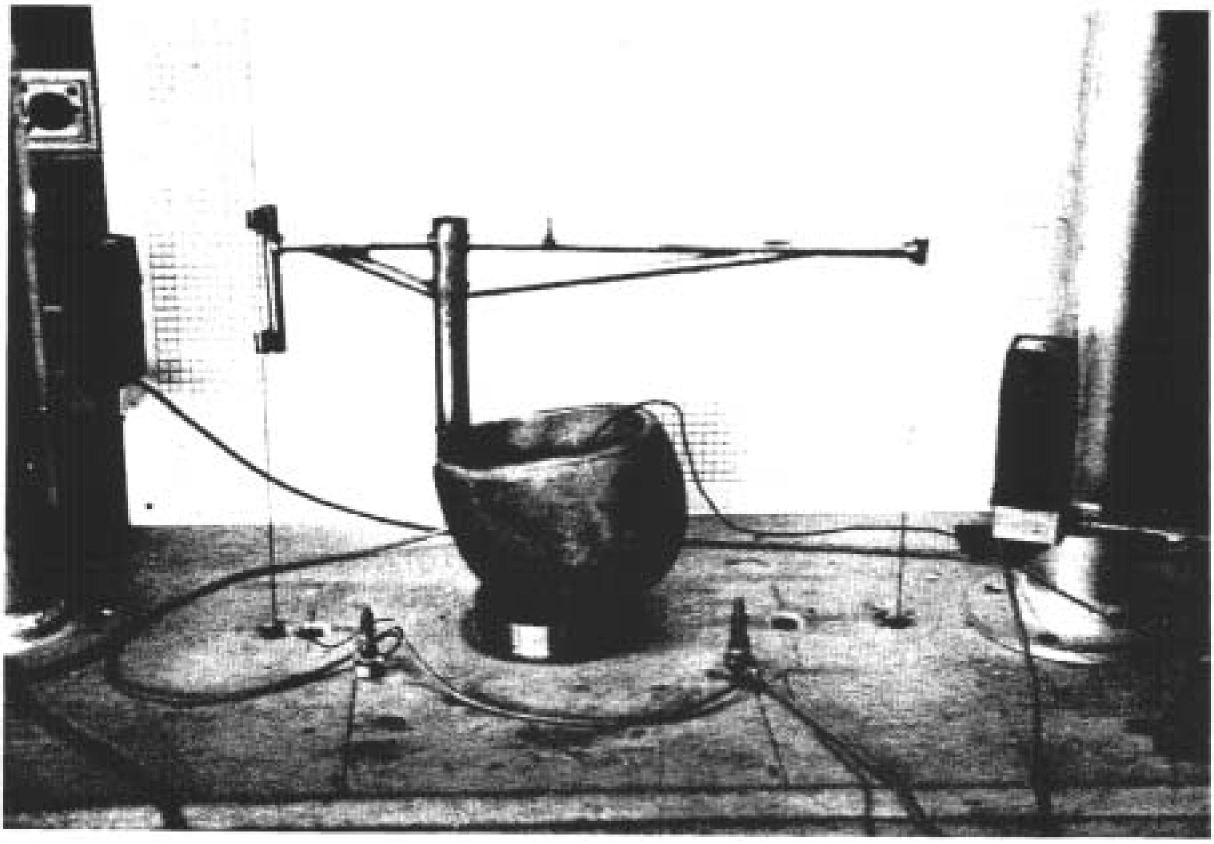
In the detailed report, a full report of both the motorcycle and bicycle post crash studies is presented and the findings of these studies are discussed and reviewed in the following two Chapters.

EXPERIMENTAL WORK

Four areas of experimentation were defined after completion of the review of previous work, and a sample of post crash results were analysed. The four areas involved were: head hard tissue stiffness, protective helmet shell and padding stiffness, jaw protection and crash simulation.

Measurements of samples of head hard tissue helped to develop a better understanding of bending strength of the head and the relationship to it of age in children. Mohan, Bowman, Snyder, Foust (1979) indicates that the strength of a child's hard tissue is lover than an adult's with very little difference by the age of 15 years. Mohan et al. postulated the difference based on what is known of body hard tissue differences with age of the child. A great deal of previous research has been carried out on the strength properties of head hard tissue but not bending strength, so a series of bending tests were carried out on samples of head hard tissue.

Protective helmet shell and liner stiffness is a complex issue which is currently largely being defined for helmet manufacturers by two rather artificial tests in the Australian Standard; the impact attenuation test and the penetration test. As discussed earlier in the Introduction Chapter, both tests fail to test protective helmets in realistic crash situations. One obvious shortcoming is the representation of the human head used for acceptance testing. As the following photograph indicates the head is represented as a solid piece of magnesium metal with only the shape of it having any resemblance to a human head.



PHOTOGRAPH 1
THE AUSTRALIAN STANDARD MAGNESIUM HEADFORM

A better representation of the human head was obtained from the Wayne State University. The HodgsonlWSU headform, was developed from load tests carried out on cadavers and is considered to have reasonable biofidelity.

The experimental *work* involved comparisons of the two headforrs, The Australian Standard Magnesium headform and a HodgsonlWSU headform in the Australian Standard vertical acceleration test rig as illustrated in Figure 1.

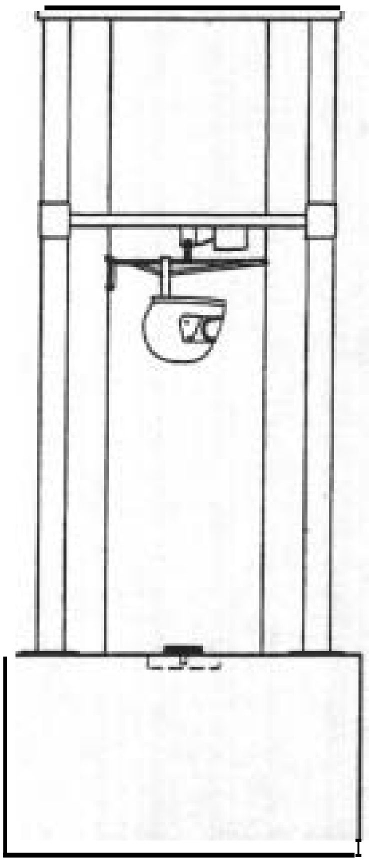


FIGURE 1 THE AUSTRALIAN STANDARD VERTICAL ACCELERATION TEST RIG

Comparative tests were carried' out on various protective helmet types using various densities of foam. Given that virtually all protective helmets complying with Australian Standards have the same padding density, tests using alternative foam densities provided an understanding of the implications of varying density and thickness. A number of previous researchers, Gale, Mills (1984) and Hearn, Sarrailhe (1978) have indicated that protective helmet paddings are too stiff.

"he jaw protection experimentation involved working up research protocols *for* non-injurious jaw impact on a volunteer and the non-destructive testing *of* cadaver material. Ethics Committee approval was obtained for both *of* these procedures. In both cases a variety of impacts were delivered using a impactor to the lower jaw area for no helmet, no face protection and two types of full-face helmet facebar. As well as monitoring the effect of the facebar, it was also possible to gain some appreciation of the effect of helmet mass as it has been considered by some researchers (Krantz 1985) that helmet mass may influence the amount of rotational acceleration during an impact. The following illustration shows how the jaw impact rig developed for the jaw impact research is operated. For the tests using cadaver material it was possible to also measure the intracranial pressure pulse during impact, which yielded important information concerning the transmission of impact pulses via the base of the skull to the brain.

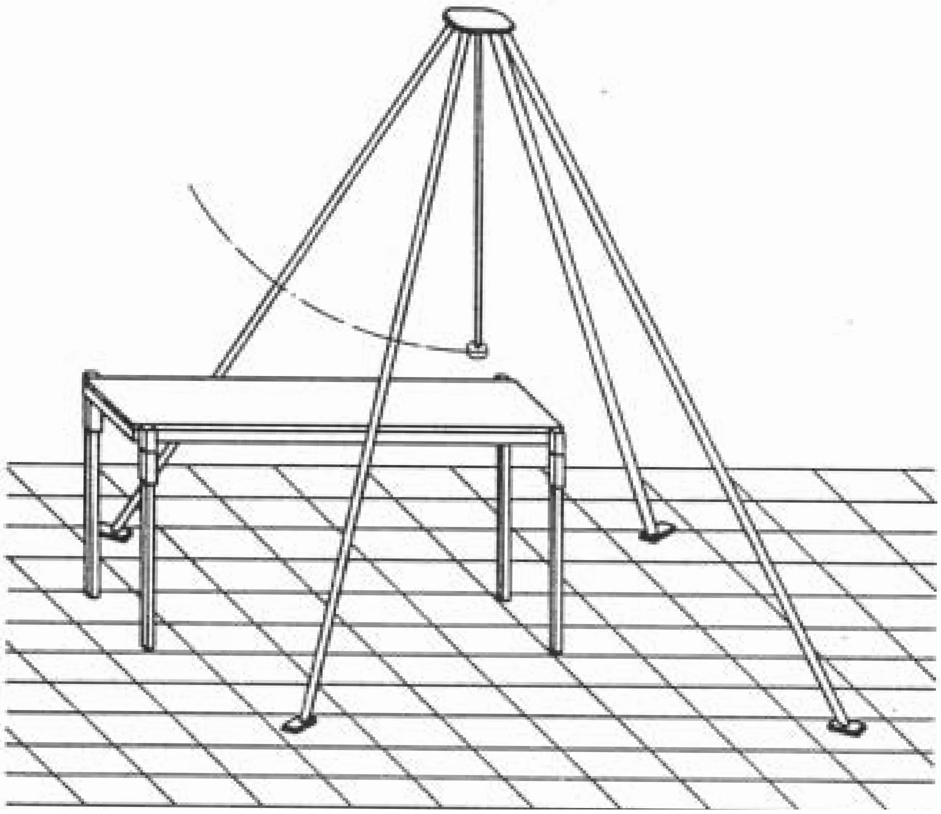


FIGURE 2 THE *JAW* IMPACT RIG

The crash simulation experimentation involved the acceleration of a dummy on a cycle sled to speeds of up to 45 km/hr-1 prior to ejection of the dummy either into a motor vehicle or over the handlebars and head first into the laboratory floor. The following photograph shows the dummy ready for launching.

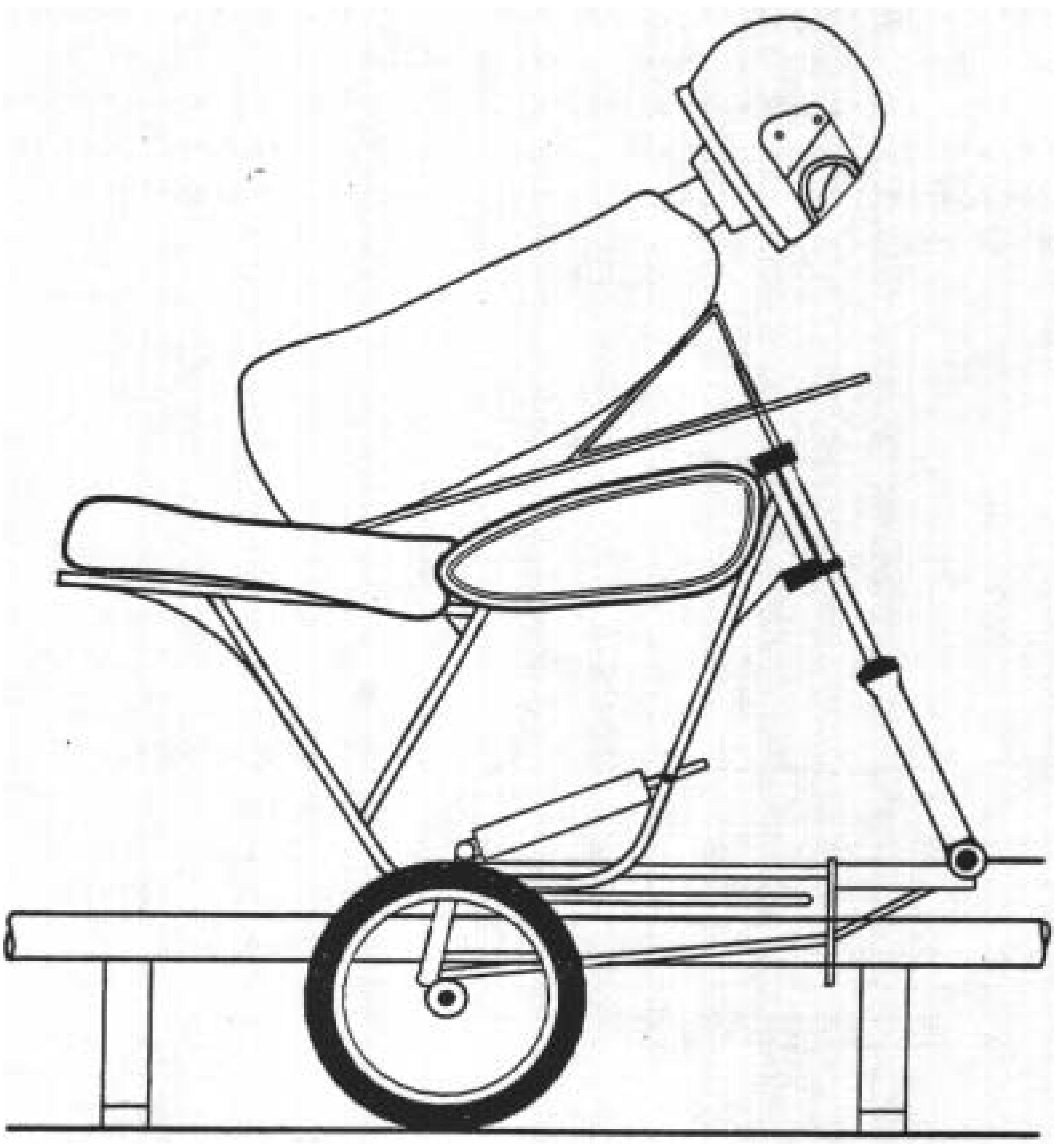


FIGURE 3 *CRASH* SIMULATION RIG

For some of the tests, high speed photography was used and for the remainder of the tests accelerometers were fitted inside the head of the dummy so that rotational acceleration could be measured.

A full account of the experimental procedures can be found in the Detailed Report section. The main findings of the experimentation are discussed and reviewed in the two Chapters following.

DEFINITION OF MOTORCYCLE HELMET REQUIREMENTS

A considerable number of issues arise from the three areas of work in this project: rwiew of previous research work, the post crash study and the experimental research. The purpose of this Chapter is to examine all of the various issues identified in the project, and to discuss the need for changes to design and the Australian Standard for Motorcycle Helmets AS1698. The following sections summarise the main project findings and the final section considers the need for changes to the Australian Standard.

PREVIOUS STUDIES

It is apparent that current motorcycle helmet standards are reducing serious trauma. There is ample evidence of this available in the United States where in some States the removal of compulsory helmet wearing laws have seen a dramatic jump in .serious trauma (2 to 3 times) resulting from motorcycle crashes.

Previous studies have indicated there is a high incidence of impact to the facial area with a tendency for more severe injuries to be at the base of the skull. This may be the result, in some cases, of impacts to the jaw area being transmitted to the base of the skull. There is also a high level of facebar failure in full face helmets involved in frontal impact. A mathematical model has demonstrated high tearing stresses in the brain stem for a blow delivered to the occipital region, which is evidence of the potentially lethal effect of rotational acceleration resulting from non-centre of gravity directed impacts to the head.

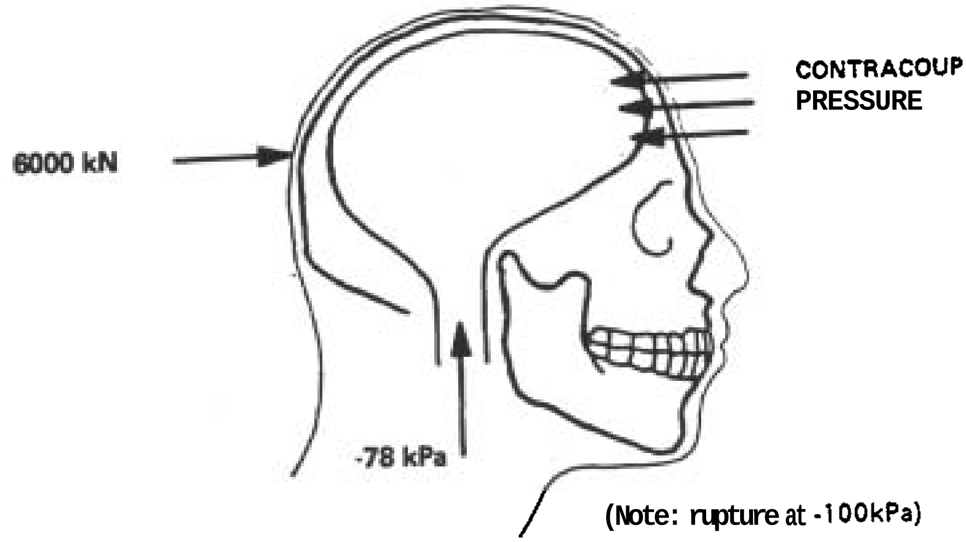


FIGURE 4 MATHEMATICAL MODEL RESULTS
(Source: Hosey)

It is possible that higher mass helmets may aggravate rotational acceleration althouqh it would seem that full face helmets, althouqh heavier, may develop shock transmission paths to the torso.

There are indications from injury patterns of a need for better distribution of energy by the shell and the point is made that fibreglass shells are less deformable as well as having lower frictional resistance to sliding. The density of helmet liners may be too high, with one researcher recommending a density of 30 kgm^{-3} instead of 50 $kg\ m^{-3}$ and another indicating that the current standard impact test may not be effectively testing the shell and padding. There are high levels of helmet retention failure but usually associated with very severe high energy crashes.

POST CRASH STUDY

A total of 158 motorcycle crashes involving head injury were surveyed, with most of the cases having serious head injury. There were 45% fatalities in the surveyed cases compared with 5.4% of fatalities in the Queensland population of motorcyclist hospitalised injuries in 1985.

From the survey reliable data was developed for 18 open face helmets and 78 full face helmets. The low level of open face helmets possibly reflects user preference for the full face helmet. The two main shell materials are fibreglass and various polymer blends, with fibreglass making up 70% of the helmets surveyed. Again, this possibly reflects a user preference for fibreglass helmets.

Impacts with fixed objects and motor vehicles often produce severe head injury, and over an impact speed 80 km hr^{-1} there is a high probability of fatality unless the rider manages to only have glancing blows to the head. There seems to be little argument in favour of attempting to design motorcycle helmets for high speed crash situations because the level of injury in the largely unprotected body (other than the head) is usually very severe.

As the following figure illustrates, the level of severe injury increases as the impact speed increases.

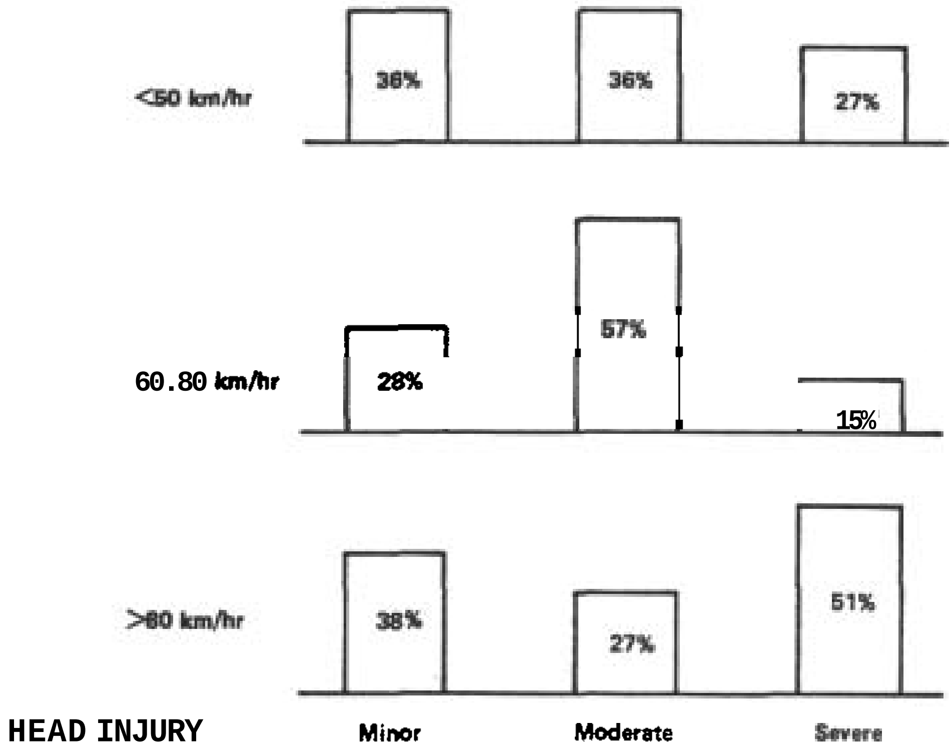


FIGURE 5 MOTORCYCLIST HEAD INJURY SEVERITY VERSUS IMPACT SPEED

The following figure shows the distribution of major injuries and the frequency observed. While the survey is not representative of typical injury patterns, as the survey has a concentration of severe cases, the figure does illustrate where the severe life threatening injuries are occurring.

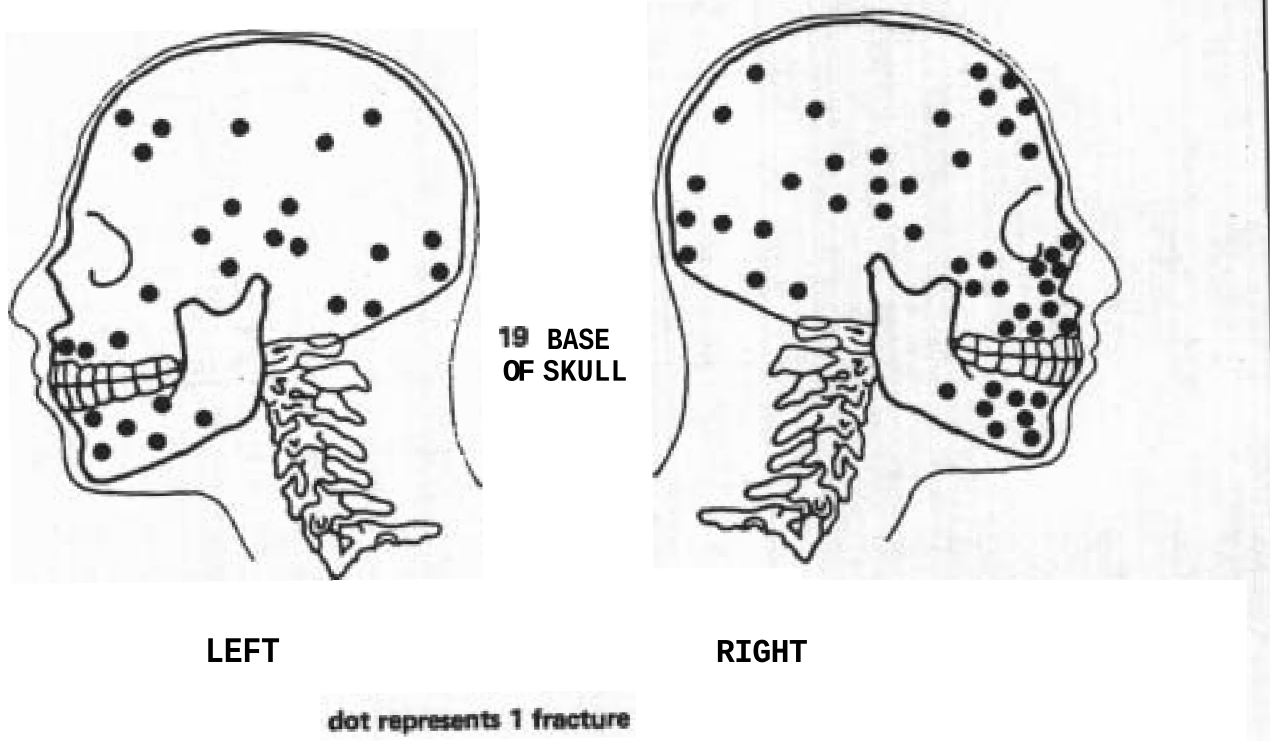


FIGURE 6 MOTORCYCLIST SERIOUS HEAD INJURY LOCATIONS

Motorcycle helmets examined in the project are all carefully designed to meet current impact and penetration attenuation standards. The thickness of the shell varies from about 2mm to 6mm with not all areas necessarily having the same thickness. The thickest shell zone, as would be expected, is within the Australian Standard Test zone which is the crown area of the helmet (refer Figure 6). Liner or padding again varies in thickness but is typically about 30mm in the test zone and there is virtually no difference in foam density between brands. Virtually all helmets examined had a foam density of approximately 50 kg m^{-3}.

Helmet mass varies from 0.7 kgs up to 1.7 kgs. Fibreglass helmets may have a slightly better outcome compared with polymer helmets including impacts to the facebar area. There is also a little evidence that full face helmets may provide some protection from neck injury compared with open face helmets. Even in very severe crashes there is virtually no occurrence of two impacts at the one location or very sharp penetrating surfaces. The worst penetrating surface observed was the edge of a light steel channel about 3mm in width.

There is some evidence found also by other researchers that full face fibreglass helmets have improved head injury outcome. The following figure illustrates the difference found in the post crash study.

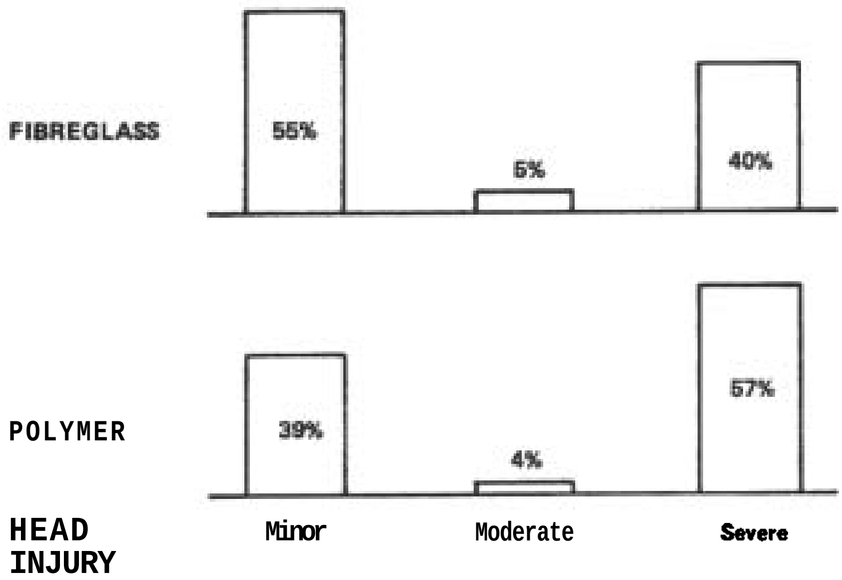


FIGURE 7 MOTORCYCLIST HEAD INJURY SEVERITY VERSUS HELMET SHELL

The stiffer fibreglass facebar of the full face helmet also seems to afford improved protection, as the following figure illustrates, but is based on quite a small subset of post crash data.

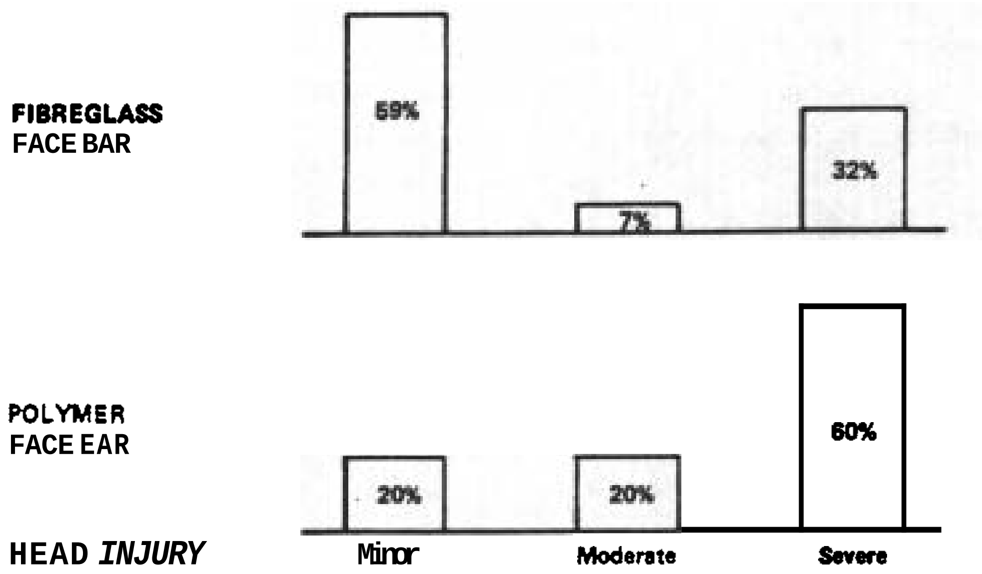


FIGURE 8 MOTORCYCLIST HEAD INJURY SEVERITY VERSUS FACE BAR TYPE

EXPERIMENTAL RESEARCH

The temporal area is a zone of weakness in the skull hard tissue, and given the significant number of impacts that occur in this area, the shell should be stiffer and the liner softer in this zone. Bone tests indicate that the temporal region has only a half to a third the strength of other areas of the skull.

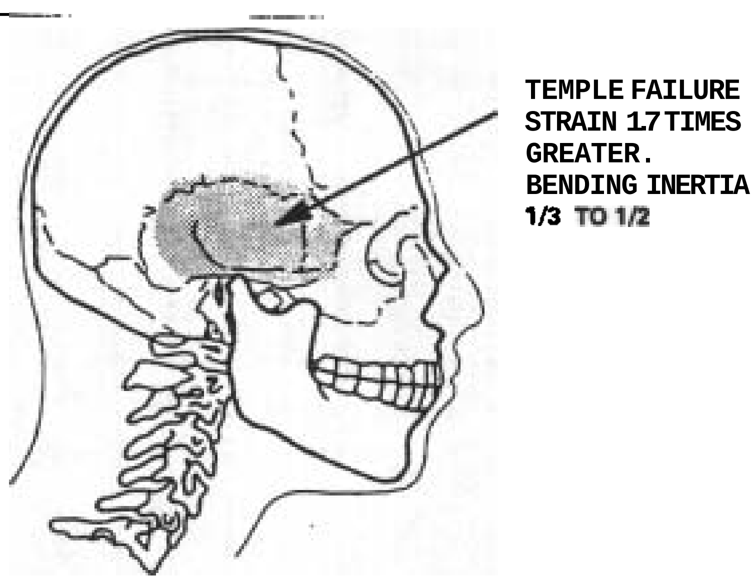


FIGURE 9 ADULT HARD TISSUE – TEMPORAL REGION

While further testing would be desirable, it would seem that the current Australian Standard impact attenuation test is quite inappropriate. Because of the unyielding characteristic of the magnesium headform used in the drop test, the outer hard shell of the helmet and the liner is more easily distorted and compressed (illustrated in Figure 10).

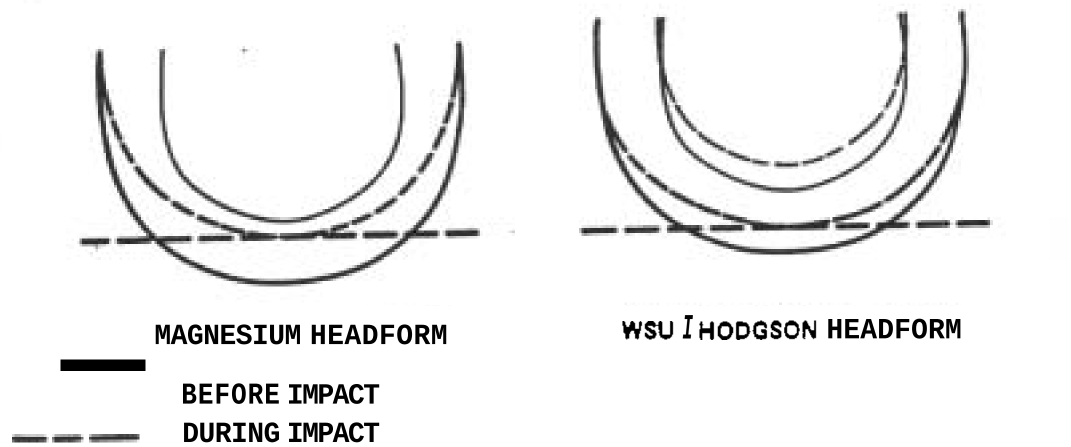
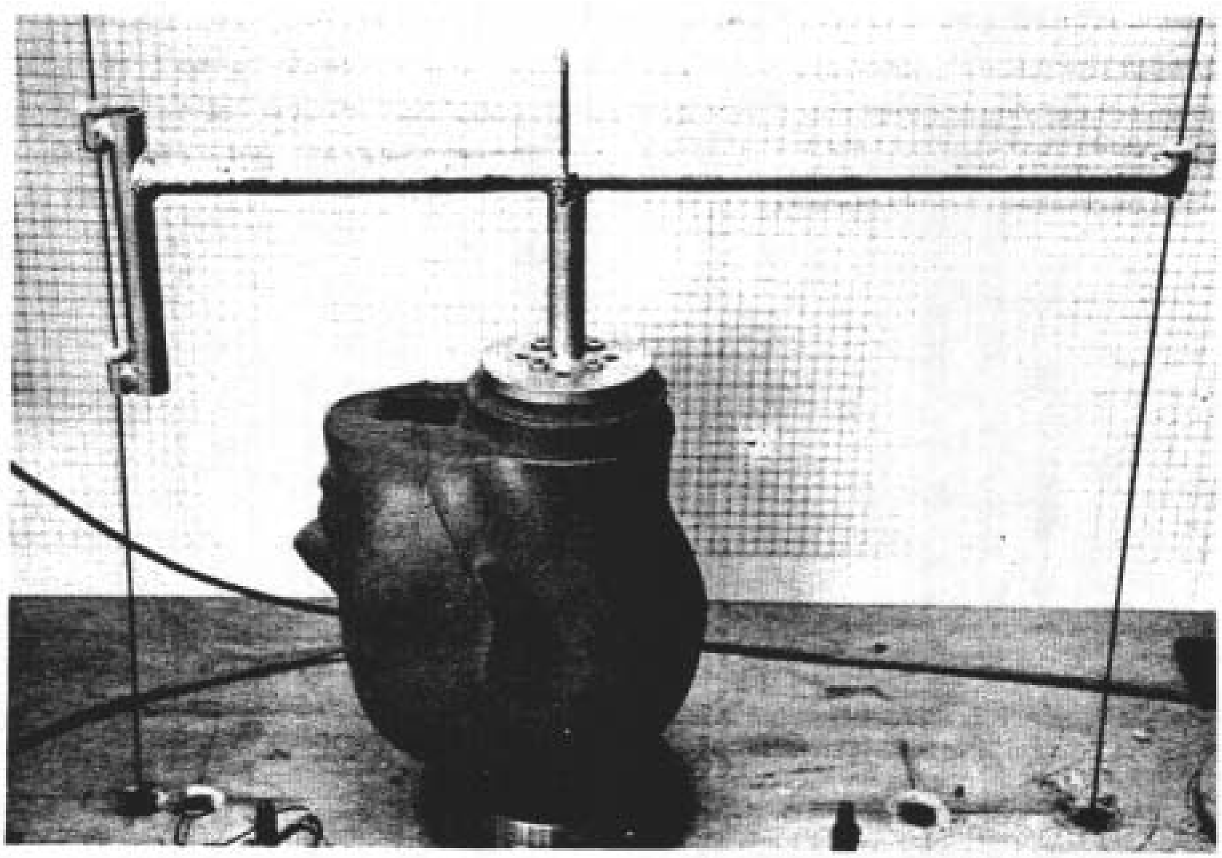


FIGURE 10 ILLUSTRATION OF THE HELMET IMPACT USING THE MAGNESIUM HEADFORM COMPARED WITH THE WSU/HODGSON HEADFORM

If a human like headform is used, and the Wayne State University Hodgson headform seems to be reasonable, flexing of the headform can occur. When this headform is used in the standard impact attenuation test, the helmet designed to the current Australian Standard is less damaged because it does not produce the same level of distortion as the magnesium headform produces, but similar impact accelerations can occur because of flexing of the WSU/Hodgson headform. Inbending must be strictly limited especially for children's helmets because of the adverse effect it can have intracranially in an actual crash impact.



PHOTOGRAPH 2 WAYNE STATE UNIVERSITY HODGSON HEADFORM

The indications from experimental work are that the current helmet liners are too stiff and a liner foam density of about 30 kg m^{-3} should be used rather than 50 kg m^{-3}.

The jaw impact experiments have indicated increased intracranial pressure pulses and increased rotational acceleration with increased helmet mass and, given the earlier discussion, a possible increase in brain stem tearing stresses.

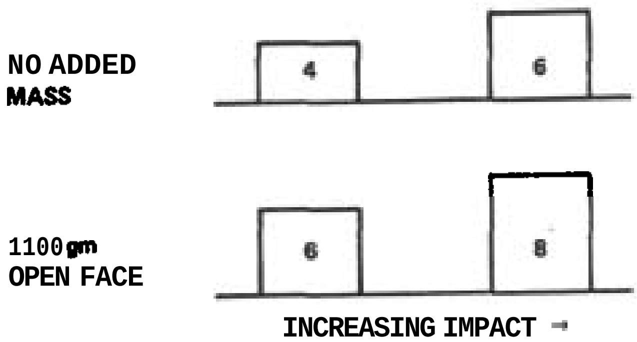


FIGURE 11 IMPACT INTRACRANIAL PRESSURE PULSE VERSUS INCREASED MASS AND IMPACT FORCE

For the full face helmet the distribution of impact shock away from the jaw results in some reduction in intracranial pressure measured in a direct line with the impact. The facebar in a full face helmet produces a marked reduction in rotational acceleration. The open face helmet has disadvantages having both high intracranial pressure pulses and high levels of rotational acceleration when facial impact occurs.

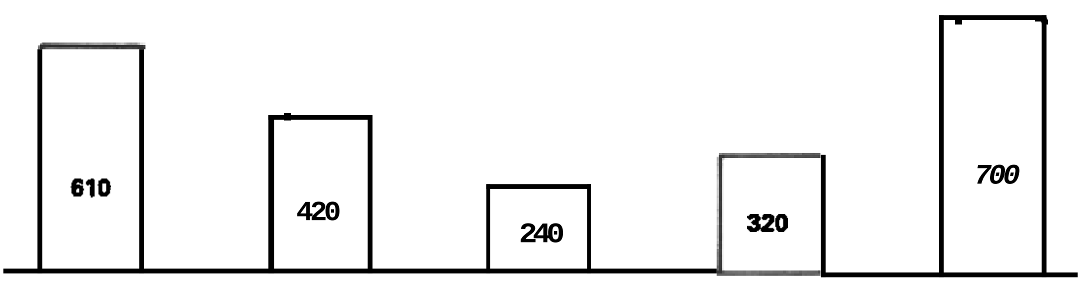


FIGURE 12 ROTATIONAL ACCELERATION PRODUCED BY IMPACTS TO THE *JAW* FOR VARIOUS TYPES OF HEAD PROTECTION

Crash simulation experiments have yielded conflicting results. There is an indication that fibreglass helmet shells have a much lower impact frictional sliding resistance but can have a considerably greater rotational acceleration. It would seem that sliding resistance of a fibreglass shell during impact can be almost one third that of polymer material. The fibreglass helmet can have 1.5 to 2 times the rotational acceleration and this difference was found in both fall impacts with a surface and impacts with a motor vehicle. But equally, one brand tested with a 1.7 kg mass had a rotational acceleration similar to that of the polymer helmet (1.1 kg). This may be the result of high quality shell sliding properties of this brand.

It is possible, but not proven, that the increased mass may be at least partly producing the difference. The fibreglass helmets used for crash simulation had about a 20% higher mass. There are also very high levels of rebound associated with the simulated impacts.

The rotational accelerations measured in the crash simulation experiments are enormous: 40,000 to 60,000 rad s^{-2} when compared with 4,500 rad s^{-2} for onset of vein rupture. Rotational acceleration is related to the magnitude of the impact forces and is reduced if the impact force is reduced by using better energy absorbing material. Rotational acceleration can also be reduced in a sliding crash by improving the impact frictional properties of the shell material.

REVIEW OF THE CURRENT AUSTRALIAN STANDARD FOR MOTORCYCLE HELMETS

Current motorcycle helmets do have a protective effect. It has been found in the US that where compulsory helmet wearing laws have been repealed, serious head injury increases two or three fold.

This review mostly deals with the fine tuning of a proven protective device. Nonetheless, the Australian Standard for motorcycle protective helmets would seem to be largely irrelevant in its current form and may be producing helmet designs which are too stiff because the required tests are artificial.

For crash situations the standard only addresses a simple fall on the head. It does not recognise real crash situations but only requires impact attenuation protection for the upper part of the head above a testline which leaves most of the vulnerable temporal region of the head unprotected by the Standard. Most crashes involve the facial area of the helmet, yet there is no requirement for impact attenuation in the lower facial and side of the head area. "he effect of considerable forward velocity when vertical impact occurs is not considered yet the extent of impact sliding resistance can have a marked effect on rotational acceleration.

The full face motorcycle helmet appears to be more capable of distributing impacts. Open face helmets do not do this and the effect of the added mass on the head produces high rotational acceleration and intracranial pressure pulses for frontal impacts. "he standard uses an inappropriate headform to test impact attenuation and in so doing produces helmet designs with liners which are too stiff. The requirement for two impact tests in the same location simply further increases liner stiffness and density. The penetration test is artificial as the closest real penetrating surface found in the cases surveyed is a narrow gauge steel channel.

It is recommended that the following issues be considered for inclusion in the Australian Standard for motorcycle helmets:

> Replacement of the magnesium headform with a more humanoid headform, such as the Wayne State University/Hodgson headform.

Set limits on the amount of headform deformation during impact.

The specification of a maximum density for liner foam of about 30 kg m^{-3}. This reduction will also help reduce rotational acceleration.

Require a maximum impact acceleration of about 250g for a 1.8m drop height.

Removal of the requirement of two impacts at the same location.

Require that helmets be tested on the sides and facebar area.

- Require that the helmet shell be very stiff, particularly in the temporal zone and facebar area. Possibly retain the penetration test to ensure a high level of shell stiffness particularly in these areas.

- Set stringent maximum limits on helmet mass and make this limit very low for open face helmets.

Develop a test for impact sliding resistance where the helmet and headform move horizontally at about 40 to 50 km hr^{-1}. A carefully controlled crash simulation test may be appropriate. In this test, limits are to be set on sliding resistance and rotational acceleration.

The phasing out of open face motorcycle helmets should be actively encouraged, and targets set for the reduction in mass of motorcycle helmets by use of new technology materials.

DEPINITION OF BICYCLE HELMET REQUIREMENTS

There would seem to be some fundamental flaws in the current Australian Standard bicycle helmet designs, and the purpose of this Chapter is to examine the various issues that have been identified by this project, and to discuss the changes needed to the Bicycle Helmet Standard. The following sections summarise the main project findings and the final section considers the need for changes to the Australian Standard.

PREVIOUS STUDIES

The main issue identified in previous studies is the high incidence of head injury in pedal cyclist crashes. McDermott, Klug (1982) found a greater incidence of head injury in cyclists compared with motorcyclists. The cyclist helmet wearing rates were very low in this study and consistent with European wearing rates. Sustained campaigns to increase wearing rates have now improved this situation.

Mohan et al. (1979) in a study of children's head injuries identified the need to consider the lower bending strength of the child hard tissue, and Bishop et al. (1984), has carried out comparative tests on various types of bicycle helmets using the WSU/Hodgson humanoid headform.

POST CRASH STUDY

A total of 171 bicyclist crashes involving head injury were surveyed in the post crash study. Only about 6% of cases were wearing a bicycle helmet which is below the average wearing level of about 10% recently observed in Brisbane, indicating a possible under-representation of cyclists wearing protective helmets in hospitalised head injury cases. This is indicative of the effectiveness of existing bicyclist protective helmets.

The crash involving a collision with another vehicle has a far greater probability of having a serious head injury outcome compared with all other types of bicycle crash, including falls from the bicycle and collisions with fixed objects. Collisions with other vehicles accounted for all of the 14 fatalities included in the crash survey.

There was found to be a high proportion of children with head injury: 89 crashes where the cyclist was 12 years of age or less, or 52% of all surveyed crashes. It is apparent that bicycle helmets are reducing the severity of head injury, as the following figure illustrates for all crash situations.

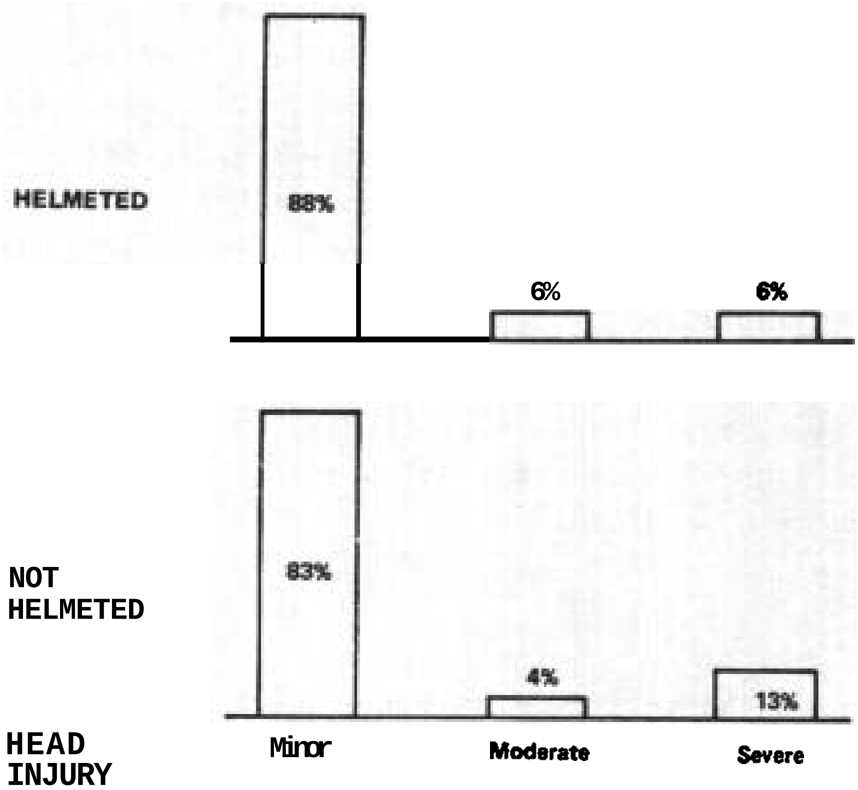
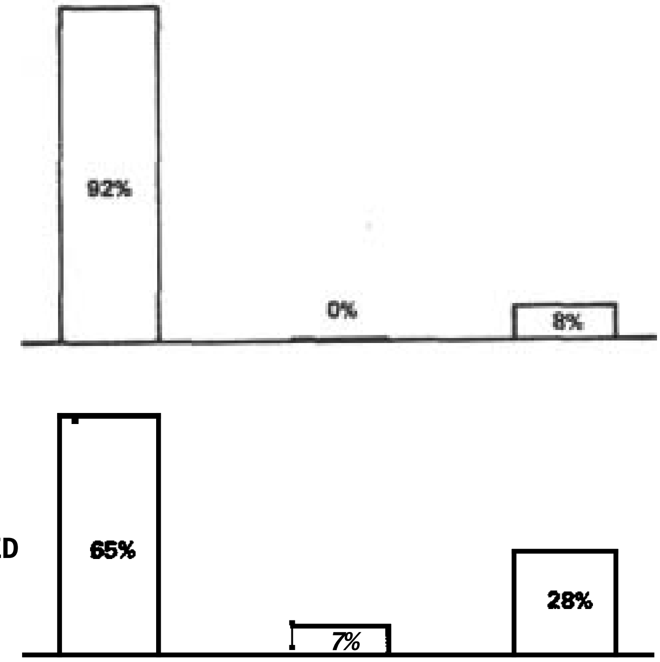


FIGURE 13 THE EFFECT OF BICYCLE HELMETS ON HEAD INJURY

If the collision with another vehicle is separately identified, as illustrated below, the effectiveness of helmets is apparent. It is also apparent that the most severe crash situation for bicyclist collisions is with other vehicles.

HELMETED



NOT HELMETED

FIGURE 14 BICYCLIST HEAD INJURY SEVERITY FOR CRASHES INVOLVING ANOTHER VEHICLE

There is a substantial variation in the bicyclist protective helmet types involved in crashes, ranging from hairnets to helmets with minimal padding to helmets with a substantial padding of stiff foam and conforming to the current Australian Standard for bicycle helmets. "he sample of helmets in the crash survey is small but only the lighter forms of head protection were found to be involved in the more serious crashes. While it is difficult to generalise, it is apparent from the crash survey results that none of the helmeted crash cases were involved in a severe high energy collision with another vehicle so the above illustrations may be showing helmet protection too advantageously.

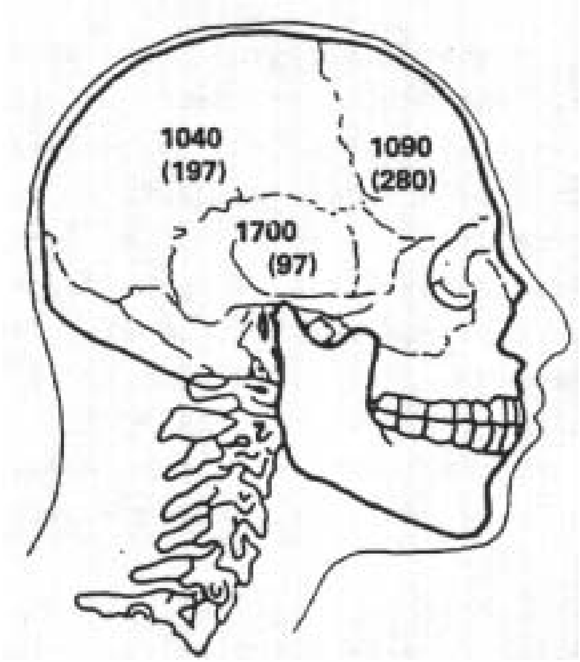
Twenty eight percent of non helmeted cases involved serious head injury and when these cases were individually assessed, 42% would definitely have had an improved outcome if a helmet had been worn. The remaining severe unhelmeted cases include severe head injury caused by impacts to the lower face, totally unsurvivable situations and impacts which caused high levels of rotational acceleration.

EXPERIMENTAL RESEARCH

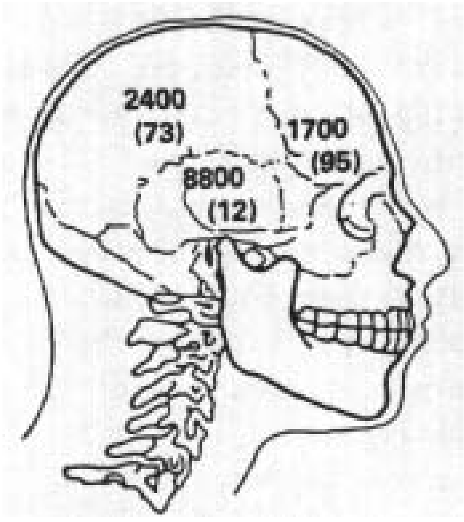
The skull hard tissue experiments have shown the child's head to be significantly more vulnerable to impact than an adult head. The child head in overall terms is considerably more flexible than an adult head. The difference is greatest at younger ages and as Mohan et al. (1979) found, by 14 to 15 years of age there is little difference between the strength properties of the youth's head and an adult head.

For the skull hard tissue tests relatively young child hard tissue from about 5 years to 12 years of age was tested. A far greater deformation at failure was measured compared with the same tests on adult hard tissue.

The following Figure illustrates the differences.



ADULT



CHILD

strain x 10^{-6} (bending inertia in brackets)

FIGURE 15 DIFFERENCES IN HEAD HARD TISSUE BETWEEN ADULT AND CHILD

The above figure indicates that the child skull has a far greater flexibility and reduced bending strength and is far less protective of the brain than an adult skull.

It is in the above context that the standard impact attenuation test for bicycle helmets needs to be considered. It is quite apparent that the solid magnesium headform used in this test can take no account of the vulnerability of the temporal area of the adult head and the greatly reduced stiffness of the child's head.

As discussed in the Chapter on impact properties, experiments conducted with the human-like WSUlHodgson headform produce impact accelerations quite different from those with the Australian Standard magnesium hcadform for impact-drop-tests onto layers of foam of varying densities. It is apparent from these tests, and *from* results of other researchers, that the helmet liner density should be reduced to about 30 kg m^{-3} if there is to be no substantial distortion of the head (causing brain damage) during a crash.

The WSU/Hodgson headform models an adult head. "here is a definite need to develop a special headform for the testing of impact attenuation of child bicycle helmets because of the greater flexibility of the child head. It is possible that liners for child. helmets may need to be less than 30 kg m^{-3}.

The jaw impact experiments have demonstrated the adverse effect of increased helmet mass on rotational acceleration. From the cadaver material head/neck model, the effect of added mass to the head on rotational acceleration seems to increase slowly up to about 1,000 gm then increase at a faster rate, as Figure 16 indicates.

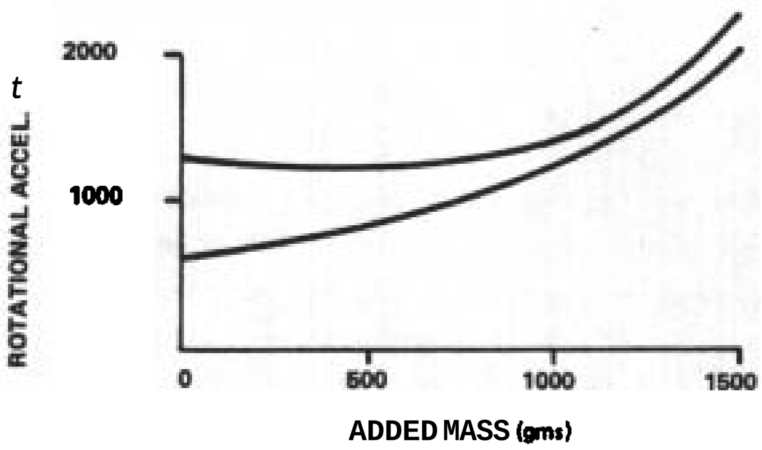


FIGURE 16 EFFECT OF ADDED HELMET MASS ON ROTATIONAL ACCELERATION

The results in the above Figure are for an adult. The effect on rotational acceleration of increasing the added mass on a child's head has not been tested, but based on the inertia properties of the head, if a child's head is one kilogram lighter and 2cm smaller in diameter, then the inertia of the child head is approximately half that of an adult head. The ability to accommodate Increased mass on the head is age dependent. A very young child, say 2 or 3 years of age would be uncomfortable with a 500 gm helmet. More work is needed in this area and it is at least postulated that a child bicycle helmet should be lighter than the comparable adult helmet.

The crash simulations using Australian Standard polymer shell bicycle helmets with the head of a dummy impacting the laboratory floor after leaving the bike simulator over the handlebars at 45 km hr^{-1} have produced high rotational accelerations. The rotational accelerations obtained were on an average of 58,000 rad s^{-2} and are about 30% higher than the polymer motorcycle full face helmets tested. More testing is needed to discover the reason for the higher rotational acceleration. The motorcycle helmets tested exhibited a quite marked difference between the polymer full face helmets and the fibreglass helmets tested, with the heavier fibreglass helmet (1.35 kg compared with 1.1 kg) tending to have higher rotational acceleration - although one brand did not.

The high level of rotational acceleration for the 0.5 kg polymer bicycle helmets could be attributable to poorer sliding properties of the helmet shell, insufficient shell impact distribution capability, or a bottoming out of the metal dummy head because of the liqht construction of the helmet.

REVIEW OF THE CURRENT AUSTRALIAN STANDARD FOR BICYCLE HELMETS

It is apparent from the post crash study that bicycle helmets produce reductions in the severity of head injury, but the experimental research has indicated considerable scope for improvements to these helmets.

The current Australian Standard does not recognise the special needs for children, yet children make up the majority of riders. Also in the post crash survey children had the highest number of head injuries resulting from bicyclist crashes. The impact attenuation test, as discussed in the previous chapter, quite inappropriately makes use of a rigid magnesium headform. *The* more human-like WSUlHodgson headform provides a more realistic impact test headform. Tests with this headform indicate that helmet padding should be reduced in density from 50 kg m^{-3} to 30 kg m^{-3}. The current peak acceleration of 400g is far too high with recent helmet researchers (Slobodnik 1979) indicating 250g for aduits.

It is very important to realise that children's heads are more flexible than adult heads. What is needed is a humanoid headform with the bending properties of the typical 10 year old and strict limits placed on the amount of headform deformation to be allowed during impact.

The most life threatening crash situation is the collision with another vehicle. Other crash situations, e.g. falling from the bicycle and the like, are usually significantly less severe and require only minimal head protection. *The* collision of a bicyclist with a vehicle is similar to a lower speed motorcyclist collision and has similar injury patterns, with a concentration of injury to the facial region. The current bicycle helmet is sinilar to the very early motorcycle helmets in that it only protects the very top of the head which has a lower probability of injury.

While it is recognised that bicyclist helmets need good ventilation and that the mass of the helmet affects rider endurance, bicycle helmets should be extended to cover the vulnerable temporal area and the facial region.

It is recommended that the following issues be considered for inclusion in the Australian Standard for bicycle helmets:

Replacement of the magnesium headform with a more humanoid headform such as the Wayne State University/Hodgson headform.

Set limits on the allowable amount of headform deformation during impact when testing smaller size bicyclist helmets.

The specification of a maximum density for the liner foam of about 30 kg m^{-3}. This requirement will also reduce the magnitude of rotational acceleration.

Requirement of a maximum impact acceleration of 250g from a 1.5m drop height for adult helmets and preferably less for child helnets.

Removal of the requirement of two impacts at the same location.

Require helmets to be extended to fully protect the temporal area and for impact tests to be carried out in this area.

Require that the shell be very stiff and have a low impact sliding reaction. Develop a standard test for measurement of impact sliding friction.

Further, it is considered that special requirements need to be be specified for children's helmets.

> The impact test headform should be designed to reflect the weakness and deformability of the child head.
>
> The vulnerable temporal area should be well protected.
>
> The mass of the child bicycle helmet should be limited to *about* one half to two thirds that of the adult helmet (250 to 350 gms).

The development of bicycle helmets to suit head properties of children should be encouraged.

RECOMMENDATIONS

MOTORCYCLE HELMETS

TITLE	NUMBER	RECOMMENDATION
Helmet impact test	1	Replace the magnesium head-form used for impact testing in the Australian Standard with a humanoid headform with a corresponding reduction in maximum allowable acceleration
Helmet liner stiffness	2	Substantially reduce liner stiffness
Two impact requirement	3	Delete the current require-ment for two impacts at the one location on the helmet
Change the helmet test zone	4	Btend the helmet test zone to include the sides and facebar of the helmet
Shell stiffness	5	The shell is to be very stiff particularly in the temporal zone and facebar area
Helmet mass	6	Set maximum limits for helmet mass particularly for open face helmets and reduce mass with new technology materials
Sliding resistance	7	Develop a test to measure sliding resistance and rotational acceleration
Open face helmets	8	If possible phase out open face helmets

BICYCLE HELMETS

TITLE	NUMBER	RECOMMENDATION
Helmet impact test	1	Replace the magnesium head-form used for impact testing in the Australian Standard with a humanoid headform with a corresponding reduction in maximum allowable acceleration
Helmet liner stiffness	2	Substantially reduce liner stiffness
Protection of the temple area	3	Extend the helmet test line to ensure the temple area is protected
Shell properties	4	The shell is to be very stiff particularly in the temporal region and across the frontal area
Facial protection	5	Encourage greater lower facial protection particularly in the temporal region
Children's helmets	6	Develop a special protective helmet for children and test it using a test headform incorporating child head stiffness properties

TITLE	NUMBER	RECOMMENDATION
Children's helmets	7	Limit the mass of children's helmets to reduce the effects of rotational acceleration during an impact by tightening shell material selection procedures
Further research	8	Actively pursue the complete definition of requirements for children's bicycle helmets. Currently, they are wearing helmets with adult padding impact tested in a way that is totally incompatible with the properties of a child's head

REVIEW OF PREVIOUS POST CRASH STUDIES

POST CRASH STUDIES

A substantial number of post crash studies has been carried out in recent years, with some investigating all bicycle and motorcycle crashes and others concentrating on helmet effectiveness. The moat notable study of motorcycle crashes in overall terms is the Hurt, Ouellet, Thom (1981) study which carried out indepth investigation of 900 motorcycle crashes in California during 1976 and 1977. Griffiths (1983) reports on similar studies carried out by the Traffic Accident Research Unit, New South Wales, where a total of 148 crashes were investigated. McLean, Brewer, Hall, Sandon, Tamblyn (1979) as part of the Adelaide In-depth Study, investigated 68 motorcyclist and 23 pedal cyclist crashes. These in-depth studies put into perspective the incidence of motorcycle crashes and the frequency of head and neck injuries.

Griffiths (1983) reports increased injury severity likelihood with increased motorcycle engine capacity, and/or increased speed at impact. The source of injuries is also indicated with about 67% of head and thorax injuries greater than AIS 3* for crashes involving cars and trucks and 33% with other objects including the road surface, roadside and road furniture. Of the 68 motorcycle crashes investigated in the McLean et al. (1979) study, 3 involved fatal brain contusions and 25% of casualties involved head injury, with 5% severe.

The Hurt et al. (1981) crash data is very comprehensive and extensively cross tabulated, enabling very accurate understanding of types and severity of motorcycle crashes. There is about 40 percent helmet usage in the motorcycle crashes because of the non

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*AIS 3 is level of severity 3 on the Abbreviated Injury Scale.

compulsory use of helmets. Hence the sample size of relevance is about 350 helmeted riders and it is interesting that of these, 196 had some head and neck injury with 6.5% of the 196 severe or worse. Hurt et al. (1981) also comment that some motorcycle crashes are of extreme severity.

Before discussing post crash studies concentrating on helmet effectiveness, the very effective work at Transport and Road Research Laboratories (TRRL) reported by Harms (1984) must be noted. Currently, TRRL are carrying out intensive in-depth investigations of fatal crashes involving extensive forensic investigation of crash sites and victims. The post crash investigations contained in this report are modelled on the TRRL post crash investigation procedure. The TRRL work is an extension of earlier, very effective work carried out by Pedder, Hagues, Mackay (1979) at the University of Birmingham. The National Health and Medical Research Council (NH & MRC) Road Accident Research Unit in Adelaide is also involved in ongoing studies of head injury crashes, with emphasis on pedestrianfcar conflicts. As reported by Gibson, McCaul, McLean, Blumbergs (1985), a very comprehensive, detailed neuro-pathology examination is carried out as part of the post mortem process to pin point the exact nature and extent of cerebral damage. The need for detailed crash assessment and, in particular, care with autopsy work is indicated by Hill (1979) in drawing parallels and differences between aircraft and traffic crash investigations.

A substantial number of post crash studies have recently been carried out in Europe and the United Kingdom, as reported in the 1984 IRCOBI* Conference proceedings, including Stocker, Loffelholz (1984), Vallee, Hartemann, Thomas, Tarriere, Patel, Got (1984), Otte, Jessl, Suren (1984), Mohan, Kothiyal, Misra, Banerji (1984) and *Harms* (1984), the latter being part of an overall investigation. These studies highlight certain crash characteristics related to helmet effectiveness.

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* International Research Committee on Biokinetics of Impacts (IRCOBI)

The location of impacts for crashes investigated is concentrated in the front and side, virtually in a band around the head with a much lower incidence on the crown area. Also of concern is the high incidence of helmet loss during collision with typically, Vallee et al. (1984) reporting 15% and Harms (1984) 18% for open face and 14% for full face. The studies mentioned earlier in this section also report high incidences of helmet loss with Pedder et al. (1979) reporting that 36% of helmets come off at some stage during crashes. Of these about a third came off without apparently sustaining major direct impact. While Pedder's results are of concern, it must be noted that Pedder only investigated fatal crashes and as always, some of these crashes would have undoubtedly been very severe.

The study carried out by the London Accident Unit, London Hospital of low speed motorcycle crashes (Cannell, King 1982), found that 50% suffered some head injury, a result higher than other studies, and that full face helmets did not confer a significantly greater degree of facial protection, but showed improved outcome for small impacts.

In summary, previous in-depth post crash studies indicate that between 25 and 50% of motorcycle crashes involved head and neck injury, with about 5 to 7% at a severe level; that cars and trucks were involved in about 2/3 of crashes. Severity was also found to increase with engine capacity and speed of the motorcycle and the location of impact was concentrated in the facial area. Between 15 and 36% of helmets were lost during collisions with helmet loss becoming more likely in very severe crashes. On the conduct of post crash studies, emphasis should be placed on accurate and detailed investigations including detailed autopsy procedures.

ANALYSIS OF CRASH DATA

A number of notable prospective studies have been carried out using records of crashes. Some have been based on police reports and hospital records, while other researchers have concentrated on fatal crashes utilising the very complete records of the Coroner's Office. Other specialised studies have been carried out in specialist hospital units.

Significant studies of motorcycle crashes involving investigation of Coroners' reports include recent Swedish work by Krantz (1985) and two Birmingham studies, Larder (1984) and Whittington (1980). The Krantz (1985) study of 132 motorcycle and moped riders shows impact sites similar to those found in the studies reviewed in the previous section, with the point of impact occurring in a band around the head with little impact below the level of the ears at the side. There is a high incidence of impacts all wer the facial area. Numerous cases of injury at the point of impact were found indicating the need for a better distribution of force by the shell and absorption of energy by the liner.

It is postulated that rotational forces may have produced a number of the injuries remote from the source of impact, a point also made by Harms (1984). It is further suggested by Krantz (1985) that increasing helmet mass produces higher traction force on the junction of the head and neck, although it was found that riders who suffered disruption at this location wore open face helmets. Larder (1984) found a high proportion of chinguard failures in full face helmets with fibreglass shells and considered this was the result of the relatively low elasticity of fibreglass. He expresses doubt that much can be gained by improving chinguards, particularly the low level of padding, because damage in this area is usually the result of a relatively severe crash. Larder (1984) also is of the opinion that liners do not effectively reduce impact energy in a significant proportion of crashes. No impact damage was shown in 44% of liners indicating a possible need to use softer liner material.

Whittington (1980) makes the valid point that some fatal motorcycle victims, although having fatal head and neck injuries, would have died anyway of causes other than head injury. While the study of fatal crashes produces some important results, it is important to realise that a proportion of these are vastly humanly intolerable.

Thomas, Foret-Bruno, Faverjon, Henry, Tarriere (1977) in a study of mopeds found that impact occurred to the forehead in 77% of cases, but for the more severe head injuries. They also found that half of the serious crashes occurred with cars and loss of helmet occurred in 25% of cases and was more likely in serious crashes. In a study of bicycle and moped crashes, Huijbers (1984) reached similar conclusions but also noted injury tolerance differences with age, with the younger bicycle riders and middle aged people being at greater risk.

Australia had a pioneering role in the introduction of compulsory motorcycle helmet legislation, and as Foldvary, Lane (1964) indicates, the use of the rather basic helmet worn in 1961 reduced fatalities by about one third. Recent studies in the United States, where compulsory helmet usage laws have been repealed, indicate a halving of fatalities where helmets are used. On the question of helmet use, a significant study is the comparison made by McDermott, Klug (1982) of head injuries for bicyclists compared with motorcyclists.

McLean (1981) found that a head injury was sustained by 65% of pedal cycle casualties compared with 25% of motorcycle casualties, which again speaks volumes about the lack of head protection among pedal cyclists in Australia. Although the incidence of head injury in pedal cyclists is high, Illingworth, Noble, Bell, Kemn, Roche (1980) puts it in context for children. They found pedal cyclist crashes in children to be similar in overall severity to skate board accidents but significantly less severe than child pedestrianltraffic crashes. This context does not necessarily reduce the importance of head protection because, as indicated later in this section, even relatively minor head injury can result in ongoing mild disability.

Yeo (1979) has studied the records of spinal units throughout Australia, including considerable follow up with victims, in an effort to better define differences in the crash performance between open face motorcycle helmets and helmets with face bars. His results seem to indicate a greater recovery from spinal cord damage in the case of helmets with face bars compared with open face helmets although the difference is just statistically significant. The recovery is often partial from tetraplegia to paraplegia indicating that the critical vertebrae for the helmet with a facebar may be lower down the cervical spine than for the open face helmet. For helmets with face bars a load path seems to develop via the clavical with the result that there is a higher incidence of broken clavicals in crash victims wearing full face helmets but reduced spinal chord damage.

In a recent investigation of neck injury. Hodgson and Thomas (1981), found that the cervical spine is weak in torsion and, where contact of the helmet curtain occurs with shoulder pads, the extent of hyperextension is reduced and cervical spine shear and bending likewise is reduced. This would seem to support Yeo's findings.

In neck injury analysis of the 900 motorcycle crashes, Hurt et al. (1981) identified a significant reduction in neck injury outcome for helmeted versus unhelmeted motorcyclists, particularly for moderate neck injury. But his neck injury data does tend to indicate an increase in neck injury for helmets with face bars compared with open face helmets, with an increase in fatalities for crashes involving facebars. Admittedly the numbers of fatalities are small, 7 out of 812 cases, and Hurt makes the point that the fatal crashes found in his survey were often totally unsurvivable irrespective of helmet usage or helmet type.

OTHER ISSUES

The protective effect of helmets needs to be considered in the context of what is currently worn and the Australian Standards. For motorcycle helmets the current Australian Standard is for an impact attenuation acceleration of no greater than 300g for each of two impacts at the one location of the helmet where a helmet and metal headform are dropped through 1.8 metres onto a solid metal surface. A stringent penetration test is also required plus a retention strap strength test. In a similar vein the Australian bicycle helmet standard defines a maximum acceleration of 300g for two impacts at the one site for a drop height of 1.5m and also has penetration and retention requirements.

The above standards, and virtually all other overseas standards, place considerable emphasis on the simple fall onto the head with no account being taken of likely impact conditions for collisions with vehicles which, as Griffiths (1983) indicates, are likely to be more severe. Equally, no consideration is given to the capacity of a helmet to slide during the initial impact making contact with a surface. As Aldman (1984) indicates, in tests involving a forward velocity plus a drop height of 1.4 metres, no sliding occurred even on relatively smooth surfaces presumably because of the high normal force acting during the impact. Recent work by Glaister, Hazell and Mortimer (1983) has resulted in the development of an oblique impact test where sliding resistance is measured. This test has been incorporated into the most recent British Standard on Protective helmets, BS 85/35265. These results indicate quite high (in the order of 1.0 and higher) levels of sliding resistance, suggesting that the vertical impact pulse can have a tangential impact pulse almost as great where there is significant forward movement.

While the Glaister test is an improvement on no assessment of sliding there is still no indication as to the effect of sliding resistance on injury outcome. Viano (1985) indicates that vascular and intracranial laceration damage can occur if the brain lags in response to head impact. Rotational acceleration is a factor which, as Krantz (1981) indicated, *may* be present in up to 50 percent of substantial motorcyclist head impact situations. Krantz further indicates that the effect of helmet mass which may magnify this type of injury.

As quoted by Glaister (1982), rotational acceleration tolerance is about 1800 rads-2 for concussion (Ommaya and Hirsch, 1970) and bridging area disruption is stated to be likely if both a critical angular acceleration of 4,500 $rads^{-2}$ and a critical angular velocity change of 30 $rads^{-1}$ are exceeded (Lowenhielm 1977). As indicated by Aldman (1984), these values can be exceeded in an impact to the forehead with an associated velocity of only 35 km/h.

Human tolerance, particularly for head impact, still remains a vexed question. Because of the involvement of the complex and only partly understood central nervous system, there is no straightforward definition of when humanly unsurvivable damage occurs. The long term effect of even relatively minor damage is still in need of definition.

The Glascow Coma Scale (Jennett 1974) has gained international acceptance as a guide to the severity of head injury. In a recent survey of all types of head injuries, O'Rourke, Costello, Yelland, Stuart (1986) found the following morbidity rates for various Glascow Coma Scale levels.

TABLE I
GLASCOW COMA SCALE INDICATORS

GCS	Morbidity (%)	Long Term Disability Moderate to Vegetative (%)
3-8	72	13*
9-12	20	28
13-15	0.2	3

Source: O'Rourke et al. (1986) resuscitated case

*Note: In the above table, only those cases surviving until admission into the neurosurgical unit and later dying are included in the morbidity proportion, so the morbidity levels for the lower GCS ranges actually under-represent the full level of morbidity.

Recent research (Rimel, Giordani, Barth, Boll, Jane 1981) indicates that emotional, psychological and intellectual problems can develop in victims with even minor head injury (defined as unconscious for less than 20 minutes and with a GCS between 13 and 15). When a group of minor head injury victims is compared with a comparative population of people that have not experienced head injury there is a significant increase in morbidity 3 months later, a higher level of unemployment as well as maladjustment. There is clinical evidence that victims with minor head injury also have organic brain damage. Gibson, McLean, Blumbergs (1985) *have* identified small areas of brain damage resulting from earlier minor injuries to the head in detailed examination of brain tissue of traffic crash victims. The same finding is reported by Ivan (1984).

There is no simple relationship between the impact intensity and the head injury sustained. In fact, the very existence of a relationship between internal brain injury and head impact kinematics has been questioned by Newman (1982). Early relationships, including the Wayne State University concussion tolerance curves, aimed basically at developing human tolerance relationships for translational head acceleration. It is now evident that rotational head acceleration can be present in most head impact situations and in some cases may be the most critical element in the development of brain injury.

Other factors produce varying severity of head injury outcome. As Newman (1975) indicates the likelihood of head injury increases with:

- the kinetic energy of the blow
- the maximum load experienced by the head
- the local pressure on the skull
- the rate of onset of loading
- the time duration of the blow

While the Glascow Coma Scale has wide acceptance as a measure of brain impairment, human survivability is dependent in large measure on the extent and severity of all injuries sustained in a crash. The most widely used injury classification is the Abbreviated Injury Scale (AIS) developed by the Committee on Medical Aspects of Road Safety (1971).

An AIS is usually separately reported for each major body component and with multiple injuries (often the case in motorcycle victims) a method of producing a combined injury scale is needed. The Injury Severity Scale (ISS) is found by summing the square of the worst three AIS rated injuries. The point of tolerance is generally accepted as about 40 (Stoner, Barton, Little, Pates 1977). One fault of the injury severity rating scales is they do not give consideration to the long term outcome for the victim.

SUMMARY OF FINDINGS

In summary recent Post Crash Studies have indicated

- Main impact areas are in a band around the head at ear level with a high incidence of impacts all over the facial area,
- A need for better distribution of force by the shell and absorption of energy by the liner,
- A problem with high mass helmet producing undesirable levels of force at the head/neck joint,
- Disadvantages of open face helmets particularly in impact force transmission to the torso,
- A high incidence of facebar failure (but often associated with severe crashes),
- A tendency for more severe head injuries at the base of the skull,
- A high level (up to 25%) of loss of helmet during the crash. In some cases such crashes would not have been survivable even if helmet was retained,
- Differences in age tolerance to crashes,
- A higher incidence of fatalities where motorcycle helmet use is not compulsory,
- A high incidence of head injury in pedal cyclist crashes,

REVIEW OF PREVIOUS HELMET TESTING AND EXPERIMENTATION

CRASH SIMULATION STUDIES

A few crash simulation studies have been undertaken recently. There is some reluctance on the part of research organisations to undertake motorcycle crash simulation because of the number of degrees of freedom and the difficulty in controlling a dummy rider so that it behaves similarly to a typical crash situation. Sone researchers (Brun-Cassan, Vincent, Fayon, Tarriere (1984), Janssen, Huijskens (1984) and Grandel, Schaper (1984)), have used full simulation of dummies (GM Hybrid II) mounted on a motorcycle and projected by a sled into various parts of cars. Otte, Jessl, Suren (1984) and Slobodnik (1979) have developed "bench top" type tests using a selected headform to attempt helmet damage replication (difficult when often helmets are only slightly damaged in a major impact). Otte et al. (1984) examined the very severe guard fence pillar impact caused by a blunt protruding steel section, which produced a well defined helmet crush and was easily replicated in the laboratory. Slobodnik's approach was to make casts of the helmet damage shape and to use the cast as an impact anvil in a helmet drop rig. A similar helmet was used for the simulation with impact speed modified by drop height. But it is worth noting that Slobodnik's interest was in helicopter pilot helmets and the types of crash situation are more controlled. The headform used by Slobodnik was the WSU/Hodgson humanoid design while Otte's work was done with a GM Hybrid II head and neck.

Where the GM Hybrid II is used both in "bench top" testing and in full simulation, it is reasonable to question the validity of using a dummy developed for in-vehicle crash studies for this type of simulation. As indicated by Hoekstra (1984), dummies need considerable improvement to come anywhere near the behaviour of an unconstrained human projected at high accelerations. Nonetheless, a significant number of full dummy impact studies have been carried out.

Care should be taken in any crash simulation work that the effect of the torso is included. Kingsbury, Herrick, Mohan (1979) indicate that in a free fall impact there is a second peak acceleration spike from the torso. Stocker et al. (1984) in comparisons of the effects of 5, 50 and 95 percentile dummies also illustrates the effect of torso. The effect of the child torso was examined by Mohan, Bowman, Snyder, Foust (1979). In extensive studies of child head injuries they found no influence from the torso, presumably because of the large relative mass of the head in children but they also found that in adult head impact the influence of the torso effectively increases the head mass by about 2.4 times.

MATHEMATICAL MODELS

There have been several attempts to model the human head and neck response to transient loading using finite element models. Two notable models are Hosey, Liu (1982) and Ward, Nahum (1979). This type of model development has been complemented by tests using cadavers and the WSU/Hodgson headform described earlier.

The human head and neck is a very complex system as the following examples indicate:

- skull material consists of two outer hard layers separated by the diploe cellular layer and thickness varies from point to point and skull to skull
- the shape of the skull and the deformation characteristics influence the distribution of pressures during impact
- the scalp material, dura layer and cerebral fluid tend to dampen impact. Outflow of fluids through the skull base opening during impact alters pressure distribution
- the neck muscle system is complex and response to impact difficult to quantify.

The more recent models use thick and thin shell elements and material property changes during impact to produce reasonable comparisons with surrogates. In relative terms finite element models can be quite helpful. For instance, Hosey et al. (1982) describes the impact on the occipital of a 6000N force (142g) and found negative pressures at the contra coup sufficient to cause injury plus a cyclic change in pressure in the brain stem from 35 kPa to -78 kPa. It was inferred that at an impact force of 7800N the negative pressure in the brain stem would have been sufficient to produce tissue tearing and at this point there has been time only for very little rotation.

Ward, in a discussion on the effect of padding on impact loading, indicates that the main effect of padding should be to smooth out the sharp spike shaped impact acceleration pulse and that there is considerable scope for optimising padding crush rates for a range of impact types. Mention is also made of injury tolerance levels and pressures corresponding to injury severity.

Walfisch et al. (1984) carried out finite element analysis of visors and was able to demonstrate relative differences between visors of varying thickness and flat versus domed shapes. Elastic properties and a statically applied force were used.

Saczalski, States, Wager, Richardson (1976) developed a fairly simplistic finite element model of head and helmet, and tried various liner stiffnesses. They compared this work with a rigid headform and were able to make some tentative comments on the need to optimise liner stiffness.

In summary, there have been some very sophisticated finite element models developed which provide some insight into the distribution of impact loads. Hosey et al. (1982) is particularly good in that respect. Simpler finite element models have been used with some effectiveness to improve understanding of load distribution and deformation of helmet shells.

TISSUE PROPERTIES

Before a human model can be synthesised, it is necessary to have measured various tissue properties. For the head and neck this includes skull and vertebra strength, muscle strengths and the strength properties of various soft tissues, as well as the combined response of such materials under impact load.

The importance of the combined effect is illustrated by Gurdjian, Lissner, Patrick (1962) where it is indicted that the strength of a cadaver head is 16 to 20 times greater than the strength of the skull bone. Much work has been done in this area with McElhaney, Melvin, Roberts, Portnoy (1973) and SAE J885 (1980) the most definitive papers in this area. McElhaney contains measurements of the dynamic mechanical properties of the skull, diploe, dura and brain tissue while SAE J885 reports on these properties as well as describing brain damage modes and providing bending and shear strength of the neck and strengths of facial bones and the mandible. An earlier paper by Melvin, McElhaney, Roberts (1970) contains further useful data on strength modulus of skull bone and soft tissue.

For several years researchers have been refining the development of finite element models, and in so doing have had to study human properties very carefully. The discipline of fine tuning a model has also helped improve understanding of human properties under impact, with Ward, Nahum (1979) reporting a variation in Poisson's ratio for hard surface impacts while Hosey, Liu (1982) describe the propagation of waves of deformation through the skull material and intracranially during and after a blow to the head. It is clear that all characteristics and properties and changes in them under impact are not fully understood and may well never be.

Mohan et al. (1979) reports on the stiffness of child skulls and postulates an age/stiffness relationship based on cadaver head strength measurement modified by what is known of progressive development of skull hard tissue in children. Based on his results the stiffness of the 10 year skull is about 80% of adult skull stiffness.

Research carried out over a number of years at the University of California by Goldsmith and other researchers and reported in the following section also serves to illustrate the complexity of trying to model the invitro response by building up a biomechanical model based on human properties.

In summary then, while a great deal of work has been done in the way of straightforward measurement of the various elastic properties of the head and neck tissues, the combined effects, particularly for impact, are not well understood.

HEAD AND NECK MODELS

Synthesised head and neck models have been developed generally to serve specific purposes. Models range from rigid headforms with a ball joint for helmet impact testing through to the Goldsmith, University of California head and neck model. The "standard" rigid headforms are well documented in the standards but the Hodgson headform initially developed for the testing of football helmets is discussed here as it is a beginning point for more realistic physical modelling of the head and neck.

The HodgsonlWSU headform consists of a self skinning eurathene representation of the skull, with a fluid silicon gel brain, silicon rubber scalp tissue and a rubber covered reinforced synthetic leather neck. It is based on anthropometric measurements of cadaver heads and has a similar load-deflection response to that of a cadaver head. It also has a steady state vibrational response matching that of cadaver heads. Hence it can be stated that the Hodgson/WSU headform has a similar elastic response to that of a real head in that it reasonably simulates the elastic squashing deformation of the head at impact. While the head of the model does have similar deformation characteristics, it does not seem to fail (fracture) at the same magnitude of impact force as an actual head. (Obviously, a reasonable headform design for use in routine testing). It is important to realise that some skull fracture can be a mode of energy dissipation. At this stage, it is unclear as to how the intracranial fluid is modelled for pressure rise. In the actual head, intracranial pressure is relieved by flow of cranial fluid out of and into the cranium via the opening in the base of the skull, as reported by SAE J885 (1980). A further complication is the representativeness of the neck. It is possible that a reinforced synthetic leather neck does provide an initial 3-4 millisecond response similar to an actual neck, but the resemblance would have to end there because the actual dynamic response of the head and neck is far more complex and has far greater total movement capability than synthetic leather would provide.

The Goldsmith work (Goldsmith, Sackman, Ouligian, Kabo 1978), has gone through several stages of development, starting off as a fluid filled sphere fixed to a GM Hybrid II neoprene neck through to a gel filled cadaveric skull fixed to synthetic cervical vertebrae with all of the significant muscles simulated using appropriate materials. The most recent model has comparative responsiveness in both the anterior/posterior direction and laterally as measured by comparison with specially stayed cadaveric headlneck material. A number of papers (Reber, Goldsmith (1979); Landkof, Goldsmith (1976); Kabo, Goldsmith (1983); Merrill, Goldsmith, Deng (1984) and Simpson, Goldsmith, Sackman (1976)), report the impact response of the various models and currently the model is being further improved to provide improved muscle representation.

It must be clearly stated that this work is mostly concerned with neck dynamics and the impact loadings used are very low, in the order of 20g, well below damage levels. Goldsmith et al. (1978) cautions that the calvarium is apparently not a linear system beyond a certain level of load and considerable caution must be exercised in extrapolating these research results to substantially higher impacts.

The work by Tarriere, Leung, Fayon (1981) and the application of Tarriere's work by Walfisch, Chamouard, Fayon, Tarriere, Got (1984) has resulted in a model capable of assessing the effectiveness of full face helmets in front impacts. The model consists of a GM Hybrid II head with the face cut away and replaced with crushable cellular material. Drop tests of the head with alternative helmets were carried out and the amount of cellular material crush related to likely facial injury using criteria developed by Tarriere.

Generally then, a great deal of interesting work has been carried out in the development of physical models with capability of providing better appreciation of what actually happens in the headfneck system on impact. The specific models of great interest to the Strength Testing Project is the HodgsonlWSU headform, the Goldsmith headlneck system and Walfisch's work on facial impact.

CADAVER STUDIES

Research in human tolerance, studies of impact and headform design for helmet testing purposes have involved tests using cadavers. Experiments using cadavers carried out at Wayne State University in the 1940s and 1950s led to the development of the Wayne State head injury concussion tolerance curve which is still widely used to set standards for safety helmets. Equally it should be stated that the early work at Wayne State may not have been completely reliable because of the positioning of the acceierometer at the rear of the head to measure a forehead deceleration (as indicated by Newman (1975)).

In general, a concern that is often raised relating to cadaver testing is the lack of muscle tone and differences in some body properties from those of the living (SAE J885, 1980). Kabo et al. (1983) also report difficulties in head and neck impact using cadavers with widely differing results because of flaccidity and dehydration. Recent work using cadavers has involved repressurising the circulatory system, SAE J885 (1980), Ward et al. (1979), Alem, Stalnaker, Melvin (1977). In particular, Alem describes in some detail a procedure for repressurising the head of a cadaver.

Mohan et al. (1979) report on the static loading of unembalmed adult cadaver heads to produce force/deflection curves which Mohan modified to provide age differences for children. Another interesting and effective use of cadaver material is outlined in the Tarriere et al. (1981) work on modelling facial crush for facial impacts. The work involved the development of cellular material with the same crush characteristics as the human face and enabled Walfisch et al. (1984) to develop improvements to full face helmets. Walfisch carried out cadaver tests to verify the simulation work, demonstrating the dilemma faced by researchers simulating human impact tolerance. One cannot be sure that dummy or humanoid simulation produces the right order of results and it is necessary to resort to "proving" the simulation using cadavers.

HELMET TESTING

A variety of helmet testing programs have been carried out by researchers ranging from standard impact drop tests with alternative headforms to a range of tests aimed at measuring the likely performance of helmets in crash situations.

The headform development by Hodgson at WSU has already been discussed and two recent drop test studies have been carried out using this headform: Bishop et al. (1984) for bicycle helmets and Slobodnik (1979) for aircrew helmets (similar in some respects to motorcycle helmets).

Hearn, Sarrailhe (1978) carried out impact drop tests, finding that impacts near the edge at the front of helmets have severe consequences because of the edge discontinuity and reduction in liner in this area. A very interesting observation is made by Sarrailhe (1984) that in the standard drop test the helmet shell is prevented from deforming because it is positioned by the solid metal headform. Consequently, the relative stiffness of liner and shell is not detected in the standard impact test because the rigid headform allows the small area of liner under the impact point to transfer the load directly from the anvil to the headform with the result that liners are not being effectively assessed.

It has been previously concluded that what Sarrailhe indicates may constitute an additional factor of safety in that the solid headform test is harsher than would occur with a human head. Kingsbury, Rohr (1981) indicates it to be conservatively based on forceldeflection measurements between "hard" and "soft" headforms. (The hard headform is the normal magnesium type and the soft headform a GM Hybrid II head, which of course is still a metal structure covered with a thin skin to partly simulate the scalp).

The indication from Sarrailhe's work is that helmet liners are too stiff, possibly because of the use of the solid headform for helmet assessment. This thought is also expressed by a number of post crash researchers. It is interesting then to find Saczalski et al. (1976), based on head elastic squashing concepts developed by finite element analysis, finds that the helmet shell helps to constrain the head from squashing and makes the further point that a hard headform acceptance test may produce a softer liner allowing too much squashing of the head with the result that deceleration is spread over a shorter time span!

Further on the question of polystyrene liners, Gale, Mills (1982) indicates that, the relatively soft 2 lb ft^{-3} (30 $kg\ m^{-3}$) foam would give a maximum acceleration below 2009 for a 30mm liner. He makes the further comment that most foams have a greater than 50% recovery which is returned to the impact surface giving a greater impulse to the head than a purely, or more, inelastic deformation would provide.

The effectiveness of the liner in crushing is also questioned by Newman (1975), refer Section 3.3, and he attempts to set energy dissipation standards. Grandel et al. (1984) indicates improved liner materials such as Hexcel allowing 90% crush.

Turning to shell property, there are two basic materials available: fibreglass or polymer, with recent research tending to favour fibreglass. Polymer tends to be more flexible (Kingsbury, 1981) readily allowing the shell structure to change shape under load. Fibreglass, however, is an inherently more rigid material which tends to crush under load. Here again is a significant difference. While fibreglass crushes, polymer tends to crack and of course once a substantial crack has developed, as Chapon, Dedoyan, Verriest (1984) indicate, a bad outcome is more likely.

Kingsbury et al. (1981) again indicate that polymer helmets show lover energy of deformation results for front loadings because of the greater flexibility of the material, indicating that front and side impact attenuation needs to be improved by thickening to make it equivalent to the top impact performance of helneta. (Possibly not at the sacrifice of a further increase in mass? Krantz (1985)). Gurdjian et al. (1962) indicates that for high velocity impact there will be a tendency to produce localised depression of the skull unless a very rigid helmet shell is used. However, it can be useful for the shell to give somewhat to redistribute the load. Fibreglass does this at the appropriate stage by crushing.

Recent work by Aldman, Thorngren, Gustafsson, Nygren, Wersall (1979) indicates that fibreglass can provide somewhat better protection against injury and that this could be a result of more plastic deformation of the shell. It is interesting to note that the improvement seems to be only evident where both shell and liner are damaged indicating that the protective effect of the liner has to be optimal and related to shell stiffness, the point also made by Sarrailhe (1984).

Aldman is currently experimenting with frictional properties of helmet shells as reported by Newman (1979). It would seem that there is some evidence that polymer helmet shells tend to grip the road surface rather than sliding smoothly.

Glaister (1982) reports similar work using inclined surfaces in a helmet drop rig which provided the impetus for the new requirements of the latest British Standard.

In summary, helmet shells are in need of strengthening in some key areas, notably the face and side; fibreglass seems to have preferred properties and possibly does not suffer as much from frictional gripping problems. The weight of opinion is that liners should be reduced in stiffness and, if economically reasonable, a material with a greater amount of permanent deformation should be used. It is interesting that only fibreglass shells meet the Snell motorcycle helmet standard.

SUMMARY OF FINDINGS

Crash simulations seem to provide a means of understanding impact forces in crash simulations. There are no reliable nathematical or computer models which properly define head and neck dynamics during impact, although the Hosey et al. (1982) model does indicate high brain stem tearing stresses due to impact on the occipital and may be evidence of an effect from rotational acceleration.

Mohan et al. (1979) postulates an agefstiffness relationship with a 10 year old skull at about 80% of adult stiffness.

The HodgsonlWSU headform design is based on load deflection response measurement of cadaver heads and has a similar elastic response to a real head. It has been used by Hodgson to develop football helmet standards in the U.S. and it is gaining acceptance as a more realistic headform.

There is no headfneck model available in any form which will reliably model a head impact and it seems that researchers resort to cadaver testing to "validate" tests carried out with various types of dummies and headforms.

Sarrailhe (1984) indicates that standard protective helmet impact tests may not be effectively testing the stiffness of shell and liner and that liner densities are too high. Gale et al. (1982) indicates the advantages of low density liners and makes an important point concerning liner elastic rebound. Fibreglass helmet shells are less deformable and have lower frictional resistance in sliding.

MOTORCYCLE HELMET POST CRASH SURVEY

GENERAL

The survey includes a total of 158 motorcyclist crashes involving head injury, of which 71 were fatal. Severe crashes are over-represented in this survey, as the ABS (1985) indicates that for all motorcycle crashes 5.4% of hospitalised injuries were fatal. Since the survey is focused on head injury, it should be viewed as a subset of the overall epidemiological characteristics of motorcycle crashes. The survey includes head region injury cases admitted to major Brisbane hospitals plus fatal motorcyclist crashes both in and outside Brisbane over a 12 month period.

Table II describes the general features of the motorcycle crashes surveyed, including 14 cases where no helmet was worn. Of these, 2 were fatal but only 3 of the 14 involved a collision with another vehicle. Most of the no helmet worn cases were off road (trail bike) crashes.

TABLE II
GENERAL MOTORCYCLIST CRASH SITUATIONS
INVOLVING HEAD REGION INJURY

Type of Crash	Total	Injured No	Injured %	Fatal No	Fatal %
Fall to pavement on ground	24	23	96%	1	4%
Collision with fixed object	34	16	47%	18	53%
Collision with another vehicle	100	48	48%	52	52%

Note: Care should be taken in interpreting the above data because it includes a high number of fatal crashes.

Table III below indicates more male victims in the 17 to 25 age group than females. This result is similar to previous studies (McDermott & Klug (1982).

TABLE III
DISTRIBUTION OF HEAD REGION INJURY CRASHES
BY AGE GROUP & SEX

Age Group	Male		Female	
	Injured	Fatal	Injured	Fatal
under 17	10	3	2	-
17 to 20	12	17	2	1
21 to 25	35	35	7	-
over 25	18	15	1	-
	(75)	(70)	(12)	(1)

There are two main types of helmet: the open face and the full face helmet incorporating a face bar. Table IV reflects helmet use amongst motorcyclists involved in crashes which resulted in head injury and, as discussed earlier, has a bias towards serious life threatening head injury.

TABLE IV
BASIC HELMET TYPES FOUND IN SURVEY

Type	Total	Injured		Fatal	
		No	%	No	%
open face	24	14	58%	10	42%
full face	97	53	55%	44	45%
	121	67	55%	54	45%

Note: The above table contains 121 helmeted cases. Of the total of 158 cases, 14 were unhelmeted and a further 23 were lost at the crash, burnt, destroyed, withheld or destroyed by next of kin or held by Police for inquest evidence.

To further define the extent of injury Table V provides information on the head Abbreviated Injury Scale (AIS) by age group. Care must be taken in interpreting the table because the more severe motor cyclist crashes involve high impact energy levels. (Severity increases to AIS 6).

TABLE V

MOTORCYCLIST HEAD INJURY SEVERITY BY AGE GROUP

		HEAD AIS						
Age Group	Total	AIS0	AIS1	AIS2	AIS3	AIS4	AIS5	AIS6
under 17	15	3	1	6	4	-	-	1
17 to 20	32	9	3	5	4	2	4	5
21 to 25	77	30	3	9	12	3	8	12
over 25	34	13	4	6	3	5	2	1
Total	158	55	11	26	23	10	14	19

So as to provide an impression of overall injury levels Table VI provides a summary of the Injury Severity Scale (ISS) levels in which the above crash cases lie. A description of the AIS and ISS Scales can be found in the Chapter, a Review of Previous Post Crash Studies. There is usually a very high probability of morbidity if the ISS exceeds 40.

TABLE VI

EXTENT OF OVERALL INJURY IN MOTORCYCLIST CRASHES SURVEYED

	INJURY SEVERITY SCALE					
	0 - 10	> 10-20	> 20-30	> 30-40	> 40-50	> 50
No. injured	47	18	6	1	1	1
No. fatal	0	1	10	19	15	21

Note: There are 140 cases in this table. 18 cases including 5 fatal, had no medical record.

CRASH SITUATIONS

Because a motorcyclist usually has substantial momentum, the type of crash has a significant bearing on the injury outcome. For single vehicle crashes, (that is where only the motorcycle is involved) outcome can often be less severe. But it can be worse where the motorcycle falls onto the rider or the rider impacts with solid fixed objects such as trees, posts, and concrete kerbs. Crashes involving another vehicle usually result in more severe head injury, particularly if the head impact is directly into the vehicle or caught by vehicle structures compared with a glancing impact.

The following tables describe injury severity in terms of head AIS for the main crash situations, for both helmeted and unhelmeted cases.

TABLE VII
MOTORCYCLIST HEAD INJURY SEVERITY FOR MAIN CRASH SITUATIONS SURVEYED - HELMETED

		HEAD AIS						
Crash Situation	Total	AIS0	AIS1	AIS2	AIS3	AIS4	AIS5	AIS6
Single Vehicle Fall	17	3	6	4	3	-	-	1
Collision with Fixed Object	30	10	1	4	5	1	3	6
Impact with Other Vehicle	97	40	3	13	12	7	10	12
Total	144	53	10	21	20	8	13	19

TABLE VIII
MOTORCYCLIST HEAD INJURY SEVERITY FOR MAIN CRASH SITUATIONS SURVEYED - UNHELMETED

Crash Situation	Total	HEAD AIS AIS0	AIS1	AIS2	AIS3	AIS4	AIS5	AIS6
Single Vehicle Fall	7	1	-	3	2	1	-	-
Collision with Fixed Object	4	-	-	1	1	1	1	-
Impact with Other Vehicle	3	1	1	1	-	-	-	-
Total	14	2	1	5	3	2	1	-

In the above tables for single vehicle falls 22% of helmeted motorcyclists sustained head injury > = AIS3; while for unhelmeted motorcyclists the proportion was 43% or almost double. Crashes involving impacts with other vehicles do not show an 'expected result and reflect the great variation in possible outcomes. It also reflects a difference in the nature of the unhelmeted rider crashes, as the majority occurred were off-road.

There are marked differences in injury outcome for different crash situations. The differences reflect the variations in the rate at which the rider's momentum energy is dissipated. While the crash situation is important the impact speed also has a large bearing on the injury outcome. Table IX shows impact speed is related to overall ISS. The impact speed is taken as the speed vector just at impact. For a collision with another vehicle it is therefore the resultant of the two vehicle speeds (e.g. if a motorcyclist at 80 km/h is about to impact with a vehicle travelling in the opposite direction at 40 km/h the resultant speed vector would be 120 km/h).

TABLE IX
MOTORCYCLIST ISS RELATED TO IMPACT SPEED

Collision Speed Km/hr	Total	Injury Severity Scale 0-10	>10-20	>20-30	>30-40	>40-50	>50
0-20	0	-	-	-	-	-	-
20-50	12	7	2	2	1	0	0
>50<80	40	17	8	4	3	5	3
80+	48	8	4	6	10	4	16

In the above table there is a clear trend towards increased ISS as collision speed increases. Below 50 km/h there is a reasonable probability of survival with no ISS greater than 40. Between 50 kmlh and 80 kmlh, 20% of surveyed crashes were over ISS 40 and for crashes at 80 kmlh and above (some were considerably higher) 34% were over ISS 40.

In the following section the influence of helmet type on crash outcome is assessed.

HELMETS SURVEYEI

Wile the two main types of helmet found in the survey are open face and full face, there is a considerably greater range if the extent of helmet protection, shell and liner material types and thicknesses are also considered.

The shell material can be broadly classified in two types, fibreglass material and polymer material. Polymer based shells include polycarbonate, ABS and thermo plastics. The impact absorbing liner material is predominantly polystyrene foam with little variation in density for different helmets. Virtually all helmets analysed have a foam density of 50 Kg/m^3. The extent and thickness of the polystyrene foam padding is variable. Most helmet standards define an area to be protected during impact.

protected during impact. For instance the Australian Standard for Motorcycle Helmets AS2512 defines a zone only covering the upper part of the head for impact attenuation. Most helmets have a curtain extension covering at least part of the laser head and the full face helmets have, in addition, a facebar protecting the lower face.

TABLE X
VARIATIONS IN SHELL THICKNESS

	Shell Thickness			
	Crown mm	Temple mm	Curtain mm	Facebar mm
Fibreulass Helmets				
typical thickness	4	3-4	3-4	3
range for various brands	2-5	2-5	2-5	2-5
Polwer Helmets				
typical thickness	4	4	4	4
range for various brands	3-6	3-6	3-6	3-8

TABLE XI
VARIATIONS IN LINER THICKNESS

	Liner Stiffness			
	Crown mm	Temple mm	Curtain mm	Facebar mm
Fibreulass Helmets				
typical thickness	28	26	26	10
range for various brands	22-36	12-30	18-32	3-16
Polwer Helmets				
typical thickness	30	24	24	10
range for various brands	22-34	22-34	16-34	2-18

Some helmet types protect with polystyrene foam for the full extent of the helmet curtain, while other types provide a less dense foam material or a thinner layer of foam in this region. The same is true of the facebar in full face helmets. The extent and type of padding varies, with most types only having a relatively thin layer of foam material while some have a polystyrene layer.

The face bar shell of full face helmets also varies in stiffness from quite flexible polymer material to very stiff fibreglass which is often further reinforced by moulded ridges. Tables X and XI summarise the variation in thickness for the main helmet types.

The above tables represent quite a range in shell and liner thicknesses and the impact attenuation is likely to be quite variable. To provide some comparison of the impact characteristics represented in the above tables, a range of helmets representing the broad range of properties were dropped through a height of 1.8 metres onto a flat steel anvil using the Hodgson Wayne State University headform.

The resulting range in impact attenuation is given in the following table.

TABLE XII
IMPACT ATTENUATION FOR HELMETS IN SURVEY

Description	Impact Attenuation
Typical Fibreglass brands	155g - 280g
Typical Polymer brands	170g - 180g

Note: Impact attenuation was measured by dropping helmets onto a flat steel anvil through a height of 2 metres using a Wayne State University Hodgson headform. This headform was used rather than the metal headform in AS2512 because it resembles the elastic properties of a human head.

The variation in facial protection, and curtain extent and material variation also results in considerable variation in helmet mass. The following table gives the distribution of mass for the main types of helmet in the survey and indicate a range from 0.7 kg to over 1.7 kg.

TABLE XIII
VARIATION IN OVERALL HELMET MASS

Helmet Type	Mass (Kgs) 0.7-0.9	0.9-1.1	1.1-1.3	1.3-1.5	1.5-1.7
open face polymer	18%	64%	18%	-	-
open face fibreglass	-	-	83%	17%	-
full face polymer	-	10%	76%	14%	-
full face fibreglass	-	-	9%	55%	36%

The following tables illustrate the effect of helmet type and shell material on injury severity for different levels of impact speed. About 40 of the very severe, virtually totally unsurvivable, crashes have been eliminated.

TABLE XIV
HEAD INJURY FOR MAIN HELMET TYPE BY IMPACT SPEED

Speed Range	Total	AIS0	AIS1	AIS2	HEAD INJURY AIS3	AIS AIS4	AIS5	AIS6
20 - 50 Kmlh								
open face	3	1	0	1	-	-	1	-
full face	8	I	2	3	-	2	-	-
>50 <80 Km/h								
open face	2	1	-	-	-	1	-	-
full face	39	13	4	7	9	1	1	4
80+ Kmlh								
open face	12	2	1	2	2	-	2	3
full face	31	11	-	4	2	1	4	9

TABLE XV

HEAD INJURY FOR MAIN HELMET SHELL PROPERTIES BY IMPACT SPEED

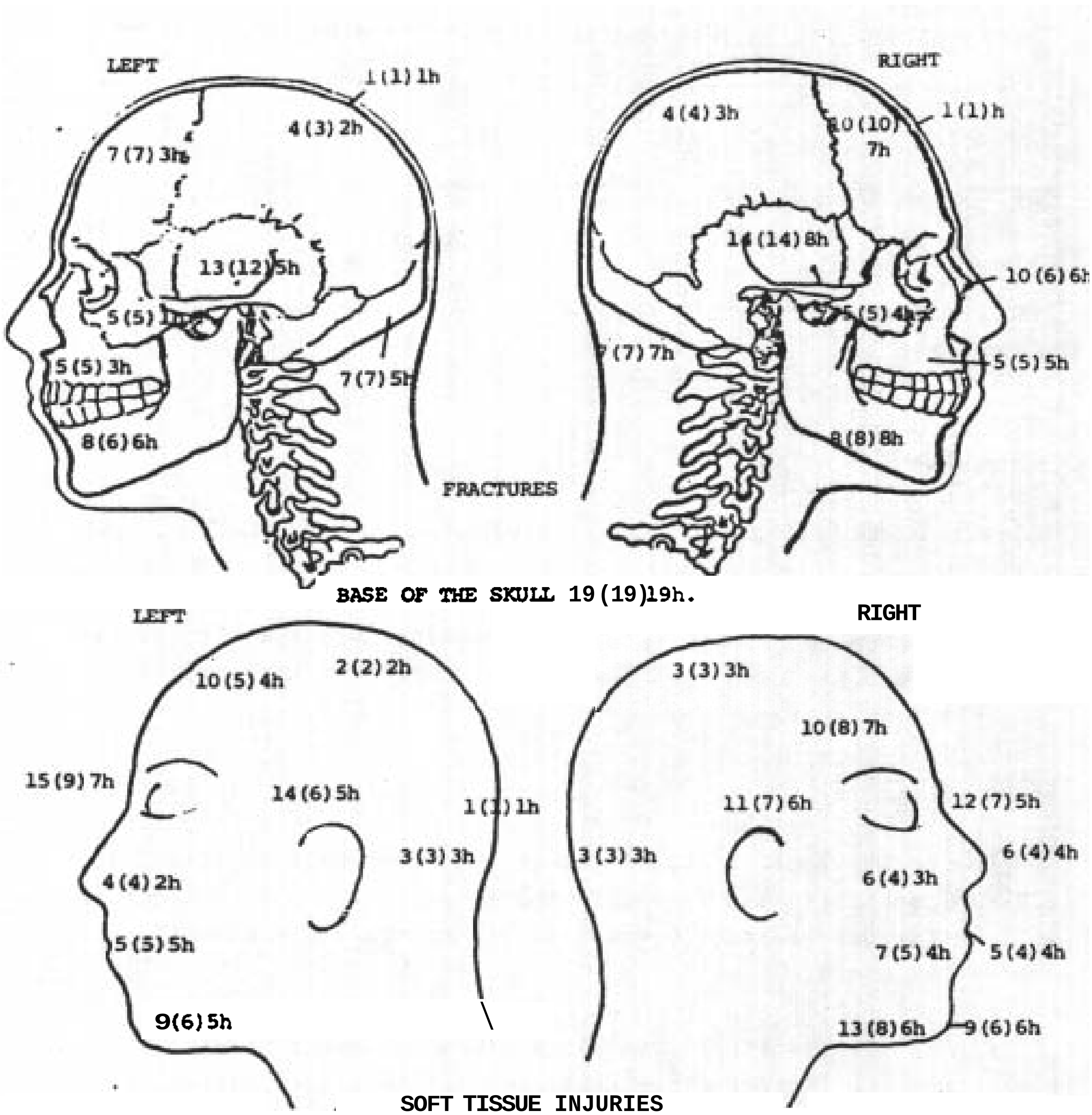
		HEAD INJURY AIS						
Speed Range	Total	AIS0	AIS1	AIS2	AIS3	AIS4	AIS5	AIS6
20 - 50 Km/h								
fibreglass	5	1	2	1	-	1	-	-
Polymer	6	1	-	3	-	1	1	-
>50 <80 Km/h								
fibreglass	17	1	4	5	4	1	1	1
Polymer	11	3	-	2	5	1	-	-
>80+ Km/h								
fibreglass	26	7	-	3	4	1	3	8
Polymer	17	6	1	3	-	-	3	4

Note: There does not seem to be any statistically significant difference between injuries sustained where a fibreglass helmet has been worn compared with polymer helmets in the above table.

INJURY SEQUELAE

In the previous sections injury has been defined using the AIS rating of head injuries. In this section the injury sequelae will be further assessed, impact sites defined and and long term outcome examined.

Figure 17 identifies the location of impacts to the head region for the motorcyclist crashes involving significant head injury in the survey.



Note: Numbering indicates ALL motorcycle accidents, where AIS scores vere recorded in each location followed by the number of accidents recording AIS > = 3, and the number of helmeted motorcyclists having this injury recorded at AIS 7= 3 follows, as e.g. 6h.

FIGURE 17 MOTORCYCLIST HEAD INJURY IMPACT LOCATIONS

The injury patterns show a high incidence of significant injury below the helmet test line and a high occurrence of base of skull injury, although the overall high severity of the motorcycle crashes in this survey must be taken into account.

In Table XVI the GCS is examined for the two main types of helmet (full face and open face) and for the two main shell types (fibreglass and polymer). Cases with ISS in excess of 50 have not been included as it is reasonably considered that these are unsurvivable crash outcomes. In some cases there has been a massive chest injury and a head injury. Those cases where the chest injury was sufficient to make the outcome morbid were also eliminated.

TABLE XVI

INVOLVEMENT IN HEAD INJURY OF MAIN HELMET TYPES

Helmet Type	Total	Glascow Coma Scale 15 - 12 Minor	11 - 8 Moderate	7 - 3 Severe
Fibreulass				
Full face all	55	30	3	22
Full face fatal	20	0	0	20
Open face all	5	2	0	3
Open face fatal	2	0	0	2
Total - all	60	32	3	25
- fatal	22	0	0	22
Polwer				
Full face all	23	9	1	13
Full face fatal	13	0	0	13
Open face all	13	7	0	6
Open face fatal	6	0	0	6
Total - all	36	16	1	19
- fatal	19	0	0	19

The above table shows that 12% of wearers of fibreglass helmets with injuries in the severe 7-3 GCS range survived, while no wearers of polymer helmets in that range did. For full face helmets, 40% of fibreglass helmets and 57% of polymer helmets were worn in crashes that resulted in injuries in the severe GCS range. Full face helmets are separately tabulated for the main helmet types in Table XVII.

TABLE XVII

HEAD INJURY FOR THE MAIN TYPES OF FULL FACE HELMET

Helmet Type	Total	Glascow Coma Scale 15 - 12 Minor	11 - 8 Moderate	7 - 3 Severe
Full face Fibreglass	55	30	3	22
Accum. %		56%	60%	100%
Full face Polymer	23	9	1	13
Accum. %		39%	43%	100%

The above table demonstrates that there is a greater involvement of polymer helmets in moderate and severe head injury.

EVALUATION OF MOTORCYCLIST CRASHES

In overall terms the fibreglass shell motorcycle helmet seems to have a lover number of severe head injury cases than the polymer motorcycle helmet. Because motorcycle helmet liners are virtually all of the same density, it is not possible by survey to study the effect of differences in liner density or stiffness on crash injury outcome.

Figure 17 indicates a very substantial incidence of impact to the facial and temporal zones. With consideration given to Yeo’s (1979) work, it is interesting to compare neck injury for full face helmets with that for open face helmets. Open face helmets are now less popular and the numbers of them found in the survey are small. Table XVIII compares the two helmet types for neck injuries AIS.

TABLE XVIII

COMPARISON OF NECK INJURY FOR FULL AND OPEN FACE HELMETS

Helmet Type	NECK AIS 1	2	3	4	5	6
Full face	2	1	2	1	3	2
(full face unsurvivable)			1			2
Open face				1		3
(open face unsurvivable)						1

The above table also indicates the relatively low incidence of neck injury. Out of 158 cases, 15 involved neck injury and of these 4 were in crash situations considered to be unsurvivable. The table shows a proportion of less severe neck injury for full face helmets but none for open face helmets.

The influence of helmet shell composition for facebar impact is assessed in Table XIX comparing fibreglass and polymer full face helmets where a first impact was made to the face bar. The cases are grouped into the various levels of GCS. Unsurvivable head and chest injury cases have been eliminated.

TABLE XIX

FIRST IMPACT TO FACEBAR FOR FIBREGLASS AND POLYMER HELMETS

	Glascow Coma Scale 15 - 12 mild	11 - 8 moderate	7 - 3 severe
Fibreglass face bar	13	2	7
Polymer face bar	1	1	3

Table XX compares the above results with all full face helmet cases surveyed.

TABLE XX

COMPARISON OF FACE BAR IMPACT WITH ALL CASES

	Glascow Coma Scale		
	15 - 12 mild	11 - 8 moderate	7 - 3 severe
Fibreulass			
All impact cases	30	3	22
Face bar impact	13	2	7
Accum. % Facebar impact	59%	68%	100%
Polvmer			
All impact cases	9	1	13
Face bar impact	1	1	3
Accum. % Facebar impact	20b	40%	100%

There is virtually the same proportion of facebar impacts for both helmet types. For moderate and severe head injury resulting from a facebar impact 41% of fibreglass facebars are in this range compared with 80% of polymer facebars.

FINDINGS - MOTORCYCLIST CRASHES

There is a high probability of fatal injuries in collisions with a fixed object or another vehicle.

Males aged 21 to 25 predominate the crashes surveyed and have the highest proportions of severe crashes.

Some crashes, because of a combination of high speed and direct impact with objects or vehicles, are unsurvivable regardless of the type of helmet.

For single vehicle (i.e. motorcycle only) crashes there is a higher incidence of more severe injury in unhelmeted motorcyclists.

The impact speed influences overall injury. Over 80 kmlhr impact speed shows a high level of severe overall injury and in the more typical 50 to 80 km/hr speed range there is still 20% with an overall very severe injury level.

There are basically two motorcycle helmet shell types, fibreglass based and polymer based. There is quite substantial variation in thickness particularly away from the standard test zone.

Virtually all liners are of the same density, approximately 50 kg m^{-3}, but there is a substantial variation in thickness again usually away from the standard test zone. Face bars have about one third of the normal helmet padding thickness and some brands have virtually no padding in this area.

- Helmet mass is quite variable ranging from open face polymer helmets at about 0.7 Kg up to full face fibreglass helmets over 1.7 Kg.

A high proportion of impacts occur to the facial area.

Full face fibreglass helmets seem to have a better outcome in severe crashes.

There is a relatively low incidence of neck injury even in relatively severe crashes. There is a little evidence that full face helmets may help stabilise the neck.

- Polymer helmets have a higher incidence of severe head injury resulting from an initial impact to the face bar compared with fibreglass helmets.

BICYCLE HELMET POST CRASH SURVEY

GENERAL

A total of 171 bicyclist crashes involving head injury were surveyed, including 8 helmeted cases forwarded from the Road Traffic Authority Melbourne. These 8 cases all involved helmet usage and an impact to the head region. Of the 163 Queensland bicyclist crashes there were only 10 or 6.2% with helmet head protection. This wearing rate is not necessarily that of the cycling population as the sample of 163 bicyclist crash cases is biased. They are all either hospital admissions to a neurosurgical unit or fatalities.

The helmet usage of 6.2% amongst the crash cases is substantially lower than the approximate average wearing of 10% based on surveys.[1]

While there is substantial variability in the type of crash and severity of impact, it is possible to group the crashes into three main categories, a fall from the bicycle, a collision with

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(1) The approximate average wearing rate was calculated by applying helmet usage for various age groups to the age distribution in the crash cases. Helmet usage for primary and secondary school age groups was taken as 13% and 2.7% respectively, based on a survey conducted in mid 1985 by the Federal Office of Road Safety following a helmet wearing promotion campaign. Helmet usage for post secondary school riders was based on users of an inner city bikeway surveyed in May 1986 and found to have an average of 28% helmet usage.

a fixed object (tree, post or stationary car) and collision between the bicyclist and a moving vehicle (truck, car, motorcycle or another bicycle). A summary by main crash types is given in Table XXI and it is interesting that all 15 fatalities involved a collision with a moving vehicle.

TABLE XXI
MAIN CRASH SITUATIONS

Type of Crash		All Crashes Helmet	All Crashes No Helmet	Fatal Crashes Helmet	Fatal Crashes No Helmet
Fall from bicycle	No.	4	83	0	0
	%	4.5	95.5	-	-
Collision with fixed object	No.	2	5	0	0
	%	28	72	-	-
Collision between bicycle and vehicle	No.	12	65	1	14
	%	15.7	04.3	7.1	92.7
TOTAL	No.	18	153	1	14
	%	10	89.5	7	93

NOTE: The above table includes the 8 unhelmeted cases from RTA, Melbourne.

Comparing helmeted and unhelmeted cases there is a much lower incidence of helmeted cases involved in the three crash categories. Out of 64 collisions between a bicycle and a vehicle 15 or 23% were fatal. Of these 15 fatalities, 3 were unsurvivable, involving severe, multiple injuries as a result of a high energy collision and would have been morbid irrespective of the type or quality of head protection. If the remaining 12 fatal cases are considered out of a total of 77 vehicle collision cases, fatal accidents represent about 16% of vehicle collision cases.

In Table XXII the head injury crash cases are disaggregated into age groupings and by sex. The age groupings of under school age (< 5 years), primary school age (5 to 12 years), secondary school age (13 to 17 years) and adult (18+ years).

TABLE XXII

DISTRIBUTION OF CRASH CASES BY AGE GROUPS AND SEX

Age Group	Total	Male		Female	
		Helmet	No Helmet	Helmet	No Helmet
< 5 years	5	-	4	-	1
5-12 years	04	7	60	-	17
13-17 years	49	5	30	-	6
18+ years	33	5	19	1	8
TOTAL	171	17	121	1	32

Note: includes helmet cases from RTA, Melbourne

It is apparent that the primary and secondary school age groups represent the greatest proportion 77% of head injury cases. Also, females seem to be less involved in head injury crashes: 19.4% of that total. Only in one female case was a helmet used. (A recent helmet usage survey conducted in Brisbane on an inner city bikeway found that female usage of helmets was 14.7% compared with male usage of 34%).

INJURY SEQUELAE

The most appropriate method of assessing brain damage is the Glascow Coma Scale (GCS), although the Abbreviated Injury Scale (AIS) ranks injury severity. Assessment using the AIS allows the overall Injury Severity Scale (ISS) to be produced. The ISS provides a description of the range of overall injury severity in the bicyclist crash sample. As Stoner et al. (1977) indicates, human tolerance is usually taken as about ISS 40. He found 50% morbidity at this level of ISS for all types of injury. The Table below indicates a 25% chance of survival when ISS is between 30 and 40 and 100% morbidity above 40 which is somewhat worse than the Stoner et al. findings for all injury situations.

TABLE IV
DISTRIBUTION OF CRASH CASES BY ISS

	Injury Severity Scale					
	0-10	>10-20	>20-30	>30-40	>40-50	>50+
Number	130	22	9	4	2	4
%	75.8%	12.9%	5.3%	2.3%	1.4%	2.3%
Number fatal	0	2	4	3	2	4
%	0%	13.3%	26.6%	20.1%	13.3%	26.6%

It can be reasonably assumed that an ISS of over 40 indicates an unsurvivable crash, although in none of the severe cases in the above table was a helmet worn. The presence of a protective helmet could reduce the head injury component and may bring down some cases in the 40 to 50 range to the 30 to 40 range and provide some survival likelihood. A detailed assessment of the 4 cases with an ISS in excess of 50, showed that 3 would have been grossly unsurvivable even if a helmet had been worn, and were the result of high speed collisions of a vehicle with a bicyclist. In the further analysis of the bicyclist crash data the 3 unsurvivable cases have been eliminated.

In Tables XXIV and XXV the head AIS is given by age group for unhelmeted and helmeted cases respectively. The head AIS includes skull fractures and brain damage and represents the most severe of all the identified head injuries in each case.

TABLE XXIV
HEAD INJURY AIS BY AGE GROUP - UNHELMETED

		Abbreviated Injury Scale						
Age Group	Total	0	1	2	3	4	5	6
< 5 years	5	-	2	2	1	-	-	-
5-12 years	84	13	26	28	10	6	1	2
13-17 years	47	6	8	27	4	2	-	-
over 17 years	30	1	5	13	6	-	4	1
TOTAL	166	20	41	70	20	8	5	3
Accum. %	100%	12%	38%	79%	91%	96%	98%	

TABLE XXV
HEAD INJURY AIS BY AGE GROUP-HELMETED

		Abbreviated Injury Scale						
Age Group	Total	0	1	2	3	4	5	6
< 5 years	-	-	-	-	-	-	-	-
5 - 12 years	7	4	1	2	-	-	-	-
13 - 17 years	5	1	-	3	1	-	-	-
Over 17 years	6	1	-	4	-	-	1	-
TOTAL	18	6	1	9	1	-	1	-
Accum. %	100%	33%	39%	89%	94%			

If the cases in Table XXIV with an AIS of zero are removed, a comparison of the proportion of injuries at AIS 3 3 and above (serious injuries) indicates that there is substantially more serious head injury in the over 17 years group although the 5-12 year age group is also high. Table XXVI indicates a substantial reduction in the presence of serious head injury (AIS > = 3) where some form of head protection has been used by the bicyclist.

A similar result is apparent using GCS for each of the cases, as Table XXVI indicates 12.5% of unhelmeted cases and 5.5% (one case) of the helmeted cases were at the severe level and for the combined moderate and serious levels 16.0% were unhelmeted and 11% wore a helmet. The comparisons between helmeted and no helmeted cases using AIS and GCS have not considered the crash situation. Table XXVI indicates a substantially higher level of fatality for a crash situation involving another vehicle. Tables XXVII and XXVIII examine the differences in more detail.

TABLE XXVI
HEAD INJURY GCS

Helmet Use	15-12 Minor	11-8 Moderate	7-3* Severe
No helmet cases	121	5	18
No helmet fatalities	-	-	12
Helmet cases	16	1	1
Helmet fatalities	-	-	1

* includes pre hospital admission fatalities

Table XXVII gives head injury AIS for two main crash situations: a collision between a bicyclist and another vehicle, and other situations including impacts with fixed objects and falls from bicycles. Table XXVIII gives the range of GCSs for the same crash situations.

TABLE XXVII
HEAD AIS BY MAIN CRASH SITUATIONS

Crash Situation	Total	AIS 0	1	2	3	4	5	6
Vehicle Collision: helmet worn	12	5	-	6	-	-	1	-
Vehicle Collision: helmet not worn	62	5	10	20	13	7	4	3
Other crash: helmet worn	6	1	1	3	1	-	-	-
Other crash: helmet not worn	88	9	29	42	7	1	-	-

TABLE XXVIII
HEAD GCS BY MAIN CRASH SITUATIONS

Crash Situation		GCS 15-12 Minor	GCS 11-8 Moderate	GCS 7-3 Severe
Collision:	helmet worn	11	0	1
	fatal	0	0	1
Collision:	helmet not worn	40	4	17
	fatal	0	1	10
Other crash:	helmet worn	5	1	0
	fatal	0	0	0
Other crash:	helmet not worn	85	2	1
	fatal	0	0	0

In Table XXVII the difference in serious injury between a collision involving another moving vehicle and other crashes (a simple fall or collision with a fixed object) is evident. For bicycle/vehicle collisions one helmeted rider (8.3%) had AIS >= 3 compared with 27 cases (45%) where no helmet was worn. In other crashes one helmeted rider had injuries of AIS >= 3 (20%) compared with 8 (9.1%) where no helmet was worn. It is evident, as would be expected, that collisions involving another moving vehicle are more severe and the injury reduction effect of existing head protection is evident and is statistically significant. Table XXVIII indicates a similar situation.

The helmeted cases have a considerable range of protection property. This is further discussed in the next section.

HELMET PROTECTION

There is a wide range of bicycle helmets available on the market, ranging from hairnet type to lightweight helmets with a polymer shell and minimal padding to more substantial helmets conforming to the Australian Standard AS 2512.

Lightweight helmets are preferred by some cyclists because of the effect of helmet mass on riding comfort and effort., while helmets conforming to the Australian Standard have a guaranteed level of impact attenuation and penetration resistance as defined by tests in the Standard.

The helmets worn in crashes in this survey can be grouped into the following types:

TABLE XXIX

HELMET TYPES

Type	Approximate Mass (g)	Impact Attenuation	No.
1. Hairnet (vinyl covered padding)	190	470	2
2. Minimal padding polycarbonate shell	300	340	5
3. Fibreglass shell minimal padding	400	280	1
4. Polycarbonate shell or similar and about 20mm of polystyrene liner	550	180	8
5. BMX type plastic with thick foam liner	740	180	2

Note: Impact attenuation was measured by dropping helmets onto a flat steel anvil through a height of 2 metres using a Wayne State University Hodgson headform. This headform was used rather than the metal headform in AS 2512 because of its considerable biofidelity.

Given the considerable range of impact protection indicated above, it is important that the head injury conditions reflected in Table XXVIII be reassessed. The following Table provides the approximate impact attenuation as defined in Table XXIX for each helmeted case in GCS ranges.

TABLE XXX

HEAD GCS AND HELMET IMPACT ATTENUATION

Crash Situation	GCS and Impact Acceleration (g) 15-12 Minor	11-8 Moderate	7-3 Severe
Collision	340g		
	340g		
	280g		
	180g		
	180g		
	180g		
	180g		
	180g		
	180g		
	180g		
	180g		340g
Other crash (fall etc.)	470g		
	340g		
	340g		
	180g		
	180g	470g	

In the above Table only two helmets were in the moderate and severe GCS ranges; these had an impact acceleration (refer Table XXIX) about an order of magnitude higher than the majority of helmets in the minor GCS range.

For the collision with another vehicle the speed of the other vehicle and the type of impact will also affect the outcome. Table XXXI indicates generally the effect of speed but more specifically Table XXXII outlines, for other vehicle involvement, the speed and collision characteristics for the main helmet types. It does not however indicate any trend.

TABLE XXXI

AIS VERSUS COLLISION SPEED

Speed Range	Total	AIS 0	1	2	3	4	5	6
< 50 kmlh helmeted crash	7	4	-	2	-	-	1	-
< 50 km/h unhelmeted crash	42	4	10	14	10	3	1	-
50 km/h and greater helmeted crash	5	1	-	4	-	-	-	-
50 kmlh and greater unhelmeted crash	18	1	1	6	3	3	3	3

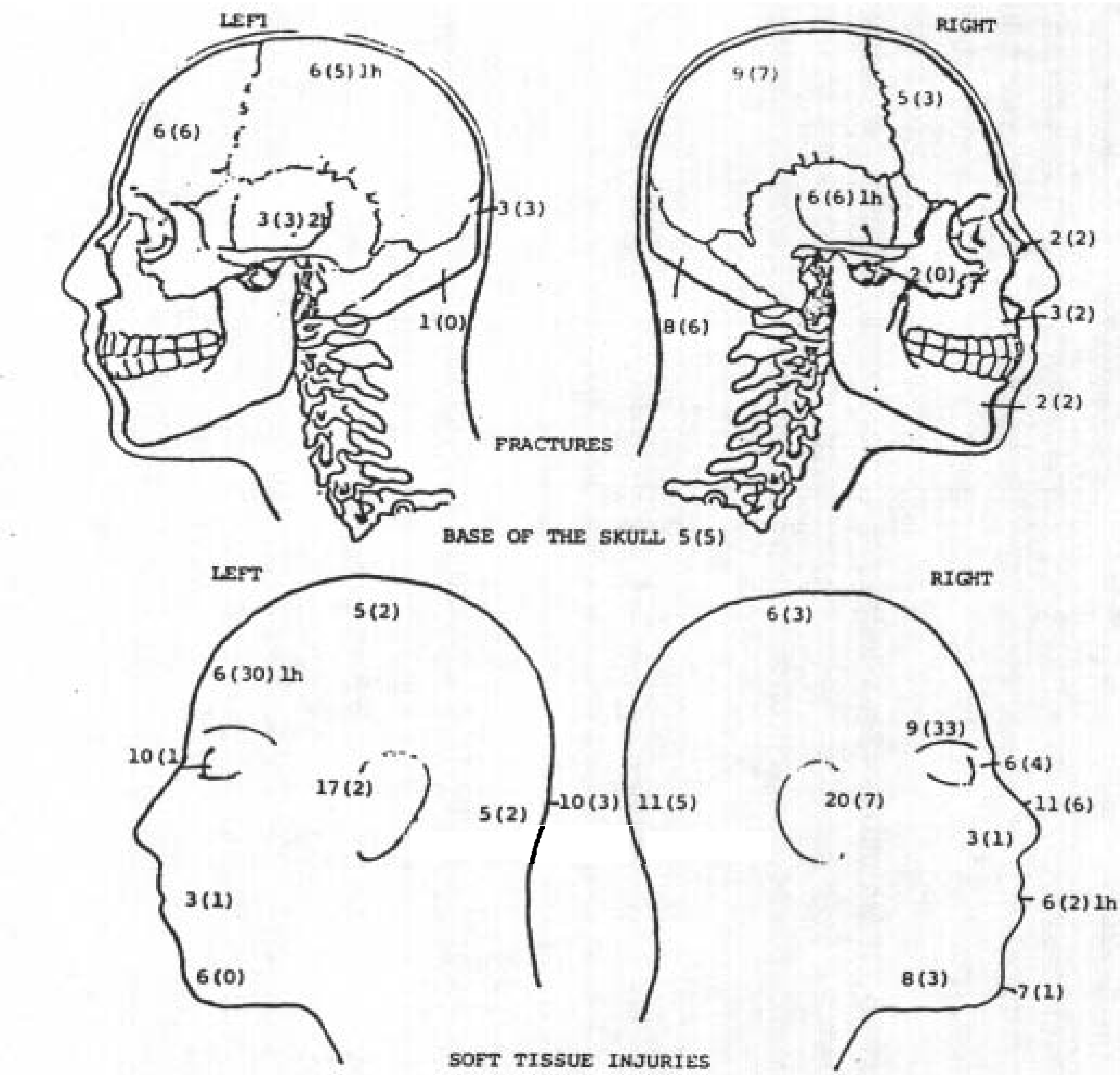
TABLE XXXII

HELMET IMPACT ATTENUATION VERSUS SPEED AND COLLISION TYPE

GCS Range (g)	Standard Impact Attenuation	Speed Range (Km/hr)	Type of Crash
Minor	180	< 50	Rode out in front of other vehicle
	180	< 50	Rode out in front of other vehicle
	180	< 50	Glancing impact
	180	< 50	Rode into side of other vehicle
	180	50	Rode out in front of other vehicle
	180	50	Rode out in front of other vehicle
	340	50	Glancing impact
	180	50	Rode into side of other vehicle
	180	50	Rode out in front of vehicle
	280	50	Rode out in front of other vehicle
	340	50	Glancing impact
Severe	340	50	Glancing impact

Table XXXI indicates that for increased speed there is a substantial increase in serious injury (AIS $\geq$ 3) for unhelmeted bicyclists. For the speed range less than 50 kmlh 33% had AIS $\geq$ 3 compared with 55.5% for 50 km/h and greater.

The location of injury is important for selection of helmet protection areas. Figure 18 indicates the impact points for helmeted and no helmet bicyclist crashes involving significant head injury.



Note: Numbering indicates ALL bicycle accidents where AIS scores were recorded in each location followed by the number of accidents recording AIS >=3 in brackets, and when a helmet is worn for one or more of these impact locations at AIS >= 3, the number of helmeted bicyclists follows as, e.g. 1h.

FIGURE 18 BICYCLIST HEAD INJURY IMPACT LOCATIONS

Figure 18 indicates a high proportion of impact points nprotected by the current extent of bicycle helmet protection (refer dotted line on figure 1). About 57% of the impact points re outside the helmet protection area. In bicycle crashes a igh energy impact to the facial region is less likely than in otorcycle crashes but there were 2 significant cases in the urvey where the main impact was to the jaw or facial area and here was associated brain damage at AIS 3 or greater.

It is interesting to examine depressed fractures of the skull ince one of the main benefits of a helmet is to spread the mpact and reduce the likelihood of this type of injury. There ere no depressed fractures for the helmeted cases while there ere 5 for the unhelmeted cases. Two of these were non fatal and nvolved a fall to the ground, and it is quite apparent that a elmet would have improved the outcome. The other three were

Neck injury was significant in 4 cases, one involving a elmet. In the helmeted case the main impact was to the facial rea and there was a partial crush of one cervical vertebra. In he other 3 cases no helmet was worn. All were fatal crashes nvolving another vehicle. One was high speed and unsurvivable, nd another case involved a complete dislocation of the neck at 1, presumably caused by hyperextension of the neck in a very iolent impact.

In both the helmeted and unhelmeted cases there are no crashes that involved two substantial impacts at the same location during the crash, indicating no need for a helmet to be designed to be capable of withstanding two major impacts at the same location.

EVALUATION OF SERIOUS BICYCLIST CRASH CASES

It is evident in a substantial proportion of bicyclist crashes involving minor head injury, as defined by a Glascow Coma Scale between 15 and 12, that use of a protective helmet can still fail to prevent minor head injury. In Table XXXIII there are 85% of non helmeted cases with minor head injury and 88% of helmeted cases with the same injury level.

It must be emphasised that minor head injury can be a source of on-going mild disability even though it may appear to be trivial (Rimel et al. 1981). The difference in minor outcome is of course better than the above differences between helmeted and unhelmeted riders. The proportion of helmeted bicyclists admitted to hospital with head injury could be up to 40% lower than the unhelmeted population of bicycle riders admitted to hospital.

Any type of bicycle helmet provides some protective effect but as Table XXXIII tends to indicate there is a risk of moderate and severe head injury for lightweight helmets. There are 10 cases involving minor head injury with helmets to the Australian Standard and 6 cases with other than Australian Standard wearing head protection devices. All crashes have some peculiar and unique characteristics so it is difficult to be completely definitive.

Referring to Table XXVIII, there are altogether 17 severe unhelmeted cases, of which 9 were fatal. The following Table summarises each case and includes an evaluation of the likely change in outcome if a protective helmet or improved protective helmet had been worn, based on a consideration of the location of injury and comparisons between helmeted and unhelmeted crash situations.

TABLE XXXIII

CHANGE IN HEAD INJURY WITH USE OF HEAD PROTECTION

Case No.	Morbidity	ISS	AIS Head	GCS	Evaluation
67	Fatal	25	5	7	A helmet would have improved outcome
82	Fatal	33	5	-	Possibility of high rotational acceleration
105	Fatal	22	3	3	Head injury caused by impacts to lower face also other severe body injury
111	Not Fatal	16	3	7	A helmet would have improved outcome
159	Fatal	18	4	-	Possibility of hiqh rotational acceleration
160	Fatal	22	3	-	A helmet would not have changed outcome
161	Fatal	36	4	-	A helmet would not have changed outcome
173	Not Fatal	17	4	7	A helmet would have improved outcome
181	Fatal	45	6	-	Possibility of hiqh rotational acceleration
183	Not Fatal	17	3	7	A helmet would have improved outcome
185	Not Fatal	22	3	7	A helmet would have improved outcome
199	Not Fatal	26	5	4	A helmet would have improved outcome
200	Not Fatal	34	4	7	Possibility of hiqh rotational acceleration
206	Not Fatal	18	4	3	A helmet would have improved outcome
267	Fatal	44	6	-	Severe lower face injury a helmet would not have improved outcome
269	Not Fatal	22	1	7	A helmet would have improved outcome
301	Fatal	30	5	5	Possibility of high rotational acceleration
302	Fatal	33	5	-	A helmet would not have changed outcome
305	Fatal	56	4	-	Neck hyper extension - severe impact to lower face

The above table indicates that of the 19 serious crash cases a substantial helmet would have improved outcome in 8 and reduced the risk of morbidity for case number 67. In a further 5 cases, while a substantial helmet may have improved outcome because of the impact location and type of crash there would have been substantial rotational acceleration. In these cases, a helmet with improved capacity to mitigate the effects of rotational acceleration would have been beneficial. A further interesting finding is that three of the fatal cases' had the main head impact in the lower face area, indicating that some form of facial protection may reduce the risk of morbidity.

In summary then, from 171 hospitalised or fatal bicyclist crash cases, 3 were totally unsurvivable; of the remaining 167, 19 were serious and only one had a light protective helmet. Eight of these would have had an improved outcome if a substantial protective helmet had been worn, and if a protective device that reduced rotational acceleration was used by all riders a further 5 cases may have been less severe. If some form of facial cushioning was used a further 3 cases may have had reduced severity.

FINDINGS - BICYCLIST CRASHES

* Bicyclists wearing helmets appear to be under-represented amongst those with significant head injury and those requiring admission to a neurosurgical unit.

* There is a higher incidence of serious injury associated with impacts of bicyclists with other moving vehicles than with falls from bicycles and impacts with fixed objects.

* Some bicyclist crashes are totally unsurvivable irrespective of the type or quality of head protection.

Primary and secondary school children make up 77% of head injury cases. Males are over-represented presumably because injuries sustained by adults can be less recoverable there is a higher proportion of adults with serious head injury than youths.

Where used, existing head protection consists of a selection of protective devices with a range of acceleration attenuation from 470g (low protection) to 180g (reasonable protection) as measured by a WSUlHodgson headform.

For the more serious crash situation involving collision between the bicyclist and another moving vehicle, existing helmet protection reduces serious injury from about 42% of crashes where no protection is used to about 15% where some protection (low protection) is used.

The speed of the other vehicle affects outcome: unhelmeted crashes indicate serious head injury in about 33% of cases below 50 km/h and about 55% around 50 kmlh. Usually by 80 kmlh the crash is unsurvivable.

While the helmeted crash sample is small, there is some evidence that the more serious head injuries are associated with helmets offering lower protection.

A high proportion of significant head injury produced by impacts to unprotected facial, lower temporal and lower occipital regions are unprotected by existing helmet coverage. Over half of the significant impacts are outside the normal helmet protection area. In 2 crash cases there was an impact to the unprotected facial area sufficient to cause serious brain damage. It is considered that a soft lower facial coverage could have reduced the severity of some of these crashes.

- Depressed fracture was only found in unhelmeted cases and no sharp penetrating injuries were identified. No cases were found of two significant impacts occurring in the same location.

- Out of 196 crash cases, 2 were identified as having significant ongoing disability as a result of the crash. In neither of these was a helmet worn.

Neck injury occurred in 4 cases and was fatal in 3 cases.

- It is estimated that 40% of serious cases would have had a substantial reduction in head injury if a good energy attenuation helmet had been worn.

- One helmet moved on impact causing a significant head injury. Two came off after the initial impact but with unaltered outcome. One helmet was unfastened and did not provide protection to the wearer.

SKULL HARD TISSUE EXPERIMENTS

HARD TISSUE TEST METHOD

The objective was to determine skull hard tissue bending properties and examine differences between adult and child skull hard tissue. Small skull hard tissue samples were taken from accident victims and tested to failure in bending. Normally the samples were tested within 2 to 3 days of cessation of life and were kept in refrigeration until the time of testing. The results for adults fall into the ranges for strength given by McElhaney (1979) and Wood (1979).

The skull is a sandwich construction with an inner and outer layer of compact cranial bone separated by the soft spongy diploe layer. Testing of these individual components was performed by McElhaney et al. (1979). These results showed that human skull bone was strain rate sensitive. The shear properties of diploe material are also given by McElhaney.

Testing of skull specimens in bending was performed to determine the variation in properties derived by McElhaney and the mode of failure of skull elements when acting as a composite. Specimens for testing were rectangular and were taken from the temporal, frontal, parietal and the occipital bones. Samples were drawn from these areas as they are least curved and were most easily loaded as simply supported beams. Samples were approximately 35mm by 10mm.

The samples were supported on flexibly mounted wedges. The flexibility of the supports was incorporated to allow for local curvature and discontinuites of the sample.

Prior to loading, a strain gauge was attached to the outer surface in a slightly off-centre position. The off-centre location was chosen so as to minimise any local stress raising effect. The gauge surface was prepared by scraping off the connective tissue and then lightly sanding the bone. Gauges were attached to the outer surface as it is much smoother. Gauges used were TML PL-3 and the cement used was TML Cyanoacrylate (CN) strain gauge adhesive.

RESULTS - SKULL HARD TISSUE EXPERIMENTS

The modulus and bending strength of the skull hard tissue was found to vary depending on the location of the sample in the skull. This variation is attributed to differences in the contruction of the hard tissue at different locations. The hard tissue is a sandwich construction consisting of an inner and outer layer of hard material separated by a soft spongy diploe layer. As would be expected, there was no significant variation in the modulus between children and adults. The adult skull is simply thicker and the sandwich outer and inner hard material layers are also thicker.

The results presented in Table XXXV indicate a substantial difference in failure strain in the child samples compared with the adult samples and the difference is particularly marked in the temporal region. In simple terms, under a similar applied load the child head will deform substantially more than the adult head.

TABLE XXXIV

ADULT *SKULL* HARD TISSUE RESULTS

Sample	Moment of Inertia (mm^3)	Thickness (mm)	Failure Strain ($\times 10^{-6}$)
Temporal	97	5.1	1700
Frontal	282	7.1	1090
Occipital	335	8.2	1060
Parietal	197	6.4	1040

TABLE XXXV

CHILD SKULL HARD TISSUE RESULTS

Sample	Moment of Inertia (mm^3)	Thickness (mm)	Failure Strain ($x\ 10^{-6}$)
Temporal	12	2.2	8800
Frontal	95	4.9	1700
Parietal	73	4.4	2400

The above results are based on a small number of samples .but they do illustrate the quite significant differences in bending strength at different localities in the skull hard tissue and between adults and children.

In the adult results, the reduced bending strength in the temporal region is quite apparent, with an available moment of inertia less than half that of the parietal region. The whole head, of course, acts together and there is natural padding afforded by the considerable outer layer of soft tissue over the skull. Nonetheless, an impact to the temporal region will need a greater amount of load spreading and impact padding than a similar blow to other areas of the skull.

The child skull hard tissue results are quite significant. The temporal area especially is obviously a zone of great vulnerability to impact, but it is alarming that current protective helmet standards give no consideration to the weak temporal region. There is no difference in protective helmet requirements between children and adults. The same impact attenuation test applies using a rigid metal headform to represent the head.

It is apparent from the above Tables that the child head is significantly weaker in bending than the adult head and will deform significantly more at failure (fracture). Medical practitiioners are aware that child head injuries have to be given careful consideration when presented for treatment. This, of course, is the reason so many children with mild symptoms of head injury are admitted for observation. With the greater flexibility in child head hard tissue there is a great likelihood of intracranial damage, brain laceration and vein rupture.

FINDINGS - SKULL HARD TISSUE EXPERIMENTS

The temporal area of the adult head is a zone of weakness, having between a third and a half the bending resistance of other areas of the skull hard tissue.

Children's heads exhibit even greater vulnerability in the soft temporal area. There is almost an order of magnitude less in bending resistance in the temporal region compared with other zones of the child skull hard tissue.

The child head in overall terms is considerably more flexible than the adult head, almost an order of magnitude in some zones. This greater flexibility means that the natural protective hard case (skull) is far less protective of the brain and the child has greater vulnerability to brain laceration and vein rupture caused by higher levels of head deformation during impact.

It is apparent that protective helmets for adult heads must provide significant additional protection in the vulnerable temporal region.

There are no separate requirements in the Australian Bicycle Helmet Standard for the design of children's protective helmets. The same impact attenuation test is used for the adult head, yet the difference in flexibility of the child head compared with the adult head is quite marked.

JAW IMPACT EXPERIMENTS

JAW IMPACT TESTS

The objectives of the experimentation were to gain a better understanding of head and neck dynamics for an impact to the jaw area, and to consider the sequelae of the impact in transmission of an impact pulse to the intracranial space and rotational acceleration of the head and to examine the effects of helmet mass and helmet facial bars.

CADAVER

For this experiment embalmed cadaver material was used. The method of preparing this material for the non-destructive impact tests is described in Appendix C.

The experimental techniques on cadaver material were worked up by first replicating some earlier work carried out by Hickey (1967) related to jaw impact of cadaver material. This work involved impacts to the jaw area with the head of the cadaver material restrained and the monitoring of intracranial pressure and temporal strain. The results obtained were similar to Hickey's results and are separately reported in Chapman, Corner, Whitney, Morgan, Parker (1986) and demonstrate that the intracranial pressurisation techniques used on the cadaver material produced similar results to other researchers.

The next step was to provide a reasonable amount of headlneck dynamic movement in the cadaver material by selective loosening of the neck muscles (Chapman. 1985). In addition, a wrestler's head harness was firmly strapped around the head of the cadaver material and elastic strappings were applied to the front and back of the head harness to provide an approximate standardised level of muscle flexibility (tone).

It was not intended that the cadaver material headlneck model provide in absolute terms the dynamic response of an in vitro human, but rather to provide a reasonable approximation from which reliable comparative results could be drawn. An additional advantage of the cadaver material model is that by use of dynamic pressure transducers connected to the intracranial space, the model can provide comparative intracranial pressure pulses for different impact conditions.

Two types of impact tests were carried out on the cadaver material: impacts to the jaw area with and without head protection. For each impact type the mass of the impacting pendulum mass was varied, refer Appendix C.

Before impact tests were carried out the intracranial space was filled with normal saline which has a similar specific gravity to intracranial fluid. This procedure was accomplished by making holes at a number of locations in the skull. Each of these holes were threaded and later capped with a brass fitting. The head was filled from the bottom and each hole was capped once saline began to flow out through it. During each impact a 300mm head of fluid was maintained in a tube above the external auditory meatus. Such a pressure head approximately corresponds to the in vitro intracranial pressure.

Head protection involved fitting to the head of the cadaver a full face and an open face helmet. Both helmets had holes cut into the crown of the shell to allow a pressure transducer to be fixed to the parietal region of the skull. Also, holes were cut into both sides and into the frontal region in the shell of the helmet to accommodate and allow easy access to the brass fittings already protruding from the skull of the cadaver material.

A pressure transducer was screwed into a tapped opening in the parietal region of the skull to measure the intracranial pressure upon impact. An accelerometer located at the rear of the pendulum mass measured the acceleration upon impact between the falling pendulum mass and the jaw of the cadaver material.

To monitor rotational acceleration, an accelerometer was fixed to the occipital region of the head harness for the unhelmeted impact and to the back of the helmet for helmeted impact. In both cases the accelerometer measured the tangential acceleration of the head upon impact thereby allowing rotational acceleration to be calculated.

Also, a displacement transducer was positioned at the back of the head to indicate the displacement of the head upon impact. A description of the instrumentation used can be found in Appendix C.

VOLUNTEER

A series of three impact experiments were carried out with a boxer as a volunteer. The first experiment involved pendulum impacts to the symphysis of the mandible where the volunteer had no head protection. The second involved wearing a boxing type head and jaw protection (generally used in sparring bouts), and the third involved the boxer volunteer wearing motorcycle helmets of three types; a full face helmet, with a flexible facebar, a heavier full face helmet with a stiff facebar and an open face helmet.

In all cases, the tangential acceleration of the back of the head upon impact was measured by means of a piezoelectric accelerometer fixed at the rear of the head by a band of elastic material. The band with the accelerometer attached was placed on the volunteer's head and positioned so that the accelerometer was against the occipital region of the head. For the three types of motorcycle helmets, the inner liner and comfort liner at the rear of the helmet were cut out to allow the helmet to be positioned over the accelerometer attached to the band around the head.

By recording the tangential acceleration, the rotational acceleration of head upon impact was able to be calculated. Also, an accelerometer located at the rear of the pendulum mass measured the acceleration upon impact between the falling mass and the symphysis of the mandible or facebar belonging to a helmet. The head movement after impact was also measured by a displacement transducer positioned against the back of the head or head protection in the occipital region. Details of the above instrumentation can be found in Appendix C.

GM HYBRID II HEAD AND NECK

Impact tests were also carried out on the jaw of the GM Hybrid II head and neck dummy, so that its displacement and rotational dynamics can be compared with the volunteer and cadaver material.

In this case the experiment involved the pendulum mass inpacting against the jaw of the Hybrid II head and against the jaw piece of a full face helmet attached to the headform.

The accelerometer measuring transverse acceleration was attached to a bracket which in turn was screwed to the back of the helmet in the occipital region.

RESULTS- JAW IMPACT EXPERIMENTS

The initial work carried out with a restrained head cadaver material head to replicate Hickey's work and demonstrated some interesting results. As the impact force is increased the acceleration pulse on impact reduces, indicating that as greater impact force is applied there is a greater amount of elastic deformation of the lower jaw.

While this experimentation was only undertaken to ensure that the intracranial pressurisation of the cadaver material was effective, the results demonstrate that the hard tissue of the facial area deforms on impact. The facial hard tissue structure provides a cushioning of the intracranial space except for upwards blows to the lower jaw, where impact force can be transmitted directly to the base of the skull via the temporo-mandibular joints.

The results of the series of impact experiments carried out on the cadaver material head/neck system are given in Tables XXXVI and XXXVII. Table XXXVI has results for the lower impact inertias level of 4 kg m^{-2} and Table XXXVII the results for the higher impact inertia level of 6 kgm^{-2}.

TABLE XXXVI

CADAVER MATERIAL JAW IMPACT RESULTS - LOWER IMPACT LEVEL

CASE	PENDULUM IMPACT ACCEL. (g)	ROTATIONAL ACCEL. OF THE (rad s^{-2})	DISPLACEMENT (mm)	INTRACRANIAL PRESSURE (kPa)
No head Protection	13.6	620	11.5	4.0
1500gm full (face helmet with a flexible facebar	12.0	2000	12.0	5.5
1100gm open face helmet	19.5	1320	9.0	7.0

TABLE XXXVII

CADAVER MATERIAL JAW IMPACT RESULTS - HIGHER IMPACT LEVEL

CASE	PENDULUM IMPACT ACCEL. (g)	ROTATIONAL ACCEL. OF THE HEAD (tad s^{-2})	DISPLACEMENT (mm)	INTRACRANIAL PRESSURE (kPa)
No head Protection	12.6	1290	13.5	6.0
1500gm full face helmet with a flexible facebar	10.8	2250	17.0	6.7
1100gm open face helmet	15.0	1470	12.5	8.0

The above results again indicate how the jaw deforms and reduces the impact force. For the no head protection case at the lower impact level the impact acceleration is higher than for the same case with a higher impact level. The same result is apparent for the open face helmet case with a lower impact acceleration 15.0g for the higher impact level compared with 19.5g for the lower impact level. It is interesting that the full face helmet with a flexible facebar seems to deform and reduce the impact acceleration.

In Table XXXVI the effect of helmet mass on intracranial pressure pulse is apparent. With no head protection intracranial pressure pulse is 4.0 kPa compared to 7.0 kPa with an open face helmet. The open face helmet of course does not change the jaw impact, so in this case it simply adds mass to the head. An increase is also apparent for the higher impact level (Table XXXVII) but the difference is not as great. The full face helmet produces a higher level of intracranial pressure than the no head protection case but less than the open face helmet case.

The Tables also show an increase in rotational acceleration, as the mass of the helmet increases, of about 1.5 times going from the 1100gm helmet to the 1500gm helmet. The helmet mass is greater than that of the open face helmet and an intracranial pressure greater than produced for the lighter open face helmet would be expected. But the intracranial pressure is lower than for the open face helmet (5.5 kPa compared with 7.00 kPa (Table XXXVI)). The presence of the facebar has reduced the direct transfer of impact to the brain via the lower jaw.

The results of impacts to the volunteer boxer are summarised in Table XXXVIII. Only the lower level of impact was used.

TABLE XXXVII

VOLUNTEER JAW IMPACT RESULTS

CASE	PENDULUM IMPACT ACCELERATION (g)	ROTATIONAL ACCELERATION OF THE HEAD (rad s^{-2})	DISPLACEMENT (mm)
No protection	17	610	40
Boxer head and jaw protector, mass 450gm	14	420	36
Full face helmet stiff (fibreglass) facebar, mass 1700gm	22	240	35
Full face helmet flexible (polymer) facebar, mass 1500gm	12	320	30
Open face helmet mass, 1100gm	14	700	47

The results show that the no protection (no added mass) cases are quite comparable between volunteer and cadaver material headtneck model but the rotational accelerations for the volunteer are far laser for the added helmet cases, including the open face helmet where there is no jaw impact protection. The volunteer experimentation could be improved by the use of myography to record the muscular tension in the volunteer's neck muscles.

There is a significantly greater impact acceleration pulse for the full face helmet with the stiff face bar compared with the more flexible facebar. The flexible facebar helmet produces an impact acceleration lower than the no protection case, as was found with the cadaver material. It is apparent that the stiff facebar does not deform as much, and it would be very desirable to try the stiff full face helmet on the cadaver material head/neck model to monitor the intracranial pressure. This was tried but it was not possible to fit a full face helmet with a stiff facebar onto the cadaver material model. Further testing is required.

The results of jaw impacts to the GM Hybrid II head/neck model are summarised in Table XXXIX.

TABLE XXXIX

GM HYBRID II HEAD/NECK JAW IMPACT RESULTS

CASE	PENDULUM IMPACT ACCELERATION (g)	ROTATIONAL ACCELERATION OF THE HEAD (rad s^{-2})	DISPLACEMENT (mm)
No protection	29	1280	17.3
Full face helmet stiff facebar	31	2360	17.5

It is apparent that the jaw of the GM Hybrid II head is far less deformable than the volunteer's or the cadaver material model. The rotational acceleration result with the full face helmet is quite similar to that obtained in the cadaver material headfneck model experiments.

FINDINGS - JAW IMPACT EXPERIMENTS

Impact energy is absorbed by the jaw and facial hard tissue, but where impact occurs to the unprotected jaw intracranial pressure pulse increases at the same rate as increases in impact energy.

The effect of added helmet mass is to increase intracranial pressure and rotational acceleration. For a 1100gm open face helmet the increase in intracranial pressure is between 40% and 70%; rotational acceleration increases by about 13%.

- The full face helmet appears to partly offset the effect of the helmet's increased mass by the distribution of an impact pulse to the facebar partly away from the intracranial space. There is still an increase in intracranial pressure pulse of about 12% to 37% but this is not as great as for the open face helmet.

- The added mass of the full face helmet still produces increased rotational acceleration with an increase of 50% for an increase in helmet mass of 18%, although this has not been reflected in the volunteer experiments.

IMPACT PROPERTIES EXPERIMENTS

The objectives of the impact properties experiments were to explore the difference in impact attenuation by using a more human-like headform, and to examine the effect on impact attenuation of variations in protective helmet liner density. All current protective helmets have virtually identical liner densities.

STANDARD DROP TESTS

The current Australian Standard impact test for both motorcycle and bicycle helmets is to use an instrumented solid magnesium headform onto which the helmet is attached and dropped (through 1.8m for the motorcycle helmet and 1.5m for the bicycle helmet) onto both a flat steel anvil and a hemispherical anvil.

The accelerometer located at the centre of gravity of the headform detects the vertical deceleration upon impact. The allowable peak acceleration values are 300g for motorcycle helmets and 400g for bicycle helmets.

The protective helmets have to be capable of producing a maximum peak acceleration less than the above value for two drop impacts at the same location. The Australian Standards also require the protective helmet to withstand a penetration test. In this test a penetrator of 9kg mass with sharpened point is dropped from 3m for motorcycle helmets and lm for bicycle helmets. For the helmet to pass this test the point of the penetrator must not make contact with the magnesium headform on which the helmet is mounted under the penetrator drop zone.

HEADFORM PROPERTIES

The present Australian Standard impact test headform is made of magnesium alloy (k-lA). Previous research has indicted marked differences in impact acceleration when a more flexible, human-like headform is used compared with the solid magnesium headform. The Wayne State University Hodgson headform was acquired as it reasonably resembles the strength characteristics of the in vitro human head.

The WSUlHodgson headform is made of a self-skinning urethane foam (similar to human bone) moulded from impressions of cadaver bones. A rubber gel material is used to simulate the brain. The headform also *has* a solid silicon rubber neck, through which runs a flexible cable allowing attachment to a cross arm. A cavity is located in the skull which is easily accessible from beneath the mandible, allowing an accelerometer to be mounted at the centre of gravity. The headform has similar load/displacement characteristics to a human head. Considerable cadaver testing was carried out to refine the headform design.

COMPARISON OF HEADFORMS

The magnesium and WSUlHodgson headforms were adjusted to the same overall mass. Drop acceleration tests were carried out on several brands of motorcycle and bicycle helmets (including both fibreglass and polymer type helmets).

The impact locations for all test drops using both headforms were restricted to the crown of the helmet, since the mounting point of the accelerometer in the WSUlHodgson headform disallowed freedom of adjustment of impact location between helmeted headform and anvil.

The impact accelerations (g) for both the motorcycle and bicycle helmeted headforms are summarized in Tables XXXX and XXXXI.

TAELE XXXX

COMPARATIVE IMPACT TESTS USING WSU/HODGSON AND MAGNESIUM HEADFORMS - MOTORCYCLE HELMETS

HELMET TYPE	MASS (gm)	IMPACT ACCEL. WSU/HODGSON (g)	IMPACT ACCEL. MAGNESIUM (g)	% WSU > MAG
ABS	1215	170	167	2%
Polymer	1160	156	195	-20%
Polymer	1325	163	165	-1%
Polymer	1159	181	191	-6%
Fibreglass	1518	195	188	4%
Fibreglass	1626	155	189	-18%
Fibreglass	1405	153	181	-15%
Fibreglass	1504	294	230	27%
Fibreglass	1412	216	206	5%
Fibreglass	1353	216	204	6%

TABLE XXXXI

COMPARATIVE IMPACT TESTS USING WSU/HODGSON AND MAGNESIUM HEADFORMS - BICYCLE HELMETS

HELMET TYPE	MASS (gm)	IMPACTACCEL. WSUIHODGSON (g)	IMPACT ACCEL. MAGNESIUM (g)	% WSU > MAG
Hairnet	190	470	1000	-210%
Polymer Minimal Padding	300	340	900	-260%
Fibreglass Minimal Padding	400	280	800	-285%
Polymer to Aust. Standard	550	180	155	16%
BMW plastic shell thick sponge foam liner	740	180	160	13%

Note: The results in Tables XXXX and XXXXI are not directly comparable because of the lower *drop* height used for the bicycle helmets.

Table XXXXI containing the results of the comparative tests on bicycle helmets indicates some dramatic differences between the two headforms particularly for the lightweight protective helmets. The difference exhibited is the result of the magnesium headform virtually bottoming out onto the steel anvil while the WSUlHodgson headform tends to bend and reduce the impact acceleration. The magnesium results are unreal because the human head is not unyielding like a block of magnesium metal. The WSU/Hodgson headform has a more realistic response.

It is intriguing to note that for the Australian Standard bicycle helmet the WSUlHodgson headform has an impact acceleration of about 16% higher than the magnesium headform. In this case, the WSU/Hodgson headform fails to compress the helmet liner material to the same extent as the magnesium headform.

In Table XXXX the comparative tests for motorcycle helmets exhibit a scatter of differences between the two headforms, although for the polymer helmets there is a greater tendency for the WSU/Hodgson headform to have a lower impact acceleration result. Does this reflect a less stiff helmet shell?

There is obviously scope for considerably more extensive testing using the WSUlHodgson headform, including various other upper head locations. While the above results are flimsy, there is some evidence that the liner material may be too stiff and the polymer shells too flexible.

HELMET LINER EXPERIMENTS

A series of tests were carried out by dropping the WSU/Hodgson and magnesium headforms from a height of 1.8m onto layers of foam of various thicknesses and densities. *Two* types of foam were used in these tests: Isothane with a density of 32 kg/m^3 and Korthane with a density of 50 kg/m^3.

Two of the test acceleration drops that were conducted included a 17mm thick hardboard layer inserted between the combined layers of foam and a flat steel anvil. This was done to give the impacting surface some elasticity similar to a pavement surface. The results of the tests are summarized in Tables XXXXII and XXXXIII.

From Tables XXXXII and XXXXIII it is important to note that the impact attenuation results for the Hodgson/WSU headform are greater than these for the magnesium headform when dropped onto the combined layers of foam with and without the hardboard. When the headforms were dropped on single layers of foam only the reading for the high density foam (i.e. korthane with density of 50 kg/m³) yielded greater results for the magnesium headform.

It is also interesting to note that there is a 14% and 8% reduction in the impact attenuation for both the WSU/Hodgson and magnesium headforms respectively when comparing the results between the layers of foam with and without the flexible hardboard.

TABLE XXXXII

EFFECT ON IMPACT ACCELERATION OF VARIATIONS IN LINER FOAM DENSITY

LINER FOAM DENSITY (kg m⁻³)	LINER FOAM THICKNESS (mm)	IMPACT ACCEL. HODGSON/WSU (g)	IMPACT ACCEL. MAGNESIUM (g)
32	38	125	77
32	25	200	125
50	25	300	475

TABLE XXXXIII
EFFECT ON IMPACT ACCELERATION OF MULTIPLE
LAYERS OF LINER FOAM

COMBINATION	IMPACT ACCEL. HODGSON/WSU HEADFORM (g)	IMPACT ACCEL. MAGNESIUM HEADFORM (g)
12.5mm of 32 density plus 2mm of 50 density	140	109
12.5mm of 32 density plus 25mm of 50 density with 17mm hardboard under	119	104
25mm of 50 density plus 12.5mm of 32 density with 17mm hardboard under	122	100

In all but one case the WSUIHodgson headform produced a greater impact acceleration, presumably indicating that the WSUlHodgson headform is less capable of producing compression of the padding foam because of inbending of the headform. These series of tests involved simply dropping the headforms onto flat sheets of foam so, in the case of the flexible (human-like) WSUlHodgson headform, there is no surrounding helmet shell and liner present to constrain the headform from squashing and expanding sidewards. Just how important this possible mechanism of head distortion is is unknown and because of comfort padding it may be unlikely that it can occur in real crashes.

It is interesting to compare the 50 kg m^{-3} foam test with the motorcycle helmet results. The 25mm thickness used in this test is less than a typical motorcycle helmet liner by about 5mm but it is of a similar density. Hence, for the WSUlHodgson headform the fibreglass helmet results in Table XXXX compare reasonably well with the 50 kg m^{-3} foam test. But the magnesium headform results for the fibreglass helmets (Table XXXX) are similar in magnitude to the WSUIHodgson headform. This is not the case in Table XXXXII where the magnesium headform has a far higher acceleration (475g compared with 300g for the WSUlHodgson headform).

One reasonable explanation for this difference is the absence of the helmet shell in the tests reported in Table XXXXII. The result is higher without the shell using the magnesium headform because of the foam stiffness with the shell present. Deformation and crushing of the shell by the magnesium headform produces the lower acceleration values in Table XXXX. The WSUlHodgson headform does not markedly distort and crush the shell but probably inbends instead. Observations of helmets after testing often show shell damage with the magnesium headform but this damage is never present after testing with the WSUlHodgson headform.

In summary, it would seem that there is a tendency for the WSU/Hodgson headform to inbend or distort (like a real head) and because of this it is less likely to produce substantial distortion of the helmet liner and shell. Unfortunately, the inbending property of the WSUlHodgson headform and the stiffness of the magnesium headform usually produce similar results but for quite different reasons. The latter headform being unyielding crushes helmet liner and shell to produce reasonably low levels of impact acceleration. The WSU/Hodgson headform also produces low levels of impact acceleration, but where stiff shell and foam are present, by distortion or inbending of the headform. This distortion is not wanted (Viano, 1985 indicating that 1 to 2mm distortion is the threshold of intracranial damage) so the helmet liner needs to be less stiff. The shell should not be less stiff because of its necessary load spreading requirement.

FINDINGS - *IMPACT* PROPERTIES EXPERIMENTS

The WSU/Hodgson headform has impact characteristics similar to a human adult head and will inbend or distort on impact.

The WSU/Hodgson headform is less capable of compressing layers of stiff padding compared with the unyielding magnesium headform.

- Often the WSUlHodgson and magnesium headforms produce similar results but the impact mechanics in each case is believed to be quite different. "he magnesium headform tends to distort the hard shell of the helmet and stiff foam while the WSUlHodgson headform tends to do less of this but the headform deforms like a real head.

- It is quite undesirable in normal protectable crashes to have substantial head distortion (particularly in children).

- Liners should be less stiff or of a lower density. There are advantages in having a combination of lower and higher stiffness liner foams.

- More work is needed to monitor the distortion of the WSUlHodgson headform but a foam density of 30 kg m^{-3} may result in significantly less skull deformation.

CRASH SIMULATION EXPERIMENTS

EXPERIMENTAL DESIGN

The objective of the crash simulation experiments was to identify the impact forces present in typical modest crash situations including translational acceleration, rotational acceleration and any helmet gripping that may occur. To carry out the crash simulation experiments an acceleration sled was constructed. The development of this test Sled is described in Appendic C but briefly it consists of a motorcycle stabilised by outrigger wheels and accelerated to a speed of 45 km hr^{-1} by a falling weight.

At the end of the acceleration the motorcycle is stopped by an impact piston over a distance of approximately 0.5m. With the sudden deceleration of the motorcycle the dummy is catapulted off the motorcycle. Two situations were tested; impacts of the head of the dummy with the laboratory floor and impacts of the head of the dummy with various points on a car body. The dummy used for the simulations consisted of a GM Hybrid Head and Neck attached to a block of wood with the *wood* enclosed in a sand filled canvas bag with an all up mass of 57 kg.

Two series of crash simulations were carried out. The first involved monitoring the impact of the helmeted dummy with the floor of the laboratory during high speed photography (2000 frames per second). This was carried out principally to determine the amount of surface gripping that can occur on impact. In the second series of crash simulations, the head of the dummy was instrumented with two accelerometers, both in the vertical direction with one accelerometer at the centre of the head and one offset by 35mm so that rotational acceleration could be measured. Simulations were carried out consisting of impacts to the laboratory floor and various points on a motor vehicle body.

Because of the somewhat uncontrolled nature of the simulations, difficulty was experienced in obtaining impacts to the head. (The dummy did not always behave and often had a first impact with the torso area rather than the head). For this reason only a very limited set of high speed photography runs were carried out with the remainder of the simulations relying on the acceleration measurements described above.

IMPACT WITH PAVEMENT - HIGH SPEED CAMERA RESULTS

The following Table lists results interpreted from the high speed photography by using frame by frame measurement techniques.

TABLE XXXXIV

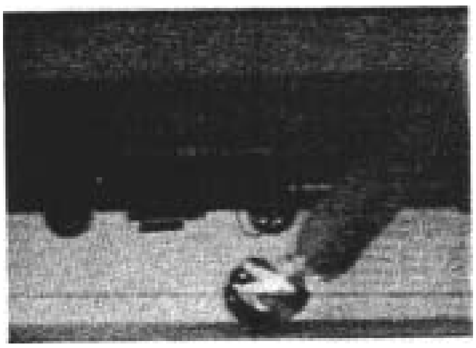
CRASH SIMULATION RESULTS IMPACTS WITH A CONCRETE SURFACE

HELMET TYPE	VERTICAL IMPACT ACCEL. (g)	HORIZONTAL. IMPACT SLIDING ACCELERATION (g)	ROTATIONAL. ACCELERATION rad s^{-2}
Polymer Brand No. 1 (1.1 kg)	250	310	37,000
Fibreglass Brand No. 1 (1.3 kg)	640	130	68,000

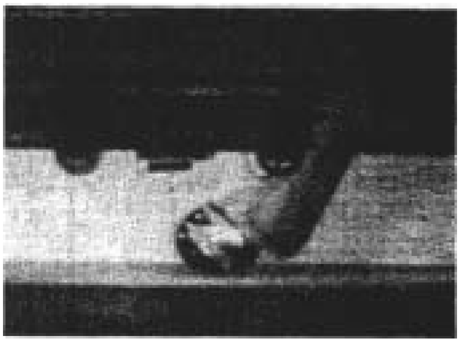
The above results indicate a substantial horizontal impact acceleration on impact. Where there is substantial forward velocity associated with a fall, the vertical impact produces a reactive horizontal frictional force which should be proportional to the coefficient of friction of the helmet material and the impact surface. Because of the magnitude of the vertical impact force, the horizontal impulsive frictional reaction will also be high. The above table demonstrates that high horizontal impact reactions occur in what is a relatively modest crash (45 km hr^{-1}). The above results are not necessarily directly comparable because it is very difficult to produce exactly the same crash simulation. The results do indicate a significantly lower horizontal sliding impact reaction for the fibreglass helmet which is probably the result of fibreglass being a harder lower frictional resistance material than polymer.

Measurements of the skid resistance of the concrete laboratory floor were taken with a standard skid resistance measurement device and compared with a typical asphalt pavement surface. The asphalt pavement surface was found to have approximately 1.5 times the skid resistance of the concrete floor. The horizontal sliding accelerations would increase by this proportion in an actual crash situation.

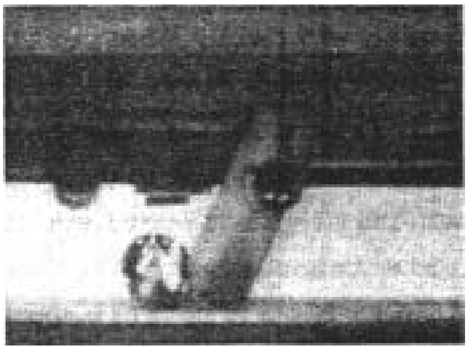
The high speed film shows a very rapid distortion of the neck followed by a rebound presumably as the elastic strain in the helmet shell, liner and neoprene neck releases. The following series of three photographs illustrate the stages of this rotation.



Exposure 1



Exposure 2



Exposure 3

PHOTOGRAPH 3 CRASH SIMULATION HIGH SPEED PHOTOGRAPHY OF AN IMPACT

IMPACT WITH PAVEMENT - USING AN INSTRUMENTED HEADFORM

The results in Table XXXXV were obtained from a series of crash simulations using both motorcycle and bicycle helmets.

TABLE XXXXV

CRASH SIMULATION RESULTS IMPACTS WITH A CONCRETE SURFACE

HELMET TYPE	BRAND NO.	MASS (kg)	ROTATIONAL ACCELERATION (rad s^{-2})
Full face polymer motorcycle helmet	1	1.1	46,000
Full face polymer motorcycle helmet	1	1.1	47,500
Full face fibreglass motorcycle helmet	1	1.3	98,700
Full face fibreglass motorcycle helmet	1	1.3	84,600
Full face fibreglass motorcycle helmet	1	1.3	125,400
Full face fibreglass motorcycle helmet	1	1.3	52.600
Full face fibreglass motorcycle helmet	2	1.7	43,000
Full face fibreglass motorcycle helmet	2	1.7	50.000
Polymer bicycle helmet	1	0.5	60,000
Polymer bicycle helmet	1	0.5	48,000
Polymer bicycle helmet	1	0.5	69,000
Polymer bicycle helmet	1	0.5	67,000
Polymer bicycle helmet	1	0.5	47,000

The rotational acceleration results in the above table tend to exhibit the same differences for different types of motorcycle helmet shell with polymer having an average rotational acceleration of 47,000 and fibreglass brand 1 having an average of 90,000. It is intriguing that fibreglass brand 2 has an average rotational acceleration of 46,000 which is comparable with the lighter Polymer helmet (1.1 kgs) yet this helmet has a mass of (1.7 kgs). Considerable further testing would be needed to properly define these trends. It is postulated that the brand 2 fibreglass helmet does not develop as much horizontal impact frictional reaction, and that this offsets the effect of increased mass on rotational acceleration.

The polymer bicycle helmets produced a higher average rotational acceleration (58,000) compared with the polymer motorcycle helmets (47,000), an increase of about 23%.

IMPACT WITH A MOTOR VEHICLE - USING INSTRUMENTED HEADFORM

In the group of crash simulation experiments involving the motor vehicle there were three impact points with the helmeted headform which were of importance: the side door panels, the door pillar and the front bonnet. The test results for the motor vehicle crash simulations with the instrumented headform are summarised in Table XXXXVI.

TABLE XXXXVI

CRASH SIMULATION RESULTS IMPACTS WITH A MOTOR VEHICLE

VEHICLE IMPACT LOCATION	HELMET TYPE	HELMET MASS (kgs)	BRAND NO.	ROTATIONAL ACCELERATION rad s^{-2}
Door pillar	Polymer full face motorcycle	1.1	1	101,800
Door pillar	Fibreglass full face motorcycle	1.3	2	91,250
Door panel	Polymer full face motorcycle	1.1	1	27,600
Door panel	Polymer full face motorcycle	1.3	2	40,800
Door panel	Fibreglass full face motorcycle	1.3	2	46,700
Bonnet	Fibreglass full face motorcycle	1.6	4	13,200
Bonnet	Fibreglass full face motorcycle	1.6	4	13,500

The door pillar collision is obviously a very lethal vehicle impact location, and both helmet types exhibit extreme levels of rotational acceleration. For the door panel case, it is of interest that the heavier fibreglass full face helmet again has a significantly higher rotational acceleration result. The bonnet crash simulations were glancing blows with the relatively soft bonnet of the vehicle and as the results indicate the rotational acceleration is relatively low.

FINDINGS - CRASH SIMULATION EXPERIMENTS

Impact of the helmeted head with a road surface where there is considerable pre-impact forward velocity produces a high horizontal frictional impact reaction as high as 450g. This tangential impact will produce high levels of rotational acceleration of the head. Fibreglass helmet$ appear to develop lower frictional impact forces.

Fibreglass helmets can develop higher levels of rotational acceleration possibly because of the greater mass of these helmets. However, one of the two brands tested goes against this trend, and while considerably heavier, has a rotational acceleration comparable to the lighter polymer helmet.

Polymer bicycle helmets produced higher levels of rotational acceleration than full face polymer motorcycle helmets, possibly because of the lighter construction and flexible structure of the bicycle helmet shell, or because of the mass of the bicycle helmet without the load distribution capability of the full face helmet.

Frontal impacts with the side panels of a motor vehicle also exhibit a lower level of rotational acceleration for polymer helmets.

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2

APPENDIX A - SURVEY DATA COLLECTION

Standardised forms were used for the crash site evaluation, medical evaluation and the helmet analysis. The forms are as follows:

ACCIDENT SITE FORM

MEDICAL REPORT FORM

HELMET ASSESSMENT FORM

The crash evaluation was further prepared for data processing using a data entry form.

The information contained on the forms was codified and keyed into the database fields indicated in Table A.I.

TABLE 1

Structure for database : C:helmet.dbf
Number of data records : 330
Date of last update : 09/19/86

Field	Field name
1	CASE-NO
2	BICYCLE
3	MOTORCYCLE
4	PILLION
5	DATE
6	TIME
7	MAKE-HODEL
8	CC
9	EST-SPEED
10	CYCLE-DIR
11	RIDER-DIR
12	MAKE_VE1
13	SPEED_VE1
14	VE1_DIR
15	MAKE_VE2
16	SPEED_VE2
17	VE2_DIR
18	IMPACT_VE1
19	BODY-VEl
20	BODY-CTCR
21	BODYSPEED
22	HEADFALL
23	BREAKFALL
24	HEADBLOW
25	OBJ_IMP
26	OTHER-OB3
27	TYPE-IMP
28	IMPACT-VEZ
29	BODYSPEED2
30	HEADFALL2
31	BREAKFALL2
32	OBJ_IMP2
33	TYPE_IMP2
34	S_COMMENT
35	HELMETWORN
36	HEL_MANUF
37	MODEL-SIZE
36	SAA_LABEL
39	HEL-TYPE
40	HEL_MASS
41	RET_SYSTEM
42	BIKE-RET
43	PRIOR_COND
44	RET-FAIL
45	RETAINED
46	OFF_IMPACT
47	RET_COND
48	FIT
49	SHELL-HAT
50	DESCRIP_SH
51	SH_THICK_C
52	SH_THICK_T
53	SH_THICK_W
54	SH_THICK_J
55	LINER-TYPE
56	DESCRIP_L
57	L_THICK_C
58	L_THICK_T
59	L_THICK_W
60	L_THICK_J
61	NUM_IMP
62	SL-IMP-DAM
63	SL_OBJ_TYP
64	SL_OBJ_MAT
65	SH_ABRAS
66	SH_PUNCT
67	SH_CRACK
68	SH_PROJ
69	L_CRUSH
70	L_DENSITY
71	L_AREA_CR
72	SL_DAM2
73	SL_OBJT2
74	SL_OBJM2
75	SH_ABRAS2
76	SH_PUNCT2
77	SH_CRACK2
78	SH_PROJ2
79	L_CRUSH2
80	L_DENS2
81	L_AREA_C2
82	H_COMMENT
83	DAYS_HOSP
84	FATAL
85	MALE
86	AGE
87	VICT_MASS
88	COMA-SCORE
89	SKULL-FRAC
90	SK_FRAC1
91	SK_FRAC2
92	SK_FRAC3
93	SK_FRAC4
94	SK_FRAC5
95	SK_FRAC6
96	SK_FRAC7
97	SK_FRAC8
98	SK_FRAC9
99	OTHER-SKFR
100	LEFORT
101	HI-COMMENT
102	COMP_CLES
103	VERT-LOC
104	CV_COMMENT
105	FAC_SOFTIS
106	LOCAT_STI1
107	LOCAT_STI2
108	LOCAT_STI3
109	LOCAT_STI4
110	LOCAT_STI5
111	LOCAT_STI6
112	LOCAT_STI7
113	LOCAT_STI8
114	FAC-EYE
115	FAC-TEETH
llb	FI_COMMENT
117	AIS-H
118	AISC_1
119	AIS-N
120	AIS-F
121	AIS-CH
122	AIS-AB
123	AIS_EM
124	AIS-EX
125	MAIS
126	ISS
127	LT_COMMENT
128	RECORDED

CONFIDENTIAL

CRASH NUMBER:

CRASH DATE/TIME:

POLICE ACCIDENT REPORT FORM NO.:

OFFICER ATTENDING: **STATION :**

CRASH LOCATION:

WEATHER CONDITIONS:

VEHICLES INVOLVED (CASE VEH. FIRST)

	Make/Model	Reg.No.	Disposal	Address	Examined
1					
2.					
3.					

PERSONS INVOLVED: M/C RIDER, PILLION

Name	Address	Police Injury Severity	Hospital/Ward

Prepare a scale drawing of the crash shoving approach paths of vehicles, final rest positions of vehicles, length of skids and direction, final rest position of motorcycle rider and pillion. Probable impact point. Indications of scuff marks, skin grazes, clothing marks etc.

Prepare a description of the crash eg., each step, each event.

Best estimate of vehicle speeds

Best estimate of body speed and impacts.

Approximate head fall height.

Fall broken by arm or shoulder.

Heed impacts *with* objects

HELMET (indicate type)

Retention railed prior to initial impact

Helmet retained?

Helmet came off after initial impact.

Indicate damage areas *on* helmet outlines attached.

VEHICLE DAMAGE

- Illustrate damage locations on attached motorcycle and car figures if difficult modify in red.
- Relate damage to likely body impacts and helmet impacts by making notes. Refer attached chart.

, Description of photographs. (Normally damage to vehicles should be photographed from above).

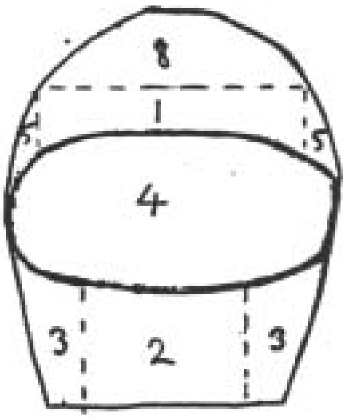
1

2.

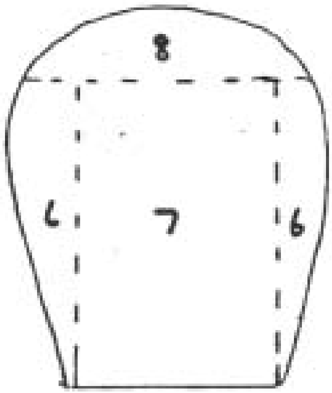
3.

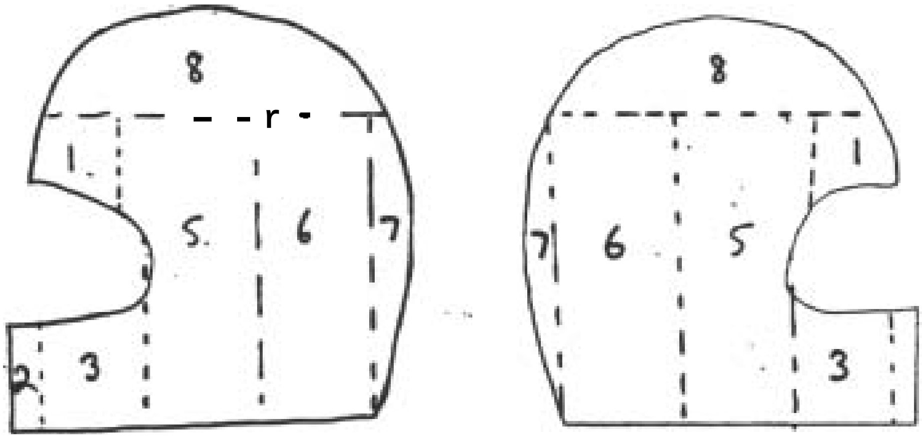
4.

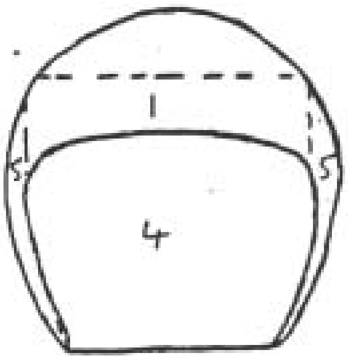
5.



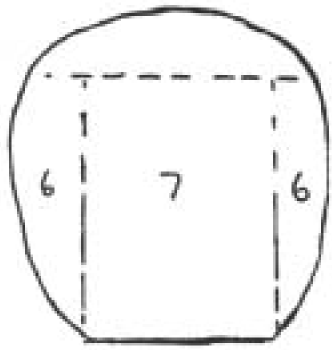
CONTACT CODE KEY
1 -Front
2 - Mid facebar
3 - Side facebar
4 - FACE only
5 - Temple
6 - Rear Side
7 - Rear
8 - Crown

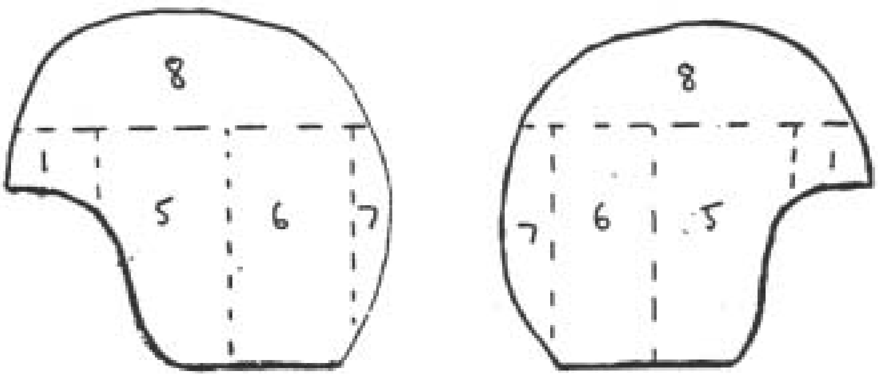


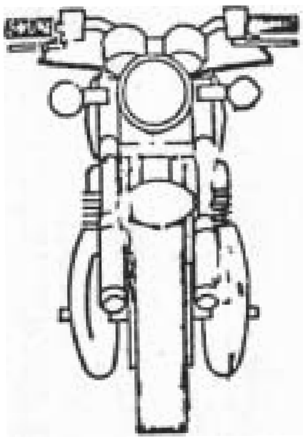


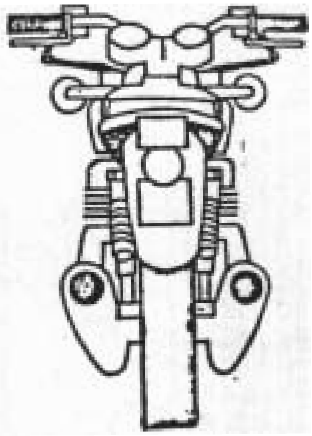


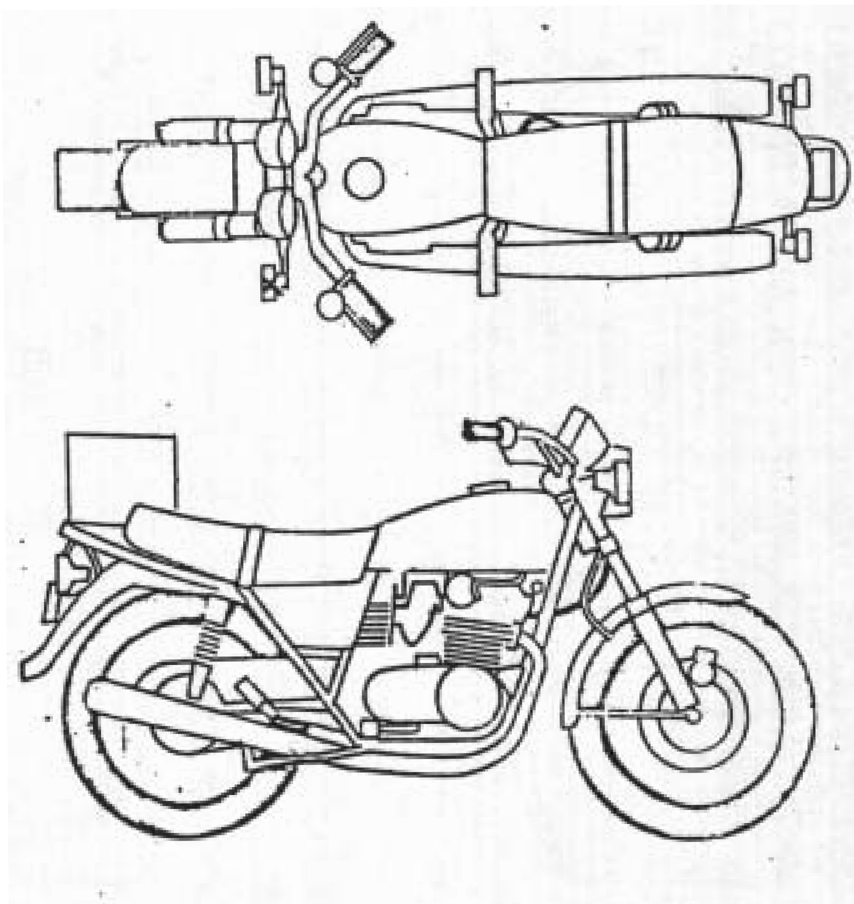
CONTACT CODE KEY
1 -Front
4 - FACE only
5 - Temple
6 - Rear Side
7 - Rear
8 - Crown

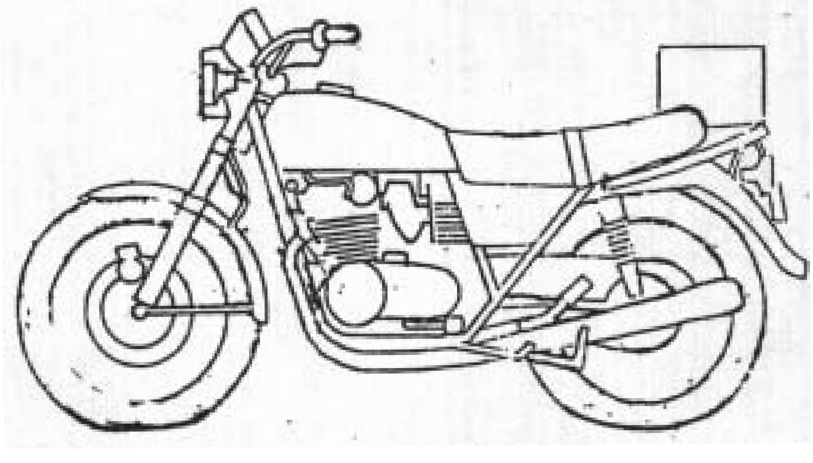










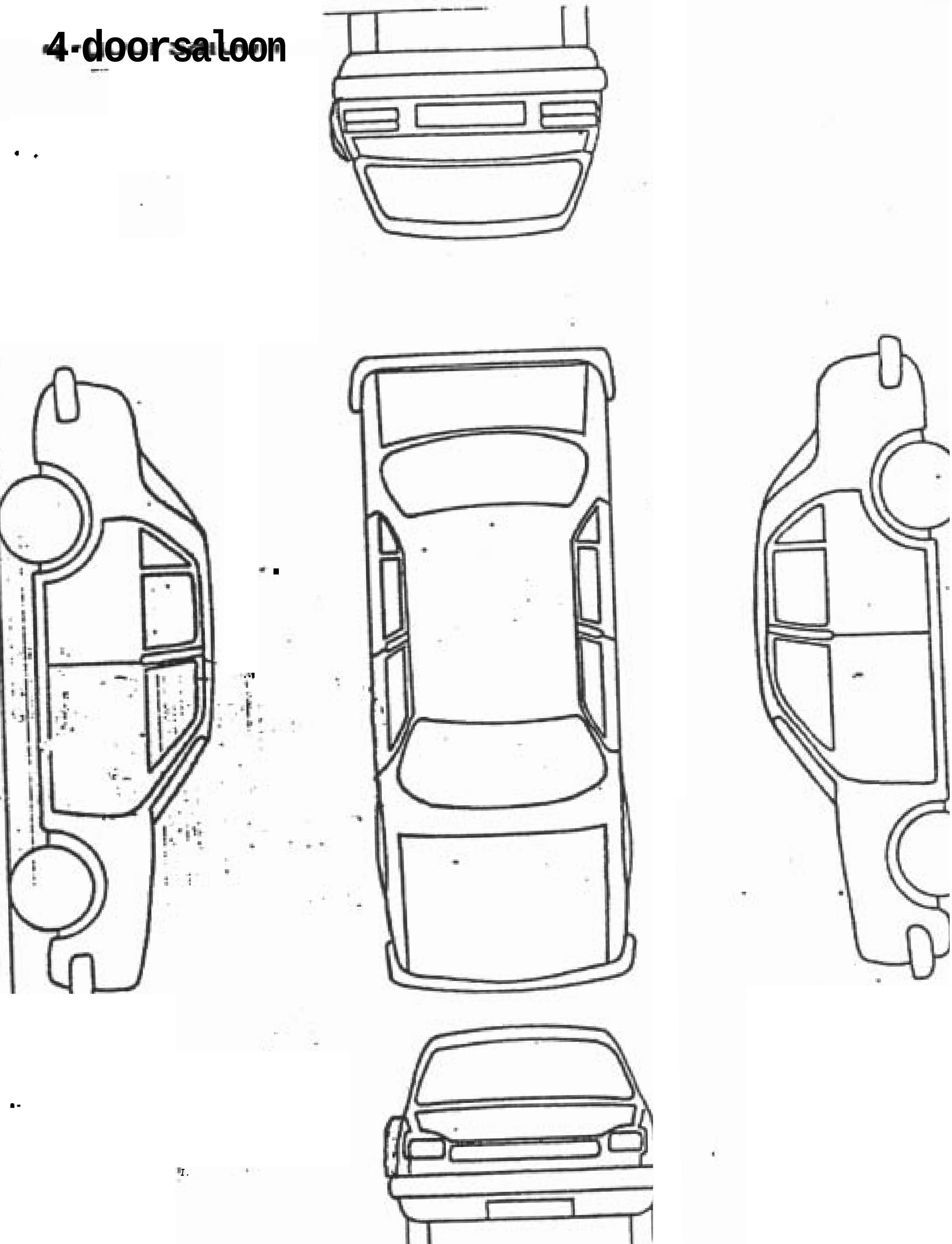


EXTERIOR DAMAGE Page 137

X	1	2	3	

& Model	Reg Year	Direct damage
meter	Body colour	Other damage

4-door saloon



POST CRASH ANALYSIS MEDICAL REPORT

DATE OF ADMISSION
DATE OF DISCHARGE — DAYS-HOSP
DATE OF DEATH — FATAL
VICTIM MASS — VICT-MASS
VICTIM SEX.
VICTIM HEIGHT
VICTIM HEAD CIRCUMFERENCE

HEAD INJURY

COMA SCORE — COMA SCORE
ARM AND LEG MOVEMENT
FOCAL NEUROLOGIC SIGNS
INTRACRANIAL PRESSURE COURSE
SKULL *FRACTURE* (refer diagram) — SKULL-FRAC

CT SCAN PATTERNS
CT SCAN BASAL *CISTERNS*
.AIS (head) — AIS.H
COMMENTS ON HEAD INJURY — HI-COMMENT (Memo)
LEVEL OF CONCIOUSNESS — AISC-I
COMMENT:

CERVICAL SPINE INJURY

NEUROLOGICAL DEFECT — transient or permanent
TYPE OF CORD LESION — complete — CORDLES-C
incomplete — CORDLES-I
VERTABRAE LOCATION (refer diagram) — VERT-LOC
AIS (CERVICAL SPINE) — AIS-C
COMMENTS ON CERVICAL SPINE INJURY — CV-COMMENT (Memo)

FACIAL INJURIES

SOFT TISSUE SEVERITY
(Location refer diagram)

Laceration
Graze
Skin Defect
Skin flaps

FACSOFTIS ☐

EYE INJURY — LEFT — RIGHT
blunt
penetrating
Visual acuity

FAC-EYE ☐☐

TEETH INJURY SEVERITX
chipped
lost
molocculsion

FAC-TEETH ☐

FACIAL FRACTURE UPPER — SEVERITY
(Location refer diagram)
frontal bone
frontal sinus
supraorbital ridge

FAC-FR-UP3 ☐

MDDLE FRACTURE SEVERITY
Nasal bone
Zygoma
Zygomotic arch
Orbital floor

Severity Lefort I
I1
III

LEFORT ☐

MANDIBLE FRACTURE

FAC-MAND ☐

FACIAL AIS

AIS-F ☐☐☐

COMMENTS ON FACIAL INJURY

FI-COMMENT (Memo)

OTHER INJURY AIS

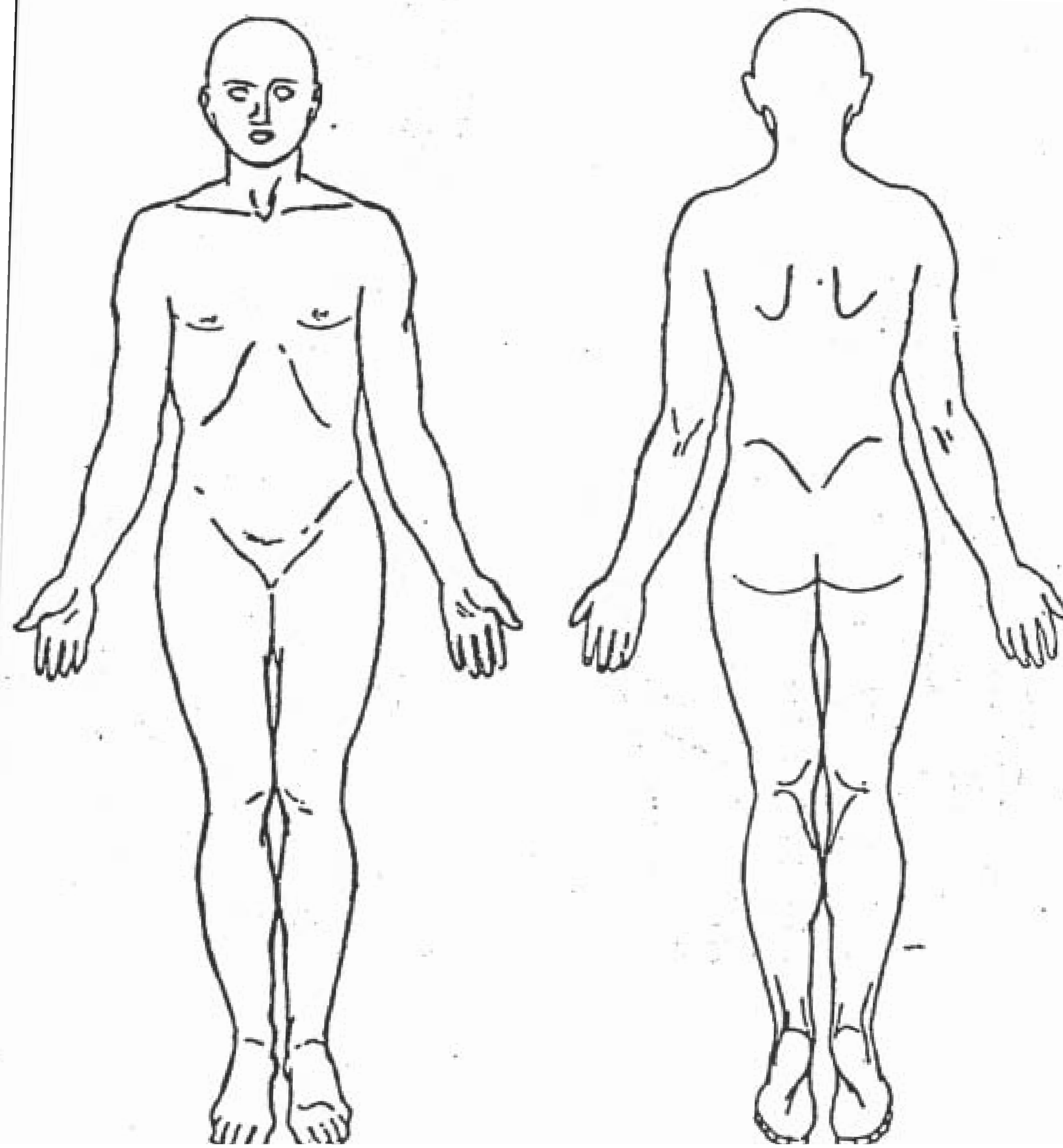
Injury	AIS	IN1-AIS	☐☐☐
Injury	AIS	IN2-AIS	☐☐☐
Injury	AIS	IN3-AIS	☐☐☐
MAIS		MAIS	☐☐☐☐
ISS		ISS	☐☐☐☐

LONG TERM OUTCOME COMMENTS

LT-COMMENT (Memo)

Accident No. [| | | | |]

Record all trunk and limb injuries. Record head injury details overleaf.



pital .... Admission Date .... Discharge date ....

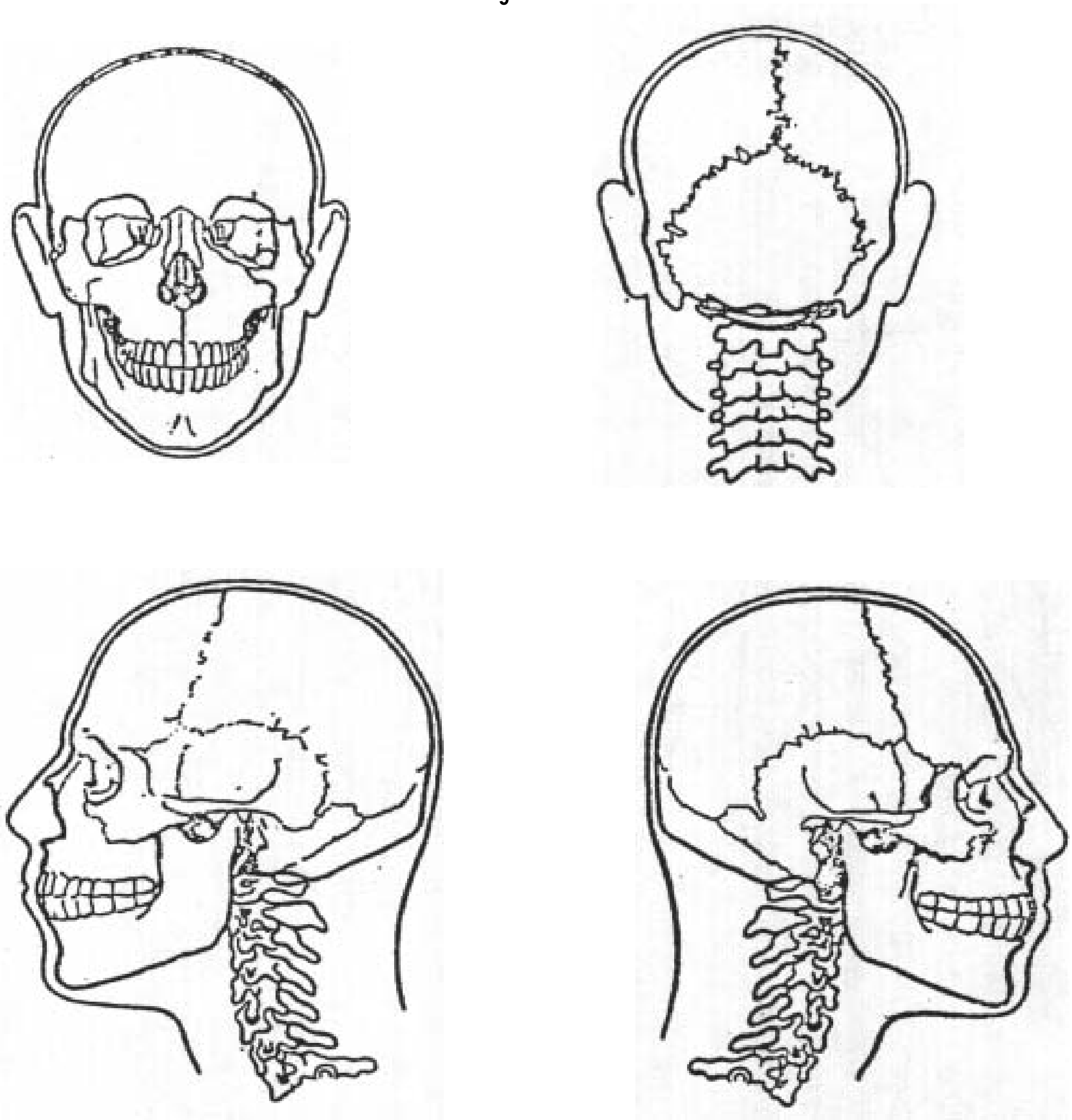
er/Pillion : Injured Fatal Age/Sex .... Height ....Wt/Bld ....

MET: F/Face O/Face Fibreglass Thermoplastic Retained in accident ....

Came off in Accident ....

ANY GIVEN DETAILS OF ACCIDENT

RECORD ALL HEAD INJURY DETAILS



PROTECTIVE CLOTHING

Trunk	Legs	Feet	Hands

POST CRASH MOTORCYCLE AND BICYCLE HELMET PROJECT

HELMET ASSESSMENT FORM

CRASH NUMBER: CRASH DATE/TIME:

CRASH LOCATION:

GENERAL HELMET PROPERTIES

HELMET MANUFACTURER		HEL-MANUF	
MODEL NUMBER /SIZE		MODEL-SIZE	☐☐☐
DOES THE HELMET HAVE AN SAA LABEL		SAA-LABEL	☐
HELMET TYPE: (Refer sketches attached)	1. basic 2. utility 3. jet 4. partial skirt 5. full skirt	HEL-TYPE	☐
HELMET MASS		HEL-MASS	☐☐☐
HELMET RETENTION SYSTEM	1. none 2. drings 3. Snaps 4. snaps & drings 5. quick release 6. other specify	RET-SYSTEM	☐
BICYCLE HELMET SUPPORTS	1. Two points 2. Four points 3. Four points +	BIKE-RET	☐
HELMET CONDITION PRIOR TO CRASH	1. good condition 2. worn 3. degradation evident 4. likely previous collision	PRIOR-COND	☐

HELMET RETENTION SYSTEM

FAILURE OF RETENTION SYSTEM

SUSPECT RETENTION SYSTEM FAILURE PRIOR TO INITIAL IMPACT	RET-FAIL	☐
HELMET RETAINED THROUGHOUT CRASH	RETAINED	☐
SUSPECT HELMET CAME OFF ATER INITIAL IMPACT (Note refer site report for answers)	OFF-IMPACT	☐

DAMAGE TO RETENTION SYSTEM	1. came off intact 2. broken at attachment 3. strap broken 4. fastener broken 5. major helmet failure	RET-COND	☐
HELMET FIT - good or poor (Refer site report, medical report may be obvious, information from victim or head measurement could be used)		FIT	☐

SHELL PROPERTY

SHELL MATERIAL

1. fibreglass
2. polycarbonate
3. other/specify , SHELL-MAT ☐

MEASURE SHELL THICKNESS AT *THE* FOLLOWING LOCATIONS:

Crown	mm	SH-THICK-C	☐☐☐
Temple	mm	SH-THICK-T	☐☐☐
Curtain	mm	SH-THICK-W	☐☐☐
Jaw	mm	SH-THICK-S	☐☐☐

LINER PROPERTY

TYPE OF LINER

1. none
2. Styrofoam large bead
3. Styrofoam small bead
4. polyurethene
5. ethafoam
6. neoprene *sponge*
7. polypropylene
8. other (specify) LINER-TYPE ☐

MEASURE TYPICAL LINER THICKNESS AT THE FOLLOWING LOCATIONS

Crown	mm	L-THICK-C	☐☐
Temple	mm	L-THICK-T	☐☐
Curtain	mm	L-THICK-W	☐☐
Jaw	mm	L-THICK-J	☐☐

IMPACT DAMAGE

NUMBER OF SIGNIFICANT IMPACTS

1. one
2. two
3. three or more NUM-IMP ☐

The following data should be recorded in order of impact i.e., first impact then subsequent impact. The location on the helmet will be determined by reference to the site and medical report. Example: Riders head impacts with side of car (impact 1). The rider then is thrown and hits the road pavement (impact 2).

IMPACT 1

LOCATION OF DAMAGE
(Refer attached diagrams showing helmet surface divided into 8 areas)

WHICH AREA — SL-IMP-DAM ☐

TYPE OF OBJECT STRUCK
1. flat
2. blunt edge
3. sharp edge
4. blunt object
5. sharp object

SL-OBJ-TYP ☐

TYPE OF MATERIAL
1. metal
2. glass
3. wood
4. soil
5. pavement

SL-OBJ-MAT ☐

DAMAGE TO SHELL IS NOW RECORDED

Abrasion depth	mm	SH-ABRAS	☐☐☐
Puncture depth	mm	SH-PUNCT	☐☐☐
Crack length	mm	SH-CRACK	☐☐

DAMAGE TO EXTERNAL PROJECTION
1. projection grazed
2. sheared off
3. evidence of catching of protection with surfaces

SH-PROJ ☐

MAXIMUM AMOUNT OF LINER CRUSH EVIDENT
mm
(Note liner normally at least partially recovers so quite small residual deformations can be evidence of a quite major deformation)

L-CRUSH ☐☐

TYPICAL MATERIAL DENSITY
(Normally found by comparison with known materials)

L-DENSITY ☐☐☐☐

AREA OF CRUSH SIGNATURE mm^2 — L-AREA-CR ☐☐☐☐

IMPACT 2

LOCATION OF DAMAGE
(Refer attached diagrams showing helmet surface divided into 8 areas)

WHICH AREA — SL-DAM2 [ ]

TYPE OF OBJECT STRUCK
1. flat
2. blunt edge
3. sharp edge
4. blunt object
5. sharp object

SL-OBJT2 [ ]

TYPE OF MATERIAL
1. metal
2. glass
3. wood
4. soil
5. pavement

SL-OBJM2 [ ]

DAMAGE TO SHELL IS NOW RECORDED

Abrasion depth	mm	SH ABRAS2	[ ][ ][ ]
Puncture depth	mm	SH-PUNCT2	[ ][ ][ ]
Crack length	mm	SH-CRACK2	[ ][ ]

DAMAGE TO EXTERNAL PROJECTION
1. projection grazed
2. sheared off
3. evidence of catching of protection with surfaces

SH-PROJ2 [ ]

MAXIMUM AMOUNT OF LINER CRUSH *EVIDENT*
mm
(Note liner normally at least partially recover so quite small residual deformations can be evidence of a quite major deformation)

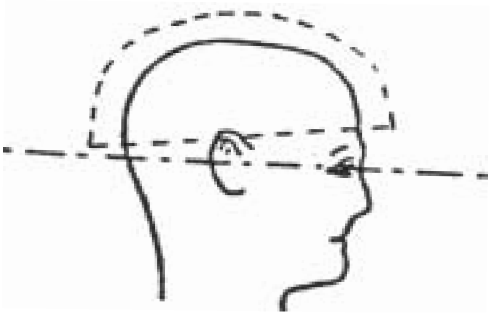
L-CRUSH2 [ ][ ]

TYPICAL MATERIAL DENSITY
(Normally found by comparison with known materials)

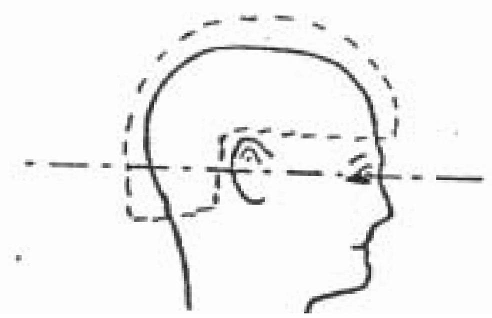
L-DENS2 [ ][ ][ ][ ]

AREA OF CRUSH SIGNATURE mm^2 — L-AREA-C2 [ ][ ][ ][ ]

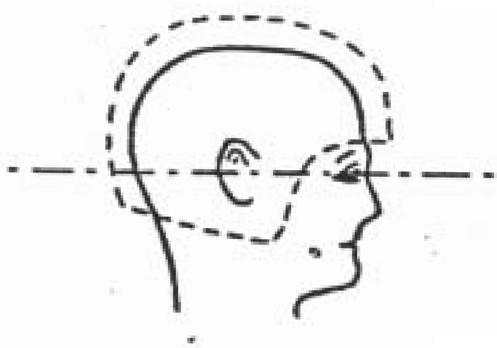
HELMET ADEQUACY COMMENT — H-COMMENT (Memo)



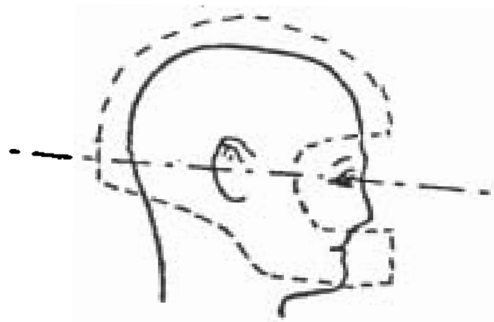
BASIC HELMET



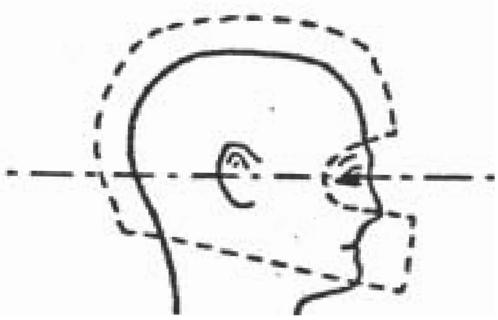
UTILITY HELMET



JET HELMET



PARTIAL SKIRT HELMET



FULL SKIRT HELMET

POST CRASH PROJECT DATA ENTRY

* *

CASE NUMBER — CASE-NO. ☐☐☐

BICYCLE OR MOTORCYCLE — BICYCLE ☐ MOTORCYCLE ☐

PILLION (Note: Only tick if this is the data entry for a pillion. Normally a pillion record should come after the riders record). — PILLION ☐

DATE — DATE ☐☐☐☐☐☐☐

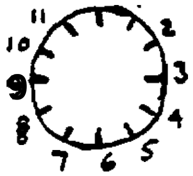
TIME (24 Hour Clock) — TIME ☐☐☐☐

MAKE AND MODEL OF MOTORCYCLE OR PUSH BIKE — MAKE-MODEL ________

ENGINE CAPACITY OF MOTORCYCLE — CC ☐☐☐☐

ESTIMATED SPEED OF MOTORCYCLE OR CYCLE JUST BEFOR IMPACT (m/s) — EST-SPEED ☐☐

DIRECTION OF BIKE AT IMPACT



CYCLE-DIR ☐☐

DIRECTION OF RIDER WHEN PARTING WITH BIKE — RIDER-DIR ☐☐

IF OTHER VEHICLES INVOLVED

MAKE Of FIRST VEHICLE		MAKE-VE1	________
SPEED OF FIRST VEHICLE	(m/s)	SPEED-VE1	☐☐
DIRECTION OF FIRST VEHICLE	(clock)	VE1-DIR	☐☐
MAKE OF SECOND VEHICLE	(m/s)	MAKE-VE2	________
SPEED OF SECOND VEHICLE	(m/s)	SPEED-VE2	☐☐
DIRECTION OF SECOND VEHICLE	(clock)	VE2-DIR	☐☐
IMPACT POINT WITH FIRST VEHICLE (select impact area no. from the attached diagram)		IMPACT-VEL	☐☐
SECTION OF BODY IMPACTING FIRST VEHICLE	1. head 2. upper torso 3. legs	BODY-VE1	☐

DESCRIBE GENERALLY HOW BODY BEHAVES
1. body caught by vehicle
2. thrown or bounced off vehicle
3. crushed by vehicle
BODY-CTCR ☐

IF THE BODY IS THROWN WHAT BODY SPEED — BODYSPEED ☐☐☐

APPROXIMATE HEAD FALL HEIGHT — HEADFALL ☐☐☐

WAS THE FALL BROKEN BY AN ARM OR SHOULDER — BREAKFALL ☐

DESCRIBE THE NATURE OF THE HEAD IMPACT WITH THE VEHICLE
1. transverse to flight
2. glancing
3. tangential
4. caught
HEADBLOW ☐

HEAD IMPACT WITH ANOTHER OBJECT?
1. pavement
2. kerb
3. tree
4. post
5. sign
6. guard fence
7. other/specify
OBJ-IMP ☐

DESCRIBE THE NATURE OF THE HEAD IMPACT WITH THE OTHER OBJECT
1. transverse to flight
2. glancing
3. tangential
4. caught
TYPE-IMP ☐

WAS THERE A HEAD IMPACT WITH A SECOND VEHICLE:
IF SO WHAT VEHICLE AREA (refer diagram) — IMPACT-VE2 ☐☐

IF NO *OTHER* VEHICLE INVOLVED

ESTIMATED BODY SPEED LEAVING BIKE (m/s) — BODYSPEED2 ☐☐

APPROXIMATE HEADFALL HEIGHT — HEADFALL2 ☐☐☐

WAS THE FALL BROKEN BY AN ARM OF SHOULDER — BREAKFALL2 ☐

HEAD IMPACT WITH AN OBJECT
1. pavement
2. kerb
3. tree
4. post
5. sigh
6. guard fence
7. other/specify
OBJ-IMP2 ☐

DESCRIBE THE NATURE OF THL HEAL, IMPACT WITH THE OBJECT

1. transverse to flight
2. glancing
3. tangential
4. caught

TYPE-IMP2 ☐

ADD A COMMENT ABOUT THE STAGES INVOLVED IN THE CRASH

S-COMMENT (Memo)

THE REMAINDER OF THE DATA ENTRY CAN BE FOUND ON THE HELMET REPORT AND MEDICAL REPORT FORMS

APPENDIX B - JAW *IMPACT EXPERIMENTAL* DESIGN

Impact was applied using a pendulum impactor especially designed for this type of work. Impacts were applied to both the jaw of cadaver material and a volunteer for in vitro comparison. Before impacts to the cadaver material were carried out, the neck was modified to provide response and the intracranial space was filled and pressurised and a pressure transducer fitted. For both the cadaver material and the volunteer impacts appropriate instrumentation was fitted to measure acceleration. The following sections describe the various experimental techniques used.

PENDULUM IMPACTOR

The pendulum impactor was supported by a tubular pyramid so that impacts could be delivered to the jaw of cadaver material in the reclining position or to a volunteer in a sitting position. Figure 19 provides a perspective sketch of the complete assembly and also provides details of the pendulum impactor. As illustrated, it consists of a heavy padded steel impactor pinned to a light aluminium shaft with an accelerometer fitted to the rear side of the pendulum to provide the actual applied impact force. The mass of the pendulum can be adjusted to three mass levels. The pendulum impact inertia and energy delivered when swung through a 30° angle are indicated in Table XXXXVII.

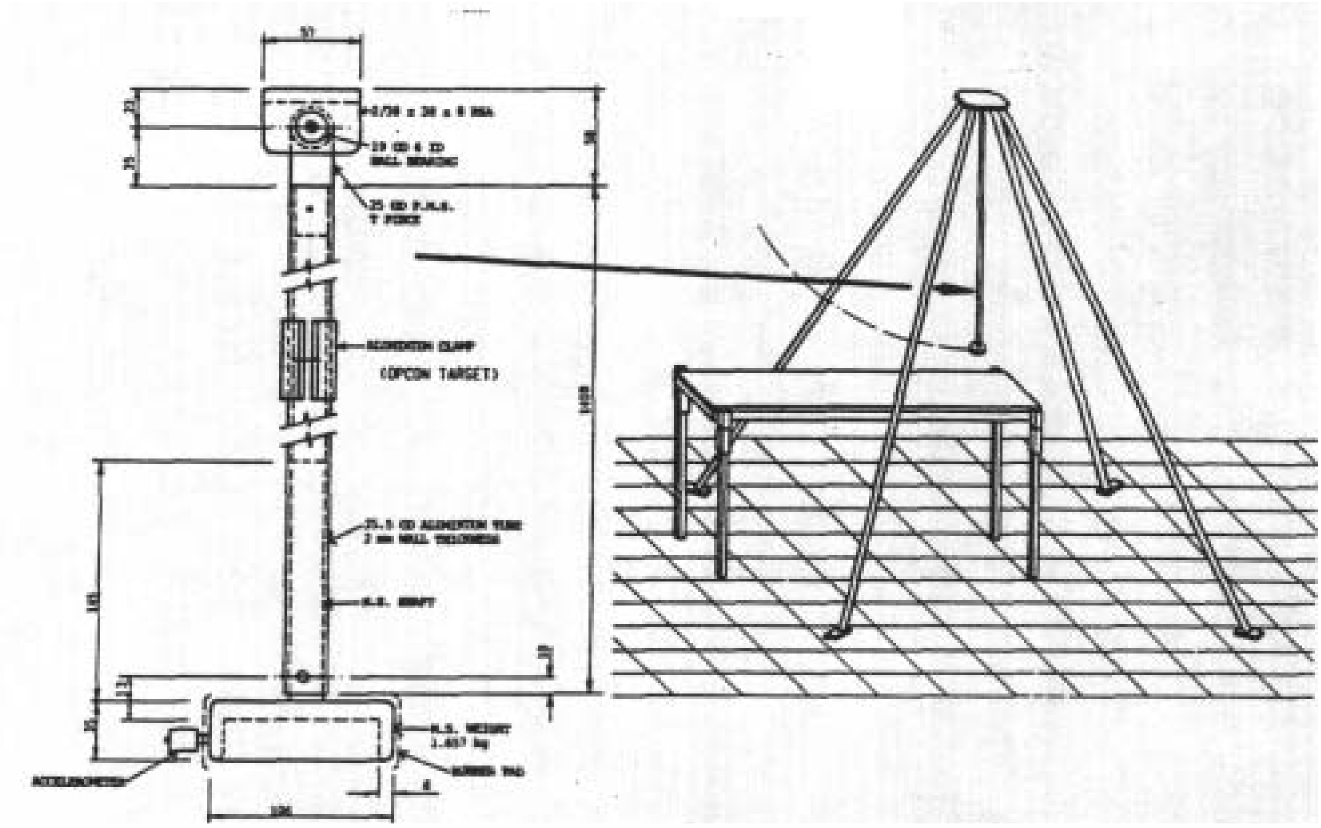


FIGURE 19 DETAILS OF THE JAW IMPACT RIG

TABLE XXXXVII

PENDULUM *IMPACTOR* INERTIA AND ENERGY

		PENDULUM	PENDULUM WITH LIGHTWEIGHT	PENDULUM WITH HEAVYWEIGHT
Total Mass	(kg)	2.694	3.137	5.302
Position of centre of gravity measured from pendulum pivot	(m)	1.058	1.156	1.231
Moment of Inertia about hinge	(kg/m)	3.912	5.929	8.955
Energy at 30° swing angle	(J)	3.146	5.678	8.731

NOTE: Distance from the pendulum pivot to centre *of* striking surface is 1.465m.

INSTRUMENTATION OF THE CADAVER MATERIAL EXPERIMENTS.

Referring to Figure 20 a piezoelectric accelerometer (2) located at the rear of the pendulum mass (1) detected the acceleration upon impact between the falling pendulum mass and the symphysis of the mandible. Its analogue signal was amplified by a signal conditioner (5) and fed on line to an analogue to digital (A/D) converter (6) which was connected to a data acquisition computer (7).

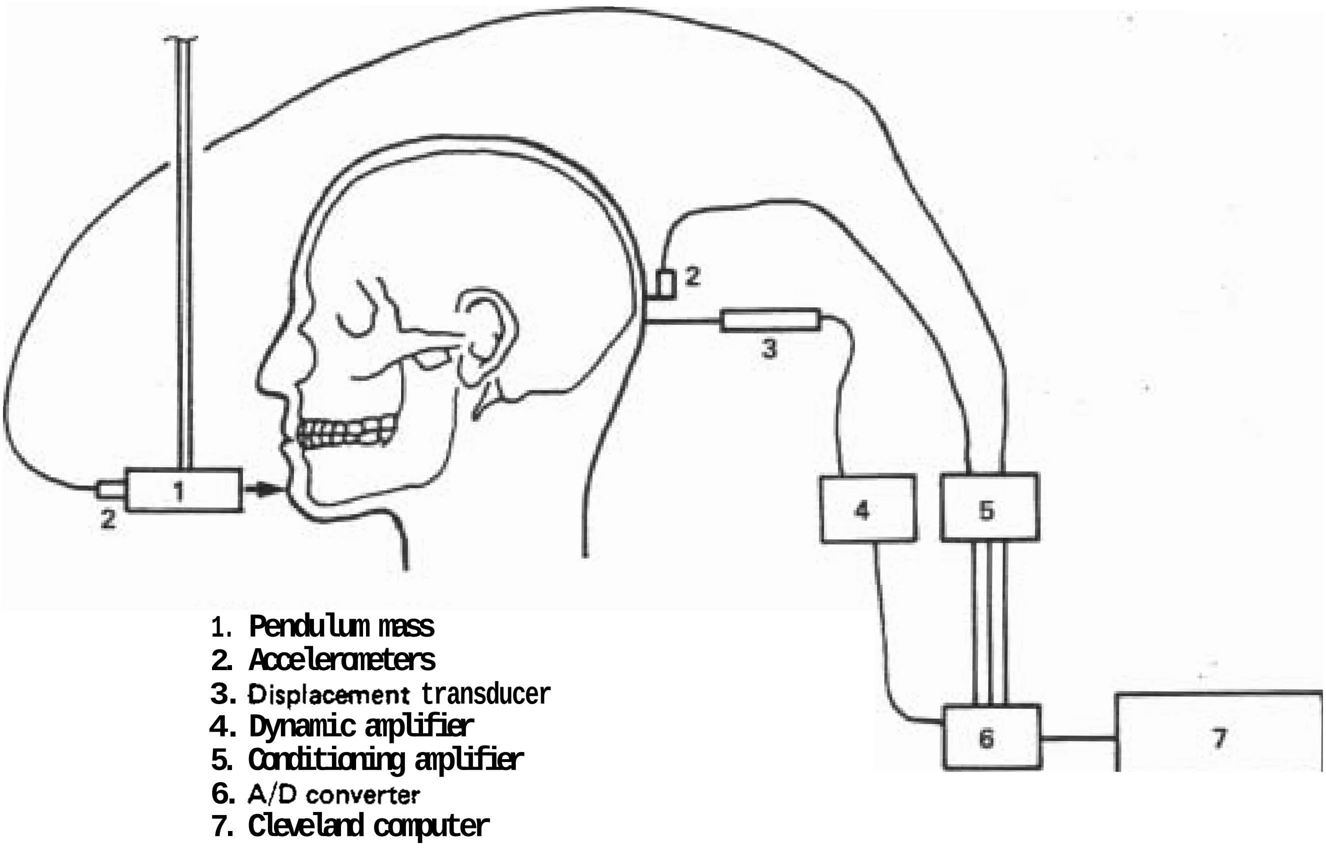


FIGURE 20 INSTRUMENTATION AND CIRCUITRY

A Schaevitz pressure transducer (5) which measured intracranial pressure was inserted in the parietal area of the skull and connected to a +15 V, OV and -15V power supply.

Displacement of the head was measured using a displacement transducer and head translational and rotational acceleration was measured by two accelerometers positioned as indicated (2) on Figure 20.

For the no helmet impacts the accelerometers were attached to the wrestler's harness while €or the helmet impacts the accelerometers were attached to brackets screwed to the helmets.

Data acquisition by the computer occurs from the instant the falling pendulum breaks the infra red light beam from an Opcon Thru Beam System. The system consists of an infra red source and detector mounted on opposite aides of the falling pendulum located close to the point of impact.

INSTRUMENTATION OF VOLUNTEER EXPERIMENTS

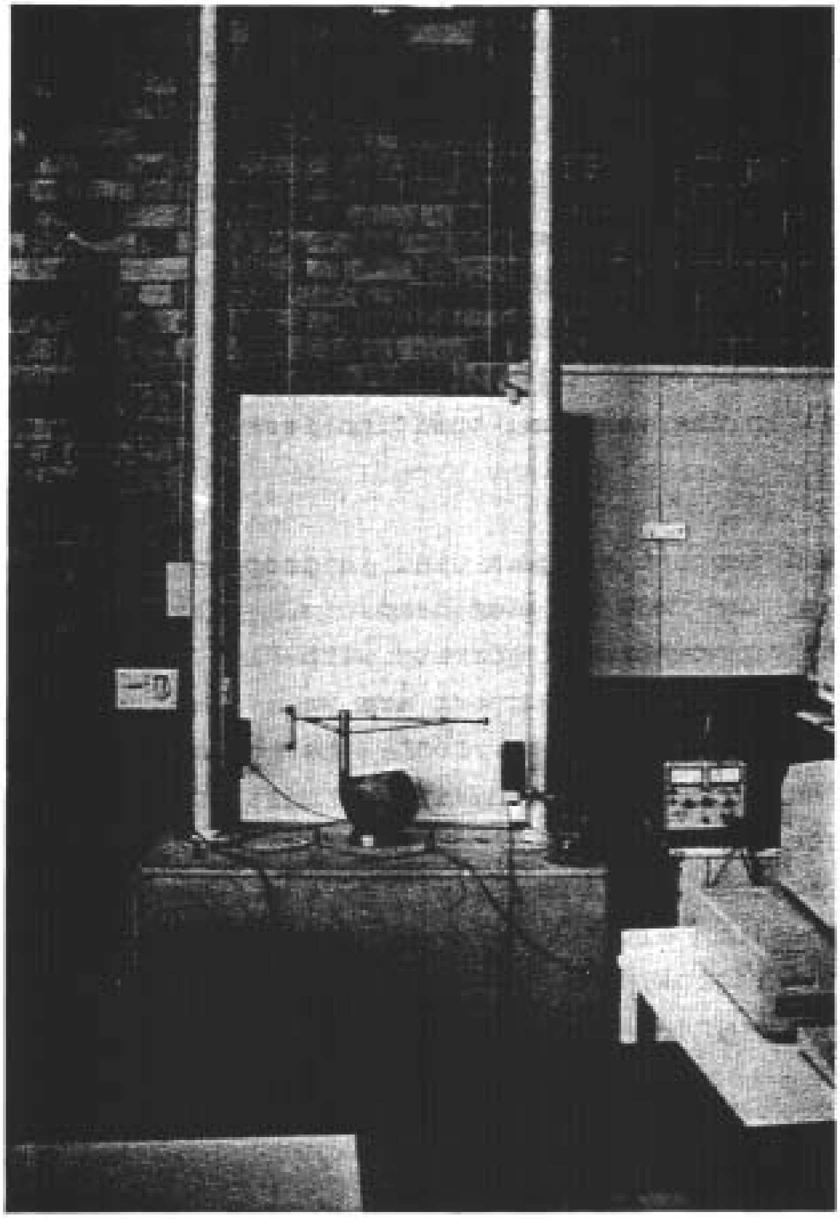
Instrumentation for the volunteer experiments was similar to the cadaver work. Two accelerometers were used. The volunteer was fitted with a band around the head above the earline with an accelerometer attached at the back of the head.

EQUIPMENT LIST

Bruel and Kjaer - type 4384 Accelerometer
Bruel and Kjaer - type 2626 Conditioning amplifier
Schaevitz Pressure Transducer - type number : P792-001
Pressure range (0-150 kN/m² VG) 0 - 15 kPa.
Opcon 1180A Source and 1280A Detector Modules plugged directly into Opcon DC/NPN control modules 8882A.

APPENDIX C - DEVELOPMENT OF A DROP RIG

The shock absorption capacity of the helmet. is tested by allowing a helmeted instrumented headform to drop onto a rigid flat or hemispheric steel anvil. The photograph below shows a vertical acceleration drop test rig. The shock absorption capacity is determined by recording the acceleration imparted to the headform fitted with the helmet at the moment of the impact.

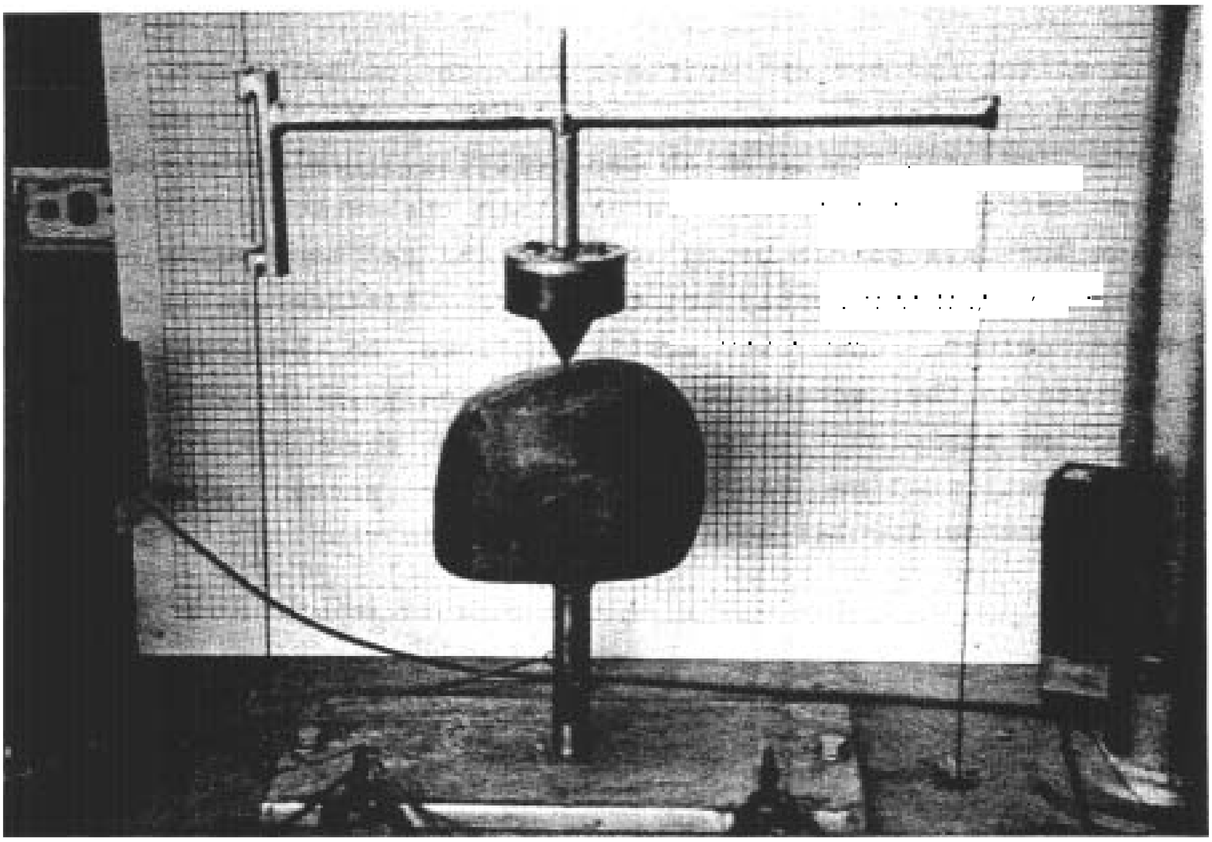


PHOTOGRAPH 4 VERTICAL ACCELERATION **DROP** TEST RIG

Two headforms were used, the magnesium alloy complying to the current Australian Standard and the WSU/Hodgson headform. The headform is attached to a support arm forming the drop assembly of specified mass. The drop assembly with helmet fitted to headform was allowed to free fall along guided vertical wires of negligible friction. The distance between the vertical wires is 530mm. A solenoid release mechanism when activated allowed the release of the drop mechanism from the centre of a cross arm which was adjustable to any height up to 3m above the anvil surface. The drop height of the assembly is specified according to the anvil used. The anvil in use is screwed into a rigid steel anvil base which was mounted on a reinforced concrete reaction block.

The support arm attached to the magnesium alloy headform uses a rotatable spherical mounting screwed to the inner cavity of the headform. This arrangement allows the adjustment of the position of impact between the helmeted headform and anvil. At the centre of the spherical mounting an accelerometer was mounted with its axis always in the vertical position irrelevant to the headform's position.

A second support a m was used in dropping helmets onto their crown using the WSU/Hodgson headform. The accelerometer was located at the centre of gravity with its axis in the vertical position. This second support arm was also used in dropping a sharp conical striker to test the helmet's resistance to penetration. The striker, having a mass of 1.93 kg and made of hardened steel, was allowed to free fall along the vertical wires to impact against the outer shell of the helmet which was fitted to the magnesium alloy headform. The magnesium alloy headform was mounted rigidly on the reaction block as shown in the following photograph.



PHOTOGRAPH 5 PENETRATION TEST ARRANGEMENT

INSTRUMENTATION

A piezoelectric accelerometer (Bruel and Kjaer type 4384) located at the centre of gravity of the combined test headform and supporting assembly detected the acceleration of the headform upon impact with the anvil. The signal from the piezoelectric accelerometer was amplified by a signal conditioner (Bruel and Kjaer type 2626) and transmitted to a Cleveland XT computer via an analogue to digital (A/D) converter.

An Opcon Thru Beam System and its associated electronics was used to trigger the computer to acquire data at a sampling rate generally set at 25000 Hz. Data acquisition by the computer occurs from the instant the falling drop assembly breaks the infra red light beam from a pair of Opcon source and detector modulues mounted on opposite sides of the impacting area. Each impact with the anvil produced an acceleration versus time pulse curve which was printed on hard copy and also stored on disc for future reference.

A scientific system software package called ASYST was used for data acquisition. One of ASYST's unique features was interactive graphics in which two vertical lines are displayed on the screen of the computer rather than the cursor. These lines follow the data points being plotted and can be moved along the axis either independently or together. This feature allows the region between the two vertical lines to be expanded and displayed on the screen for a closer analysis at a particular section of the plot. Also, a readout of the X and Y values at either vertical line can be shown on the screen by moving the line to the particular point of interest on plot.

APPENDIX D - DEVELOPMENT OF A TWO WHEELER CRASH SIMULATOR

The simulator is depicted in Figure 21.

The simulator launches a human dummy from a motorcycle, the dummy is then projected into or onto the desired impact obstacle. Thus the prime purpose of the simulator is to launch the dummy so that the impact can be simulated.

The design constraints for the launcher were that it be inexpensive, easy to construct and reliable in operation. Further. as space inside the laboratory was limited, the dummy had to be accelerated rapidly (between 2 and 3 g) so that a moderate speed accident could be simulated.

A number of means of acceleration were considered including compressed air and electrically driven flywheel types. However, these were rejected on the grounds of complexity and safety of operation. A falling weight system was used, the falling weight being connected to the cycle carriage by a 5/32" stainless steel cable through a 3:1 gear up pulley system. That is, the carriage moved 3 metres for every 1 metre of falling weight movement. Total movement of the carriage was 10.1 metres and movement of the falling weight was just over 3 metres.

One of the problems with all such systems is that kinetic energy has to be absorbed at the end of the stroke. In the case of the falling weight (which weighed 750kg) the energy was absorbed by a bed of sand approximately 0.4m thick and covered with hessian that had been laid out on the laboratory floor. The energy of the carriage was absorbed by a water filled shock strut, the water being driven through an orifice formed simply by a tap that was set in an appropriate position. The pitching motion of the cycle carriage was dampened by an automotive type shock absorber fitted to the carriage.

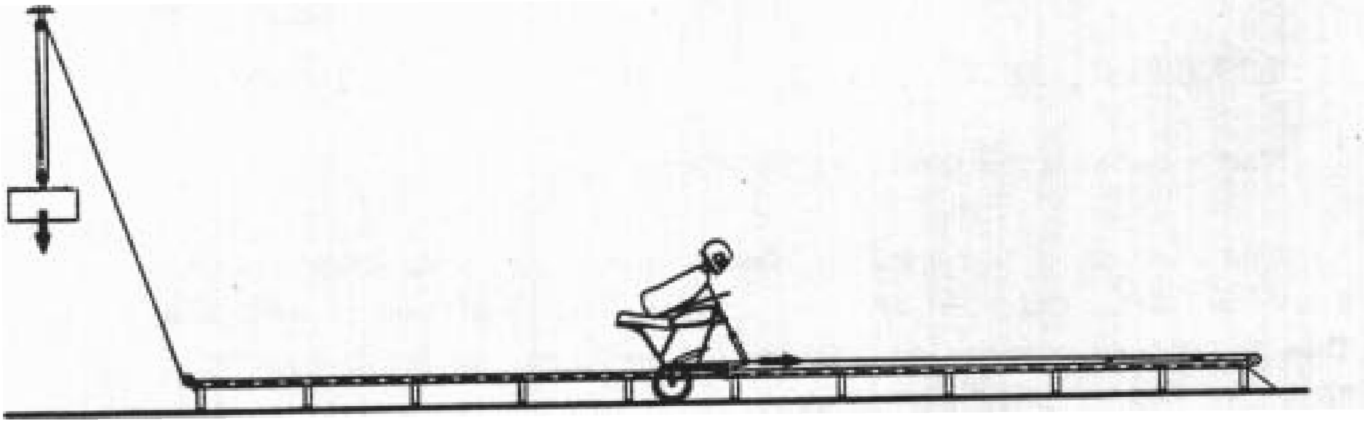


FIGURE 21 TWO WHEELER CRASH SIMULATOR

FIGURE 21 *TWO* WHEELER CRASH SIMULATOR

The cycle carriage was guided by a tubular rail, the actuating cable leaving the carriage going forward to the end of the rail, around a pulley, hence down the centre of the rail to the opposite end and from there via 2 more pulleys to the falling weight. Initially there was some problem with the cable binding in the pulleys resulting from cable springback after impact. This was eliminated by fitting cable guards around the pulleys. The system now works very consistently and reliably.

The carriage weight was limited to approximately 20kg and given the dummy weight, one may calculate the final velocity by doing a simple energy calculation. There is very little friction in the system, all pulleys were specially cast for this job and are fitted with bearings.

The launcher accelerates the dummy to 12 mlsec ($\simeq$ 45 km/hr). This launcher could be used in other impact work in the future.