CRASH INJURY BIOMECHANICS

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ABSTRACT: This Conference addressed issues important in biomechanical research aimed at reducing injuries to persons involved in road traffic accidents. The development and use of dummies, crash barrier testing and finite element modelling variously in Australia, Canada, France, Japan, Sweden and the United States of America was discussed with some emphasis on the formation of standards for vehicles, restraints and helmets. Specific research described was on: facial injury, with development of a Hybrid III load sensing face by Volvo in Sweden; the development of an infant impact head form, using stereophotogrammetry and computer modelling, in Canada; thoracic injury in France and the USA, involving experimental belt loading tests with dummies and cadavers, and side impact tests with the SID dummy, respectively; pedestrian accidents in Japan and France, using extensive data analysis and, at INRETS, impact tests with mechanical and cadaver legs; neck injury in Sweden, leading to recommendations on modifications to vehicle design, and in the USA; and head injury, which included crash reconstructions related to neuropathological studies of brain injuries in South Australia, experimental work on head impact tolerance using sub-human primates and human cadaver skulls in Japan, and brain injury work in the USA.

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The speakers, particularly those from overseas, gave generously of their time and the proceedings were very capably prepared by Dr. Jennifer Barker with assistance in the final preparation of the manuscript from Mr Craig Kloeden. The detailed organisation of the conference and appropriate secretarial services were carried out in a cheerful and highly competent manner by Mrs. Jodi Bock, assisted by Mrs. Elizabeth Kosmala.

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CRASH INJURY BIOMECHANICS CONFERENCE

OPENING FUNCTION

INTRODUCTION

Jack McLean

Good evening, Ladies and Gentlemen. You may have imagined that you would not get away without at least one or two brief comments this evening. In this room we have, I believe, the majority of people who have an active interest in injury biomechanics in Australia and clearly, because of the relatively small number of us, it's great that we have this opportunity to interact with each other and with colleagues from overseas. I'm particularly grateful, as I'm sure all of you are, to the State Government Insurance Commission, and to our overseas guests for travelling to this end of the earth or, as one of them might prefer to say, *la Finistère*.

Some might question the need for research in this area in Australia. Unlike the automobile, however, which comes in a bewildering array of makes and models in any given country, we more or less have the same make of human being in all countries, 2 basic models in a variety of age and condition. This means that research conducted, say, in France or Japan has direct relevance to Australia, but the converse is also true. Those of us who are working in the research area are well aware that there are many practical difficulties as well as ethical ones in crash injury biomechanics research. Here in Adelaide we have been fortunate in being able to overcome many of these problems in our work on head injury. In some respects, notably in the neuropathology side of the work, we have been able to go at least as far and possibly further than any other group. In other respects I think it's entirely obvious that we need all the help we can get, in technical terms. Apart from active research, those among us whose professional responsibilities include attempting to reduce the incidence or severity of injuries in crashes have much to gain from familiarity with the current state of knowledge in crash injury biomechanics and we hope that the next 2 days will prove to be of value in that respect.

I'd now like to introduce Professor Gavin Brown who is Deputy Vice-Chancellor for Research at the University of Adelaide, to say a few words of welcome and officially open the Conference.

OFFICIAL WELCOME

Gavin Brown

Dr. McLean, Ladies and Gentlemen. As the delegate from *ultima Thule*, I'm happy to welcome you to *Finistère*. It is a particular pleasure to welcome the overseas participants. We have Dr. Dominique Cesari from France, and Madame Cesari; Dr. Kennerly Digges from Charlottesville, Virginia, and Dr. Rolf Eppinger from the National Highway Traffic Safety Administration in the United States. As well, we have Mr. Koshiro Ono from the Japan Automobile Research Institute; Ms. Ingrid Planath from Volvo Car Company in Sweden, and Mr. Nicholas Shewchenko from Biokinetics and Associates in Canada. As well as welcoming these overseas delegates, of course I'm very happy to welcome the participants from Australia.

I note, in the material in the conference folder, that the economic costs associated with road crashes are quite staggering. Non-fatal brain injuries alone are estimated to cost Australia about $800 million per year. And of course, that is not putting a price on the human cost, because many of those who survive with serious brain injury are in a sense the lepers of our modern society, except for the burden and the trauma which they produce for their families.

As the Deputy Vice-Chancellor for Research of the University of Adelaide, I am both pleased and proud that this University, with support from the National Health and Medical Research Council, is at the forefront of research in reducing and understanding this problem. The objective, of course, is to minimise the appalling human and economic losses that are arising from head impacts. I hope that this conference will strengthen the links which have already been forged between the NHMRC Road Accident Research Unit here at the University, and the various overseas organisations which are represented here tonight. I also hope that the conference can act in some way as a catalyst for increased biomechanics activity in this area in Australia. You will find that an attempt has indeed been made to ensure a wide representation from invited participants from both State and Federal Departments of Transport, from vehicle manufacturers, motoring organisations and research workers in universities and other appropriate laboratories. I know that Dr. McLean has tried very hard to include all of the people with professional interests in the field in Australia at this time, but with a certain lack of taste he told me he'd had a sudden crash course in understanding that you cannot please all of the people all of the time. And, in fact, he found himself eventually turning away participants. You, of course, are the privileged few. But I'm glad to report that he is merely bruised and not traumatised, and he is looking forward to energetic participation rather than passive audiences for the next two days of the conference.

The NHMRC Road Accident Research Unit has received support from the State Government Insurance Commission (SGIC) for a large number of activities over at least a ten year period. Initially, this was in a study providing information on the cost of road crashes, conducted by Carolyn Somerville, now Carolyn Hewson. And then essential seed funds were provided for the head injury laboratory at the Institute of Medical and Veterinary Science, a development that was eventually rolled into the National Health and Medical Research Council funding. SGIC also later agreed to a request by Dr. McLean for a substantial contribution to enable a second neuropathology position to be funded. This resulted in the appointment of Dr. Grace Scott, who had recently completed specialist training in the neuropathology of trauma under Professor Hume Adams at the Institute of Neurological Sciences in Glasgow. Six years ago SGIC also funded a closed course study of alcohol-impaired night driving by the Unit. The principal investigator in this study was Hans Laurell, who was then Head of the Traffic Medicine Group of the Swedish National Road and Traffic Research Institute. Clearly the willing support of SGIC has contributed a great deal to what I regard as the outstanding success of this NHMRC Road Accident Research Unit.

The assistance and encouragement of the Senior General Manager-Insurance of SGIC, Mr. Richard Daniell, has been particularly appreciated. SGIC has provided support for conferences on road safety over many years, notably the annual road safety forums convened by the South

Australian Road Safety Advisory Council. However this conference is in many ways quite different. It is not aimed at a large audience, or at a topic that most people understand. One is therefore, perhaps, in some sense surprised, but pleasantly so, that an organisation such as SGIC should offer to support it and to support it wholeheartedly. I have rarely experienced such a clear desire on the part of a sponsor to do all that is necessary to ensure the success of a scientific meeting in order to further its pure scientific aims. It says a very great deal about the excellent management and judgement of the senior staff of SGIC and their view of the role of SGIC in the wider community. In particular, I would thank Mr. Russell Cowan, SGIC's Corporate Marketing Manager, for his part in making this Conference possible.

Now in the next two days, I suppose you will be discussing matters that most people would prefer to ignore: the force required to dislocate a knee joint, to shatter the facial skeleton, to damage the brain, to destroy the personality. Sadly, these events are quite commonplace in road crashes, but of course they have little meaning until they affect us or those who are close to us. As a community, in Adelaide, in Australia, and indeed worldwide, we should be grateful that highly trained persons such as yourselves are prepared to work in such an area: work to bring about amelioration of this modern day epidemic of trauma.

Thank you, and I trust that you will have a very fruitful conference.

KEYNOTE ADDRESS

THE IMPORTANCE OF BIOMECHANICS IN THE DEVELOPMENT OF VEHICLE SAFETY

Ken Digges

The purpose of this address is 3 fold: to give a brief history of biomechanics and how it has contributed to automotive safety to date, at least in the United States; to describe some recent developments in the United States which have an impact on the direction in which biomechanics work is going, at least in the USA; and finally, to identify some of the current challenges in the future work that should be done in the biomechanics area.

Crash Dummies. The basic tool for biomechanics is the crash dummy. The crash dummy has become very popular in the United States as almost a cult figure. It started in 1988 when a number of commercials were aired to suggest that people buckle up.

> *A video of one of these commercials, the original Vince and Larry commercial, which is extremely popular now in the United States, was then shown. This commercial shows two humans, dressed as dummies, being subjected to a crash, and advising that only dummies don't wear seat belts.*

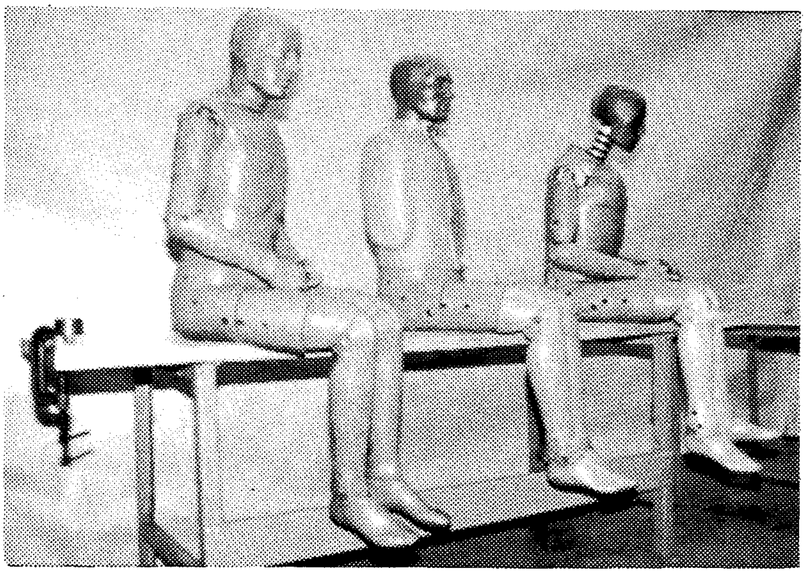
These two dummies now make regular appearances with wide coverage. For example, at the last SAE international meeting, people could have their photographs taken with Vince and Larry if they signed a pledge that they would always wear a safety belt. The latest addition to the dummy cult has been a toy car which is now on sale in Australia. If it is crashed into a wall, the front end collapses and, if it's got an airbag or a seat belt in it the dummies come out fine. But if none of those are available, then the dummies' arms and legs come flying off in all directions. It has become reasonably popular in the United States. And so biomechanics has come a long way, particularly in the popular area, and part of the challenge now is to bring the technical area up to the expectations that are seen in the public these days.

The basic tool, of course, is the anthropomorphic dummy. In the 1950s, the principal use of dummies (Fig. 1) was for testing ejection seats in the Air Force and the principal use was as a ballast; there were some accelerometers in the pelvic region, but all that could be measured was whole body acceleration and that was good enough. That was basically the way it started. About the same time, in the late 1950s, the University of California Los Angeles started running the early road crash tests, in which they used dummies that were very much ballast as well. The kind of thing that they found in those early tests was that ejection was very bad: doors came open and there were conditions where dummies went through windshields and so forth.

Standards. When the National Highway Traffic Safety Administration (NHTSA), which is the agency in the USA that develops the Standards, was established in 1967, the initial approach to Standards was still to use dummies only as ballast. Most of the Standards didn't require biomechanics data, because there just wasn't enough of it available back in those days.

Figure 2 shows some of the early Standards that were incorporated beginning between 1967 and 1970. They dealt with different protection features, none of which involved biomechanics. Included in the Figure are the kinds of protection that the Standards dealt with and the number of lives that were saved each year by that particular Standard. This is based on data from the National Accident Sampling System (NASS), so that the performances of cars before and after the Standard can be observed.

Figure 1:
Crash dummies - the basic biomechanics tool



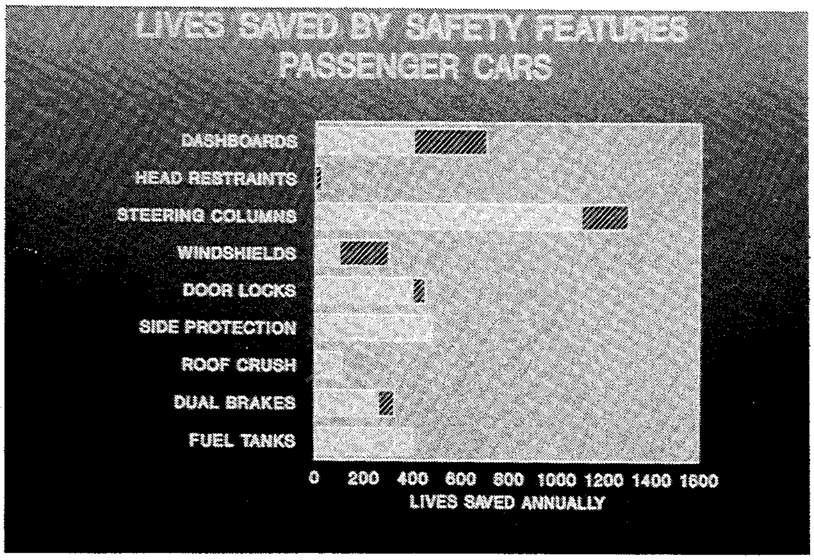
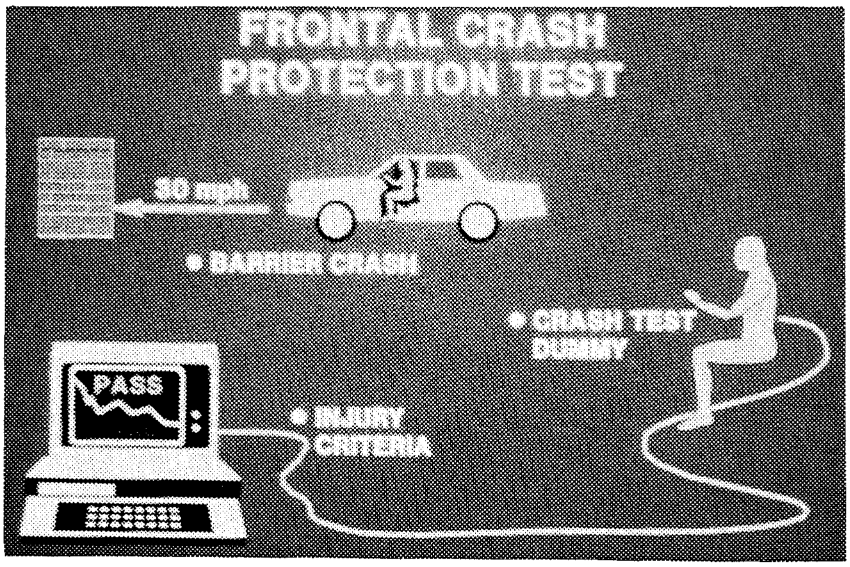


Figure 2:
Lives saved by car safety features

Figure 3:
Standard 208 requirements



There were four basic areas that these Standards dealt with. The first was setting minimum strengths of seats and head restraints, of belt anchorages and of belts themselves. The second was ejection prevention which included such things as improved door latches, hinges, and windshields. The third area was intrusion limits, involving such things as the static test of side strength and the static test of the roof, and the fourth was in the general area of friendly interiors, including padded dashboards, laminated windscreens, steering systems which didn't intrude excessively, and flammability of the interior materials.

Restraints. In those tests there was some fundamental biomechanics; for example, the impact test for the padding on the dashboard, but nothing approaching a full dummy test. The dummy tests came with the child restraints Standards and with Standard 208 for frontal protection. The sum of the savings from these early Standards was about 3,000 lives a year. However that doesn't match the savings from the use of restraint systems. As the data in Table 1 show, for child restraints the savings are about 250 lives a year, and for occupants over 4 years of age, the belt savings are about 4,500, so there are savings of about 5,000 lives a year from those kinds of Standards. Now these are the kinds of Standards that require biomechanics testing. They require dummies, and these are the Standards that are the big life savers. And there are the opportunities now for the future, to continue to develop these kinds of Standards that are based on human tolerance and the way that the human body behaves in a crash.

Table 1: Lives saved by safety restraints (USA)

	1987	1988
Occupants over 4 years	4,020	4,500
Infants and toddlers	213	250

The first real Standard to deal with occupant protection using biomechanics was Standard 208, the frontal protection Standard, and it required three things: a 30 mph barrier test, a crash dummy and some sort of injury criteria to go along with the dummy (Fig. 3). In 1972 when this Standard was first proposed, the principal objective was to measure the difference between an unrestrained occupant and an occupant protected by an airbag, because it was basically an airbag Standard. In trying to measure something that's that gross, a very sophisticated dummy is not needed. If no restraint was used, a 30 mph test with a dummy would give a Head Injury Criterion (HIC) of 1500 to 2000, chest g's of well over 60 and very high femur loads. If an airbag was used, the HICs would come down to under 500 and the chest g's would be well below 60. So such experiments would measure something that would have a very wide difference in the presence or absence of the safety system. Under those circumstances, the dummy sophistication was not nearly as detailed as that needed in this day and time.

The three principal criteria were for head, chest and leg injury. The Head Injury Criterion, which is an acceleration criterion, was based on the combination of cadaver testing conducted principally at Wayne State University, some human volunteer testing for the long duration acceleration time periods, and some animal testing. The human volunteer testing was done principally by Colonel Stapp in New Mexico, who rode an acceleration sled and measured accelerations on the chest and head — and survived! Similarly, the chest criterion came mostly from human volunteer testing. The human volunteers in some cases were people who were instrumented to dive off a high platform and their chest acceleration was measured when they hit the water. In other cases, accelerations from Colonel Stapp's restraint system, which was a complete belt system, were used. Finally there was a series of tests of some of the early airbags with human volunteers who were instrumented, so that there was that body of data for the chest acceleration. Lastly, there were the femur loads, which came from cadaver testing. It was fairly straight forward to test the strength of the femur.

This Standard (208) went into effect in 1987. The outcome has been two-fold. First, most of the manufacturers decided to meet that Standard by using passive belts — the type which automatically attach when the door is closed. However, as a consequence of public demand, there has been a move toward airbags and currently the airbag very rapidly is becoming the restraint system of choice in the United States. It has come in very rapidly — when the Standard started in 1987 there were almost 100,000 cars with airbags in them. By September 1991 there were almost 6 million cars with airbags and that number has now risen to almost 9 million. Last year Congress enacted a requirement that by 1998, all passenger cars, light trucks and vans be equipped with both passenger- and driver-side airbags.

There is a video which shows the way that the airbag has come into being and the way it's been sold particularly by Lee Iacocca, President of the Chrysler Corporation. He put airbags in all Chrysler cars and sold them very vigorously in the United States. The following video, produced by the Insurance Institute for Highway Safety, illustrates the education process which is ongoing in the United Sates.

This video was then shown. The dialogue follows.

> ***Voice 1:*** *"Keeping people alive in crashes means keeping them from being ejected or slamming into the hard interior of the car. Safety belts are good at this in many kinds of crashes. At low and moderate speeds they keep people from hitting the steering wheel, dashboard or windshield. But safety belts aren't the complete answer. In crashes at higher speeds, people wearing belts can be injured very seriously. Their faces and heads are especially vulnerable. Frontal crashes are the most common kind of crash. They account for more than half of all deaths and serious injuries. Airbags prevent many of the crash injuries that safety belts still allow. Airbags are in millions of new cars, but they're still not as familiar to motorists as safety belts. They're stored out of sight in the steering wheel. If there's a passenger bag, it's in the dashboard. When a crash begins, sensors located in the front of the car signal the bag to inflate. Almost instantly the occupant is cushioned by a pillow filled with air. Then the bag begins deflating, to further the cushioning effect. It all takes less than a second. People who use their safety belts may think they don't need airbags, but they do. Both save lives. And, as former Transportation Secretary Elizabeth Dole says, the combination of airbags and belts provides protection at higher speeds than safety belts alone do. And they will provide protection against several kinds of extremely debilitating injuries, like brain and facial injuries."*

> ***Voice 1:*** *"Airbags aren't designed for rear or side impacts or rollovers, but in frontal crashes airbags and safety belts are the best protection. Auto makers today recognise this, so new cars from virtually every car company are being equipped with airbags and the number is growing."*

> ***Lee Iacocca:*** *"In every single passenger car we build in the US, Chrysler puts a driver's side airbag in as standard equipment. We were the first American car company to do so. That's 1 million airbags a year. They've proven reliable and in some cases have flat out saved people's lives. They aren't intended to work alone. You still have to buckle up, and believe me you're a fool if you don't, because together, bags and belts are the best protection around."*

That, particularly the last bit, sort of summarises the sell that Lee Iacocca has put into it. Subsequently the other manufacturers that have been offering airbags have put a great deal of air-time and ink into the safety of the airbag and as a consequence of that, the consumer demand has been very strong.

The National Highway Traffic Safety Administration has been monitoring the performance of both the airbags and passive belt systems, and published a report last month on the effectiveness of these systems as compared to the cars that were in the fleet before this Standard came into effect. The base year was 1983. The study found that the effectiveness of passive belts, depending on the type, ranged from 7 to 16%, and the effectiveness of airbags was around 23%. That's overall effectiveness, not just frontal effectiveness. So the numbers are extremely encouraging. Both systems are working as they should work, but the airbags and manual belt systems, as used, have a higher effectiveness, as is expected.

Another piece of good news that was reported in the study was that there has been no reduction in the belt use of people who bought airbag cars. So one of the concerns - that people would let down their guard on safety if they had this additional safety equipment - is proving to be unfounded.

Part of the study also provided some very interesting information that could be used in determining needs in biomechanics. The group who conducted this study examined all the fatal accident cases in the National Accident Sampling System, in cars equipped with either passive belts or airbags, to see how people were getting seriously injured and killed in these cars. And they were able to find 81 cases in which there were fatalities in cars equipped with automatic belts or airbags, and in these cases they found the following:

- no belts were used in 50% of the cases, so that half of them were still unrestrained occupants - and most of these were passive belt cars;
- 34% were unsurvivable because of the massive nature of the crash;
- in 53% of the cases, intrusion of the vehicle structure played a major role;
- ejection was present in 20% of the cases, and it is well known that ejection is a very dangerous situation. Occupants who just stay in the vehicle are 6 times better off than those who are ejected; and finally,
- occupant age and size played a role in 19% of the fatalities.

So the same old problems recur — no belt use, and ejection. But the new ones are intrusion in 53% of the cases, and the elderly and size in 19% of the cases. Those are some of the new challenges in biomechanics.

Age and size issues associated with biomechanics. Leonard Evans, who has been visiting Australia for the last 2 weeks and has been speaking around in various parts of the country, published a landmark paper in 1988 in the Journal of Trauma, in which he used the FARS (Fatal Accident Reporting System) study to examine the effect of sex and age on people's susceptibility to fatality in crashes of equivalent severity. Some of his findings are set out below.

Fatality risk from physical trauma in motor vehicle crashes

For comparative severity exposures:

- Fatality risk is least at age 20 years
- Risk for females is 25% higher than for males (ages 15-45 years)
- Risk at 70 years is 3 times greater than at 20 years

His finding that the fatality risk was least at the age of 20 years is basically good news, as that is also the age where the the risk of being in a crash is greatest. The higher risk of females compared with males in these equivalent severity crashes suggests a possible lower biomechanics tolerance. And finally, the high risk at 70 years compared with that at 20 is a fairly well known biomechanics phenomenon which has been seen in most of the testing done to date.

The meaning of this in terms of the performance of restraint systems must be considered. In a paper published in 1982 in AAAM (Association for the Advancement of Automotive Medicine), Ziegler examined the performance of different sized dummies in restraint systems. His restraint system was designed ideally for a 50th percentile male dummy and he found that, if he took the position for the 50th percentile male dummy which produced a chest g of about 65% of the maximum allowable and a HIC of about 55% of the maximum allowable, and then replaced that with a 95th percentile dummy and did nothing else, the chest g's went up a little bit but the HIC stayed about the same, so the system worked reasonably well for a 95th percentile dummy

(Fig. 4). But when a 5th percentile female was used, both the HIC and the chest g's went up significantly. Now the same thing applies if the dummy is made even smaller, so that if a child dummy is used, the situation gets even worse. So there is a real concern about the performance of restraint systems for both the 5th percentile female and also for children. The injury criteria for small adults and for children are not well established.

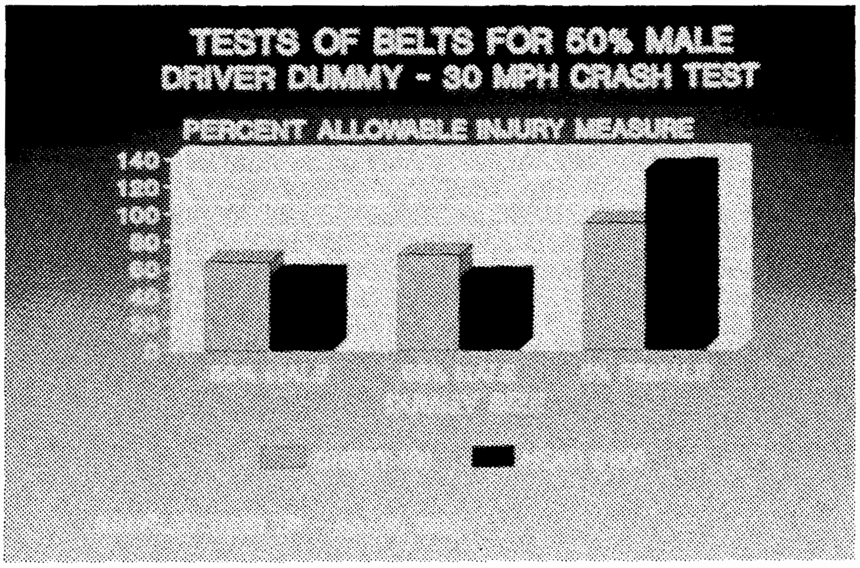


Figure 4: Effect of dummy size in restraint tests
[stippled columns = chest (g); black columns = head (HIC)]

To consider the age problem, some background on belt loads is needed. The kinds of belt loads that are reported by the NCAP program that's run in the United States are shown in Figure 5. This is for the 35 mph frontal barrier test. The average load for the driver is around 1700 lbs and for the passenger it's around 2000 lbs, as measured on the shoulder belt. The reason that the driver is lower is that some of these vehicles had airbags and in some cases the steering wheel may have mitigated the acceleration to some degree.

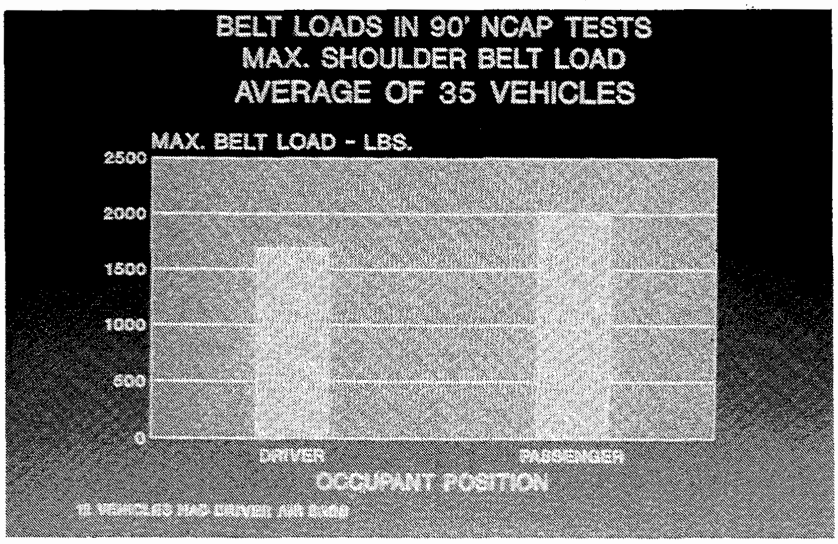


Figure 5: Belt loads in NCAP tests

This data should be kept in mind when considering Dr Eppinger's injury criteria published in 1977 (Fig. 6). In this case he looked at age as a function of belt force. For a 20 year old, a load of 1700 lbs may cause one or 2 broken ribs, but at up around 70 years the injuries would be likely to be a flail chest with 19 or 20 ribs broken, so the tolerance here is quite different.

Currently there is not the capability to make these kinds of biomechanical measurements with the dummy so that belts can be designed which are suitable to deal with young people and old people at the same time.

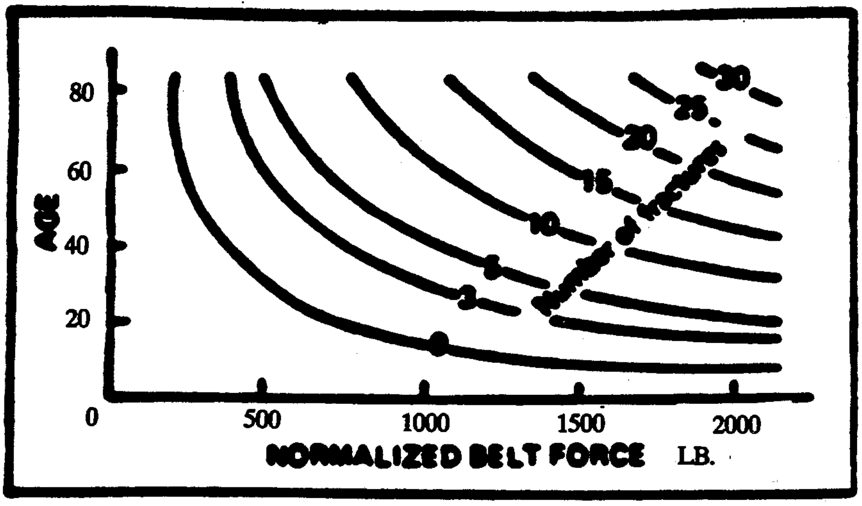


Figure 6: Effects of age on chest injury tolerance in frontal impacts
(after Eppinger, 1976)

Other challenges in biomechanics. In Australia, the annual cost of road accidents is around \$3.1 billion. In the USA it comes out to about \$40 billion for the monetary cost and if the comprehensive costs associated with loss of quality life are included, the cost is up to about \$150 billion. Therefore it is probable that in Australia also there is about a 4:1 ratio for comprehensive costs to monetary costs.

If this total cost is broken down into its components, the opportunities inside the car for improving occupant protection can be examined. Table 2 shows the percentages of the total \$3.1 billion that are associated with particular types of harm, so that if a particular percentage is multiplied by \$3.1 billion, the approximate cost of the harm and injuries associated with this particular combination is obtained. For example, damage from the dashboard to the lower extremities contributes about 5% of the total harm for restrained occupants. This is certainly an interesting challenge. At the present time, our dummies will measure femur loads, but a lot of leg harm is in the ankles and the lower legs, for which there are no current requirements as far as regulations are concerned.

The seat belt has been highlighted in Table 2 to show the opportunities for making some improvements with it. In a lot of cases, the problems are with elderly people in them; in some cases it's mis-use of some kind that is involved. In any case, the criteria and the dummies to allow better belt designs for the range of population that is exposed is one of the biomechanics priorities. Other combinations in Table 2 also have significant amounts of harm - the dashboard versus head/face, the windshield versus head/face, the steering assembly versus chest and the steering assembly versus head/face all carry fairly high numbers. However, most of these will be addressed by airbags and so it is probable that the first 4 combinations listed in Table 2 are the ones that may need the most attention in the long term. The remainder are short term needs until airbags come in, and since bags are becoming standard in the United States, the demand will doubtless be world wide as well over the next 10 years.

One of the issues that is highlighted by some of these data is that intrusion plays a very large role in the dashboard/lower extremity problem, as was noted in the study that was published recently by NHTSA. Consequently there has been a lot of interest created worldwide in the offset crash case because that is the one that produces the large amount of intrusion. There is currently a world-wide discussion on what should be the next level of crash testing: should it be an offset test, and if it's an offset, what should be the degree of overlap. There's a big argument that a 50th percentile overlap is correct, another one that it ought to be 30%. The intrusion from this range of overlap is quite different. But the main point of interest is the fact that massive intrusion injuries occur in 53% of the fatal cases and that kind of intrusion is not seen in the full frontal barrier test. Should some offset test be a requirement? And if this is the kind of test to be run, then dummies and criteria will be needed to deal with the lower limb injuries that come along with it.

Table 2 : Opportunities for restraint improvements

Component	**Body region**	**Harm**
Dash	Lower extremities	4.8 %
Seat belt	*Chest*	*3.7 %*
Seat belt	*Abdomen*	*2.5 %*
Non-contact	Neck	3.0 %
Dash	Head/Face	1.3 %
Windshield	Head/Face	5.3 %
Steering assembly	Chest	6.1 %
Steering assembly	Head/Face	4.1 %

The same Act that required airbags in cars by 1998 also required NHTSA to publish a Standard to provide head protection for A pillars, headers and roof rails - some kind of padding Standard. The approach to date has been to use the Hybrid III dummy test device in which the dummy head impacts in various places inside the vehicle - there have been 2 or 3 papers published by NHTSA on this testing. The criterion which is being used at the present time is HIC. The question that arises, though, is whether HIC, which has been around since 1967, is still the right criterion for this kind of impact. Are there sufficient rotational accelerations to cause injury? So there should be concern about the rotational acceleration. An even bigger concern arising from the accident data is that there are a very substantial number of injuries associated with striking concentrated objects like the bolt that attaches the D-ring to the B pillar. At the present time this test device doesn't pick up that concentrated load.

Finally, there is the area of side impact (Fig. 7). NHTSA has just published a Standard in that area and there is still a considerable amount of research going on. Part of the Standard is a new dummy and a new set of criteria called the Thoracic Trauma Index (TTI) which uses the rib and spine accelerations as the basis for that Standard. There has been an international discussion on whether or not the acceleration criterion is the best criterion, or should it be a deflection criterion, or should it be a combination of deflection and velocity. One of the things that this criterion does give is a very sensitive relationship between the age of the victim and the characteristics of the injury.

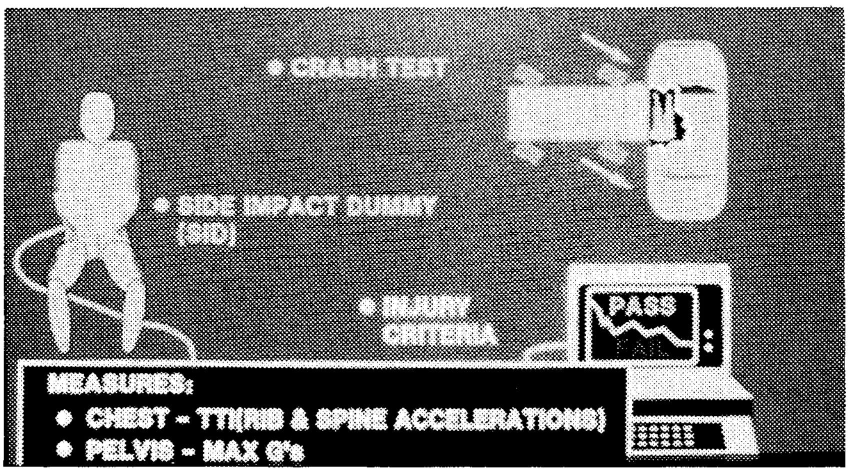


Figure 7: Side crash protection test

Figure 8 shows that a TTI of 75 or 80, which is roughly the region where the Standard was written, provides very good protection for someone at the age of 20, but at 70 years old the protection is considerably worse. So the area of side impact still needs to deal with this kind of dichotomy. For example, work concentrating on protection of the elderly, who some believe are over-represented in the population involved in side impacts, may produce a totally different padding than which would protect a younger group. This whole area of side impact is one that needs continuing biomechanics research and development and enlightenment.

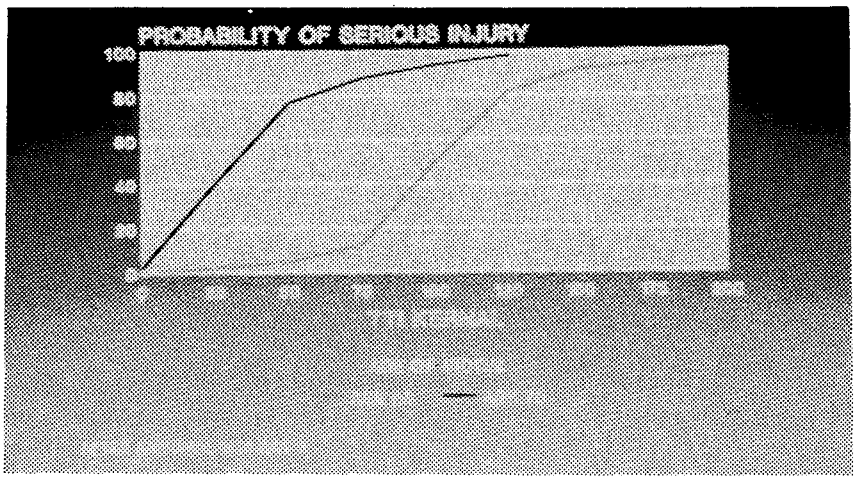


Figure 8: Effects of age on chest injury tolerance in side impacts

Observations in conclusion. Firstly, biomechanics technology has been the key to some of the great safety improvements that have been seen in recent years. Certainly it's the key to the occupant protection Standard that will be able to save up to 10,000 lives a year when it is fully implemented. It is the key to the side impact Standard and to the interior head protection Standard, which both rely on biomechanics.

The easy part of safety - that associated with occupant containment and with interior-friendly materials - has been done. The challenge now is to provide dummies and criteria to measure adequately the kinds of injuries that are being seen to the head and lower limbs when they contact vehicle interior, and to the chest/abdomen in association with belts. Also the combination of belts and bags is a real biomechanics challenge because the current injury criterion does not address that kind of loading. Again, injury criteria for children, the elderly and women are needed. Finally, the interpretation of biomechanics test results versus injury risk is needed so that benefit calculations can be made.

How is this to be brought about? Jack McLean mentioned yesterday that work in the biomechanics of injury area is simpler than that of the design of vehicles, because there are only 2 models. However, the various sizes and ages that those models come in adds a great deal to the complication. To support and continue his idea, the comment can be made that it does indeed take world-wide interaction to develop the capabilities and technologies in this area. The basic tools available to biomechanics research include about 5 or 6 different technologies. There are human volunteer tests to provide biomechanics information. But this can be obtained only at relatively low threshold levels. There are animal tests from which information can be derived, but the problem is translating from one species to another. With cadaver tests, the interpretation of the living tissue versus the dead tissue is open to speculation. It is particularly difficult in the case of brain injury because both the physiological functions and the psychological functions are very difficult to understand in dead tissue. There is the use of real world accident data, but there is no instrumentation on the people in the accidents, so reconstructions and models and very careful analysis of the accident data are needed to get biomechanics information. Also mathematical and physical models can assist. Finally there is the development and associated testing of dummies, which are really physical models.

Each of those present has a piece of the biomechanics puzzle. Each has capabilities in at least some of the areas and information from each needs to be brought together. In particular, it is of interest to cite some of the activities that have gone on in the different countries that are represented at this Conference.

In the USA, of course, the strength has been in the development of dummies and criteria for frontal and side impact Standards and the development of injury surveillance systems which allows calculation of the benefits of the regulations and safety features that are promulgated.

Sweden has made some very significant contributions in the development of accident data, the analysis of accident data, the development of safety rating systems from that accident data and the calculation of the benefits associated with that accident data. They've done some very pioneering work in dummy testing, in the development of injury criteria and methods of measuring facial injury and in examining pedestrian injury and how to prevent it. And finally, and most importantly, they've done an outstanding job of implementing the results in their cars, particularly in the case of the three point belt. A great debt is owed to Nils Bohlin and Volvo for developing that particular technology which has been the largest life saving technology and probably the most cost-effective safety feature, the most cost-effective public health feature, that was ever developed.

France has been very much in the forefront in developing injury criteria. They've done some pioneering work in human volunteer head injury testing where they measured the head acceleration of prize fighters. They've done extensive work on cadaver testing to develop chest, pelvic and lower extremity injury criteria. They pioneered in reconstructing crashes with both cadavers and dummies which was unique in the world. They've been in the forefront of developing analytical models and of developing dummies for pedestrian and side impact.

In the case of Japan, it has been fortunate to have Koshiro Ono here as he has done some pioneering work in animal testing in the study of head injury, which needs a living animal to get much of the information that is needed. And in general, the work in Japan on analytical models, particularly finite element models for safer vehicle structures are world class.

And of course Australia is last but certainly not least amongst safety accomplishments. They have led the world in the development of dummies and test procedures for people who travel in cars in wheelchairs, and should be very proud of that accomplishment. Their accident data and accident analysis has been very beneficial world-wide. One of the greatest contributions has been a consequence of the world's first belt-use laws and the large belted population which resulted. Consequently the analyses that have come out of Australia have led the world in promoting safety belts and have been extremely beneficial. Also, the initiative at the present time with helmet laws and bicycle helmets will surely provide some pioneering information in the future. Australia also has the ability to get motor vehicle injury cost data which is available nowhere else, because of the universal insurance coverage and the way that road accidents are paid for in this country. When that insurance data is coupled with the cost data and the road accident investigation results, there is a unique opportunity of measuring benefits and total costs of vehicle safety systems that are in place. And finally, the head injury research in reconstructing crashes and collecting detailed information on the trauma to the brain, has been leading the world. This research at the University of Adelaide provides a singular basis for really understanding what is causing brain injuries and for developing validating models to understand those brain injuries.

It is therefore very appropriate that Jack McLean and the RARU group have organised this meeting so that all participants can share their understanding of biomechanics. We are particularly indebted to SGIC for hosting this group and bringing us all together. Each of us has an important piece of the biomechanics puzzle and it is important that all of the pieces be put together so that we can move forward to reduce the deaths and injuries on the highways. I certainly look forward to participating in this conference.

QUESTIONS/COMMENTS:

Peter Makeham: I was very interested in your comment that the United States Congress has issued an edict that by 1998, all cars and recreation vehicles would have airbags. What are the Standards for airbags *per se*? Is there a technical Standard, and where would one find it?

Ken Digges: The 208 Standard is a performance Standard which can be met either by airbags or automatic belts. What the Congress has said is that the only way that the 208 Standard can be met is by airbags - it can't be met by automatic belts any more. There would have to be airbags on both driver and passenger sides. Congress has accepted the popular demand and gotten in front of it. NHTSA was neutral on it. In their testimony with regard to it, they took the position that the legislative branch shouldn't monkey around with the Federal Standards which are the responsibility of the executive branch. Manufacturers had already made the commitment to provide airbags. They'd all said that they were going to offer airbags as the only option by 1998, so no-one opposed the legislation vigorously. Consequently the legislation went through. And that legislation also provided the requirement for upper interior padding.

Bryan Knowles: I'm intrigued a little bit by the attitude with regard to seat belt use in the States. I've lived in the United States for 4 1/2 years, so I'm aware of the constitutional rights of the individual, but the challenge is going to come on the motor industry for elderly and small children to meet these requirements, which is a compromise when you have to meet in the States unbelted personal requirements. The problem is, is the Government going to help the industry by mandating the wearing of seat belts?

Ken Digges: It already has. Part of the benefits that I talked about for airbags and belts is a benefit associated with an increase in belt use over that same period of time. It has been very rewarding to see the belt use go from 14% in the early 1980s to 50% at the present time, and that increase constitutes saving 3,000 to 4,000 lives every year - just from that increment of belt use. There has been a major federal program over the last year with a goal to get the belt use to 70%. There are advertisements, promotions before every holiday and there is improved enforcement. Indeed most States now have belt-use laws and even in those States that don't have belt-use laws the belt use seems to be going up. Whether or not it will ever get to 70% remains to be seen. Over the past few years it has tended to go up by about 2% a year. So I think the Federal Government is doing everything they know how to get the belt use up. I think that you're seeing the advertisement of airbags consistently mentioning that if you have airbags, wear your belt also. However, you have a continuing problem in Australia because you've got 6% that don't buckle up and that's 30% of your harm. We've got 50% that don't buckle up and that's 65% of our harm. So it is indeed a problem that everybody, including the auto industry, needs to look at.

Bryan Knowles: Well, with respect, I think the Government can do what Australia has done and make it illegal to drive a car without a belt. Now in the States where they do have belt-wearing laws, in my understanding it's not a primary offence. You can only be booked for not wearing a belt if you've been pulled up for some other traffic offence and if you've not wearing a belt then they can book you.

Ken Digges: There's a considerable variation in the state laws. A few states have that stipulation, but in most of them it is a primary offence and indeed part of the enforcement program is to go to primary enforcement . I'm not sure that we're the ones to debate this particularly. It's a complex issue but I think we're both on the same side, basically. It is just a very frustrating situation that people take risks with terrible consequences when they don't have to.

IMPACT BIOMECHANICS FOR SAFETY IMPROVEMENT

Dominique Cesari

The biomechanics of impacts is a relatively new science which was developed to provide knowledge of mechanisms producing injuries and human tolerance data which is necessary to improve the safety of vehicles and to provide a better protection of road users in case of accidents. During the last 20 years, the number of traffic fatalities in European developed countries has been halved and if the increase in traffic is taken into account, this should be considered as being divided almost by 3. Impact biomechanics have greatly contributed, and are still contributing, to the improvement in road safety. The first part of this talk will be rather theoretical, but it is important to understand how biomechanical research should contribute to the global results of improvement of safety for road users; this means safety in terms of passive safety or crash protection.

Figure 1 gives an illustration of a series of interactions through which such improvement in road user safety should be brought about. First, if the number of fatalities has been reduced, it becomes more and more difficult to continue research, especially when the scientific approach is needed, that will continue to decrease the number of fatalities and severe injuries, because it's more and more difficult to find further solutions to the problems. Therefore, what is needed is knowledge of what happens in accidents, in real life. The accidents give information in terms of frequency and severity but also, to some extent, in terms of injury mechanisms. Nevertheless what is found on analysing accidents is the total research and it is always difficult to get details in terms of what happened, at which time during the process. So the first part of the activity is to study accidents at two levels, the statistical level and the detailed level, to correlate characteristics of the impact and the consequences of that impact.

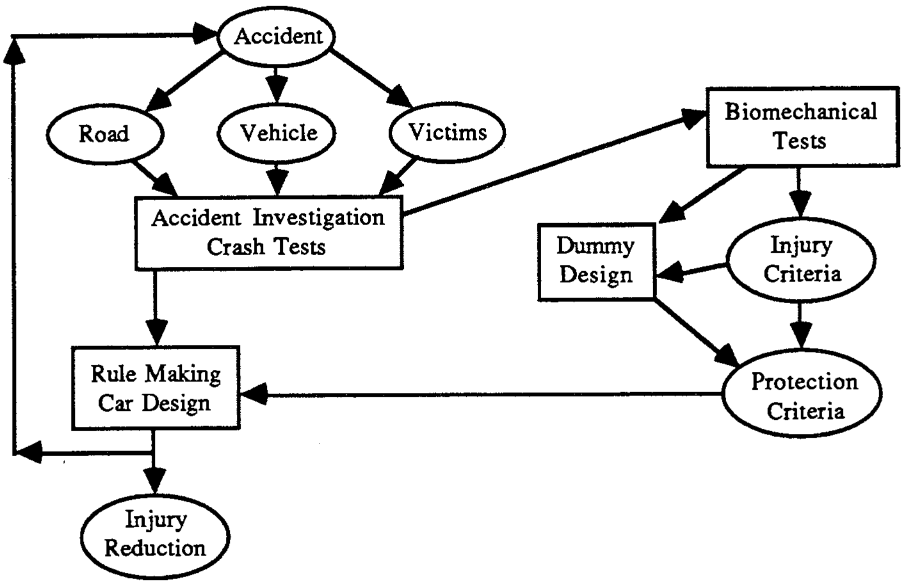


Figure 1: Flow chart for improvement in road user safety

The final objective is to reduce the severity and the consequences of injury and for that some input from the biomechanical tests is needed, because if the objective is to be able to reproduce what happened in accidents or to select test types of test conditions which correspond to accidents, it is necessary also to provide information in terms of injury criteria and to provide models of humans to put in the cars in the global test — the full scale test. So this is generally the part of the biomechanical research which is necessary to provide test conditions for better safety.

There are several types of problems. One is the individual variation in the human population:

because two persons are different they do not react the same way for specific loading. Nevertheless what is wanted is to develop what are called injury criteria, which are relationships between the severity of the injury for a specific injury and a physical parameter, which may be a force, pressure, acceleration or other. But the large variation within a population means that the injury severity parameter is merely an average for the population. But that means that for a specific parameter, some of the population will sustain more severe injuries and some less severe injuries (Fig. 2).

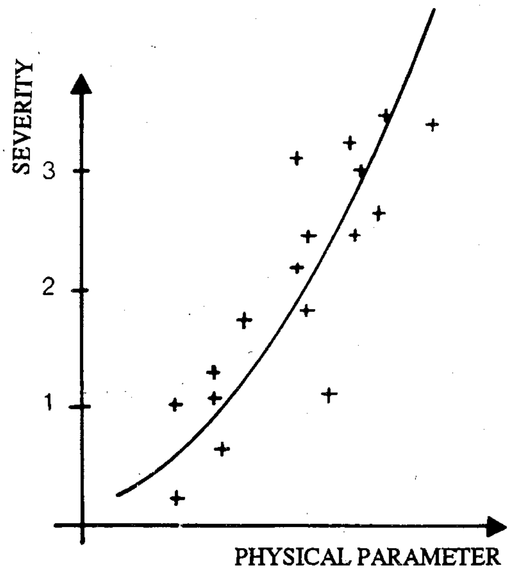


Figure 2: Relationship between injury severity and physical parameter

The second point is that part of this variation in the population is due to age and it is clear from Figure 3 that the young sustain more severe impact than the elderly. That means that it is necessary to consider the population at risk for a specific accident situation. For example, the pedestrian population is completely different from that of car users. The pedestrian population consists mainly of children below 10 years, or old people, whereas car drivers are aged mainly between 18 and 60 to 70 years. So they are quite different populations.

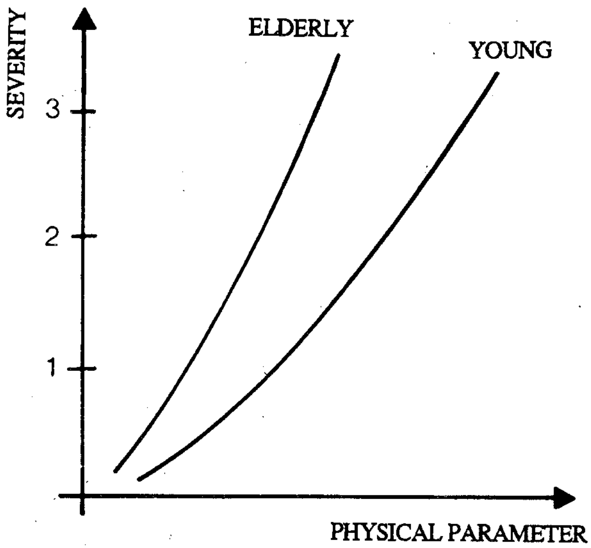


Figure 3: Effect of age on relationship betwen injury severity and physical parameter

The population also varies according to type of accident. For example, there are population differences between front and side impacts, or in pedestrian accidents, between those striking and those struck down. This is because people don't react in the same way according to their age. But this has to be considered in terms of biomechanics, taking into account the resistance of human tissues according to the age of the population. Figure 4 is an example of variation of pelvis tolerance in side impact, according to the age of the population between 20 and 80 years. This is the population in the struck vehicle and it is clear that this population cannot be represented by the value for the youngest, which is about 12 to 13 kN: a reference value for that specific problem is somewhere around 10 kN.

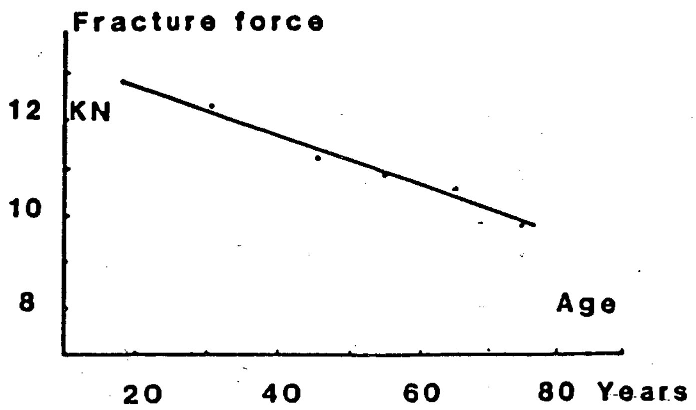


Figure 4: Variation of pelvis tolerance in side impact, according to age

Another difficulty is that for a specific type of injury, there can be different severities (Fig. 5) and what has to be considered is the population to be protected.

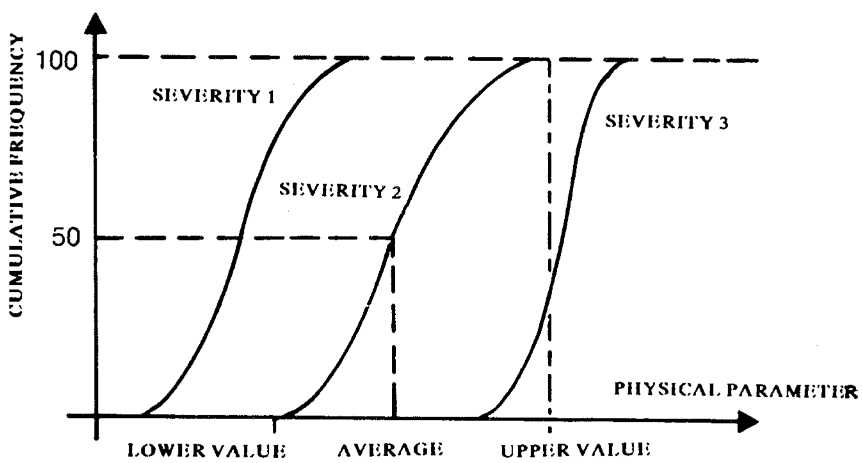


Figure 5: Levels of injury severity

In this field of accident protection it is not possible to protect everybody against all types of injuries. What must be decided is which part of the population has to be protected and at which degree of severity of injuries. So this means that some injuries are considered to be acceptable and some are not. That depends on severity not only in terms of risk of death but also in terms of long term consequences of injuries. For example, this explains why nowadays injuries to the lower limbs are considered to be more important than they were in the past, not because there is any risk of death but because the consequences in the long term are quite important. It is also a difficult question in biomechanics because some of the injuries if treated correctly may have almost no consequences, or they can be fatal, or some others may have more long term consequences, so the choice is not easy to make. But in principle, most measures that are proposed are aimed to protect half of the population from a certain degree of severity of injury.

Regarding the models for biomechanical research, there are several and the aim of their use is to reproduce in biomechanical tests the behaviour of living humans. It is clear that probably the first model which will produce the best results is the human volunteer, but this necessarily involves work at a pressure level which is below injury level and then this does not provide information on human tolerance. Work using human volunteers includes that on boxers who were equipped with accelerometers. That situation might be close to the injury level but nevertheless it would be hoped that it remain below severe injury level. Some other work has been done with volunteers, especially in the USA, in terms of neck behaviour and head motion. So human volunteers are seldom used, but they are very important for some specific areas — the neck is one of them — because if tolerance data is being looked at, it is important in the design of a dummy to reproduce the correct kinematic of the head and then the neck is the interface, it is really the key for that kinematic.

The second model, which is used more frequently, is the human cadaver and in terms of anatomy and the distribution of mass, it is probably the best model for work at the injury level. But there is some limitation in using cadavers and care is needed in interpreting results of the relationship between injury and physical parameters, for instance. This is, firstly, because the population of cadavers is in general older than the population at risk. The cadavers used in different countries in Europe, as in the USA, are the bodies of people who gave their bodies for scientific research, but the age distribution shows that many are in the oldest age group of 70 years or more. This means that in selecting cadavers, there is a risk that the cadavers do not really represent the part of the population at risk that should be reproduced in the test. So selection procedures must be developed to ensure that the experimental results are not completely different from those applicable to the population at risk. Also there is a risk that the person had a disease which may have affected resistance to the impact. There is another difficulty, for example, if tolerance to bone fractures is considered, that some of the other tissues, just because they are dead tissues, may have different force deformation characteristics from living human tissues. This is mainly because the physiological functions controlling the internal physicochemical balance has stopped with death. So some of the tissues may have decreased tolerance or, more commonly, different force deflection characteristics. Then if those characteristics are wanted for use in development of a dummy, they may be wrong. Nevertheless, up to now and currently, most of the data used in biomechanical research for dummy design and protection criteria, are based on cadaver research.

The third model is the living animal which was used up to 10 to 15 years ago but, perhaps for different reasons, this has been stopped in France and also in the USA. This has stopped the improvement of knowledge in some specific areas such as brain injuries. Animals were very interesting to study. For some injuries, especially brain injuries, there is no other model because the cadaver and human volunteers cannot be used. But this has been stopped and the knowledge needed to improve safety in such matters has also been stopped at the same time.

Then there are mechanical models, or dummies, and mathematical models. What must be remembered is that these models are really created by scientists. The design of dummies or other mechanical models, or of mathematical models, is really only the transfer of data obtained from biomechanical knowledge. So what is needed first of all is basic research in biomechanics so that it is possible to design the mathematical models or dummies. Nevertheless, the mathematical approach especially is used increasingly because more and more is known in terms of human response and also because computer science is growing faster, with increasing capacity for data and decreasing costs.

So this confirms that there is not really one model which can cover all the field of biomechanics but a step by step approach is needed and the putting together all the data that can be obtained from different approaches if there is to be sufficient knowledge for dummy design and for global tests to check the safety of vehicles or their components.

To go on now to the approach which has been used, especially in Europe, in a joint project to develop a full scale test procedure for side impact protection: this is not as advanced as in the USA, but in Europe discussions are in progress and an example of how this global approach was used, with accident data as well as biomechanical data, to develop a complete set of test procedures.

One part of that test procedure is a dummy and an almost completely new dummy has been developed which is called EUROSID. There was EUROSID Prototype and then EUROSID I. In collaboration between four European institutes, the TNO, TRRL, INRETS and Peugeot Renault, specific components have been developed which should be human-like — that is, they have bio-fidelity — and behave, as far as is known, as a human behaves in an impact (Fig. 6). For example, the neck was developed by Peugeot Renault who took into account the tests made with volunteers in New Orleans, and tried to reproduce the same force deflection characteristics. The prototype stage of the dummy in a test situation is shown in Figure 7. This dummy is now used in different countries in Europe as well as in Japan and North America.

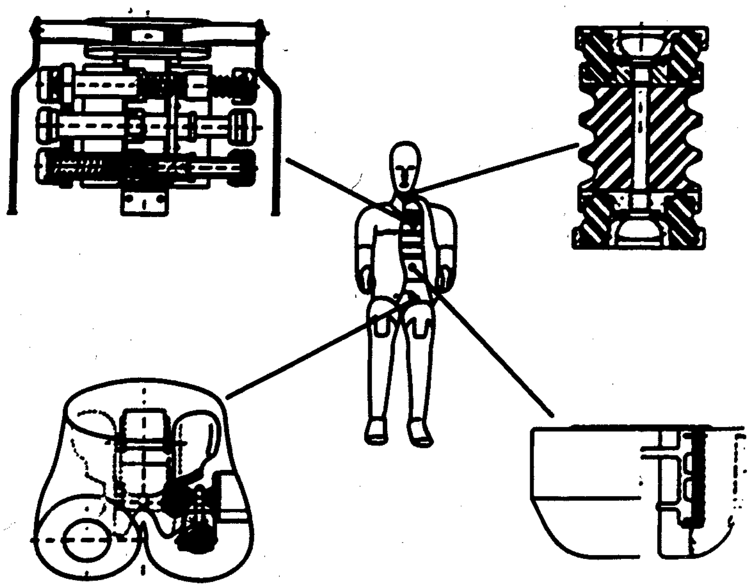
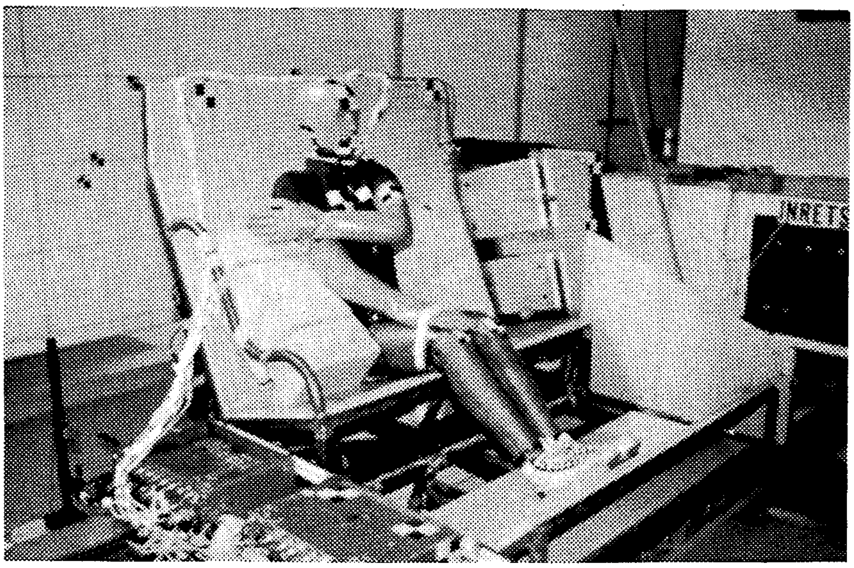


Figure 6: Overview of specially designed body parts of EUROSID:
thorax (top left), neck (top right) pelvis (bottom left), and abdomen (bottom right)

Figure 7: EUROSID prototype in test position



The main components to be discussed are the thorax, the pelvis and the neck and these should react the same as the human in different tests which were already conducted or were specially conducted for that program. For example, Figure 8 gives a corridor for deformation characteristics of the human thorax, which was the objective to be fulfilled by the EUROSID dummy. The corridor for the abdomen in terms of penetration and deflection is shown in Figure 9 and this gives to some extent the image that has been kept for the injury parameter. With the corridor for the pelvis (Fig. 10), it was attempted to put the dummy response inside that corridor, which was constructed from a cadaver test in which the cadaver pelvis was directly impacted by a rigid impactor or a padded impactor, to provide data to set up such corridors.

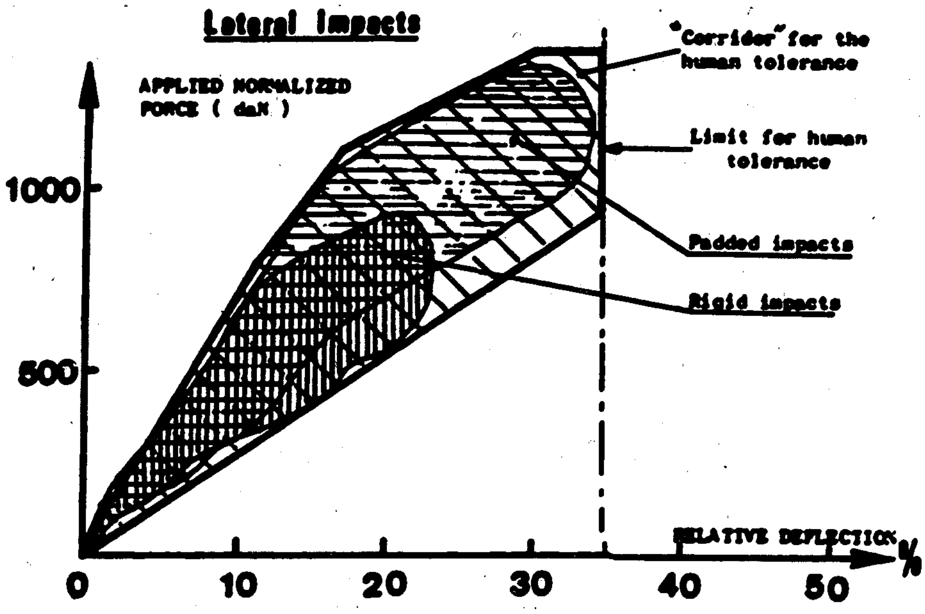


Figure 8: Force/deflection characteristics of the male impacted thorax

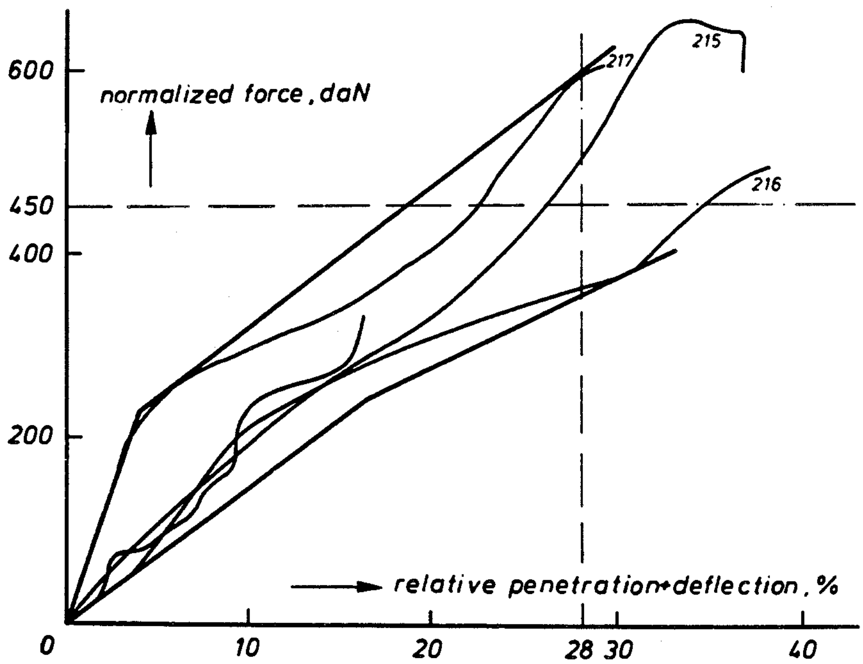


Figure 9: Force/penetration+deflection characteristics of the impacted abdomen

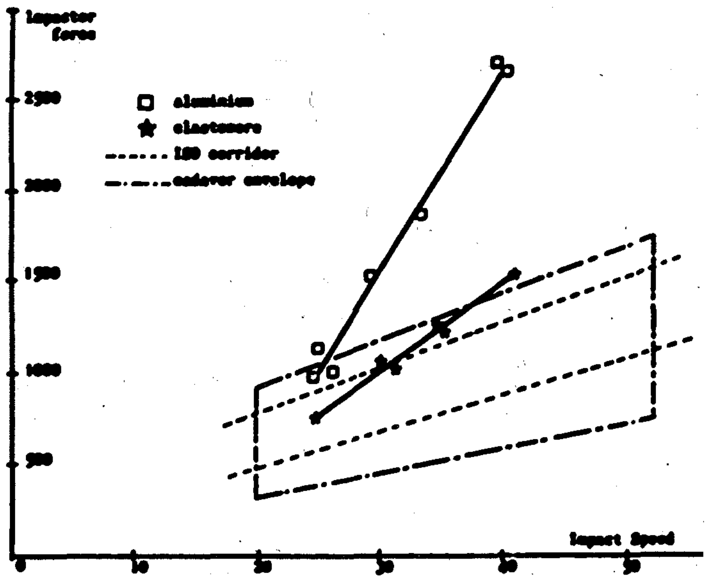


Figure 10: Improved EUROSID response for the impacted pelvis

The other part of the procedure was the test procedure. The characteristics of the barrier, which is the deformation face, are based on the characteristic of the stiffness of the front of the common cars in Europe. As well as the weight of the barrier, the test conditions are based on statistical analysis of accidents to determine the most frequent and severe accident conditions. These have been shown to be nearly 90° impacts on the passenger compartment (Fig. 11). Also, the positions of the dummies which were on the impacted side also came from accident analysis. This then allowed the development of a complex approach to full scale tests for side impact, of which the objective was really to improve the protection of the occupants of the crashed vehicle and to provide reference to the industry in terms of what should be passed to get such improvement in protection. That this really is a way to improve safety has been proved by a large amount of testing in different laboratories.

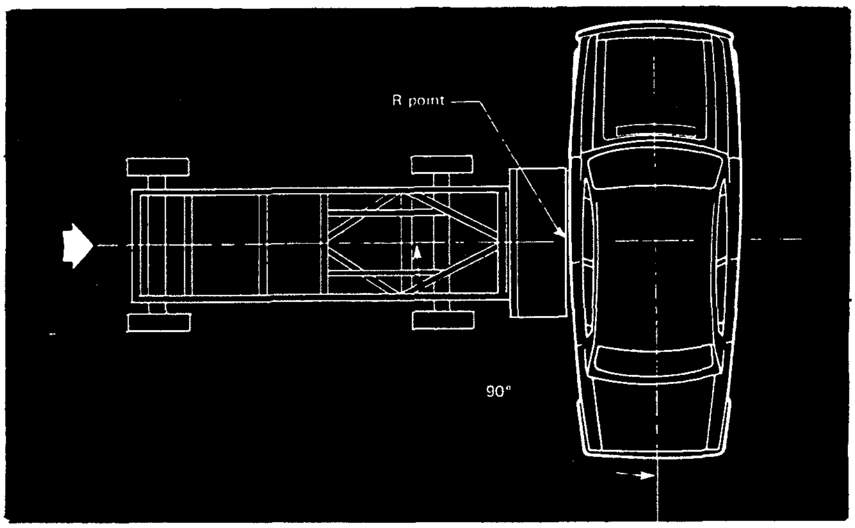


Figure 11: Configuration of the EEVC side impact test procedure

Finally, a reminder that passive safety is a continuous improvement that is already proved from last year: it is possible to propose better protection to road users, not only to car occupants, but also to pedestrians for example, or motorcyclists. Improvement is continuous and once a step has been passed, there is no going back — it's an irreversible improvement.

QUESTIONS/COMMENTS:

Jack McLean: As is apparent in what Dominique Cesari had to say, from the development of EUROSID and SID and BIOSID, we have three side impact dummies around, and there is still considerable debate as to which is the approach to be preferred. Do you care to comment?

Dominique Cesari: It's difficult — I cannot be objective to make a comment on that. What I would say is first, it's not a good situation. I mean we think that the model of the living human used in different parts of the world should be the same. And the first comment I would make is that there is some work going on now to produce an improved frontal impact dummy and this work is really a joint European and American project. We have already set up working groups to work together, to agree on specifications for such a dummy and this seems to work quite well. What we want is not to duplicate that again in the future and as far as the scientists can agree well enough, we hope that those rule makers will also agree later to have the same harmonisation of specification around the world.

Coming back to the side impact dummies, it was difficult for us in Europe to follow the procedure because the USA had already issued a DOT dummy and we considered that, with the knowledge we had about 5 years ago, we could do better than this dummy and it was for this reason that we developed the EUROSID program. But at the same time, General Motors developed BIOSID which is, I think, very similar to EUROSID, and I think there is still a possibility of harmonisation of dummies concerning the parts which have the most biofidelity from the different dummies. But there are not only scientific aspects in this. For scientists this can be done, but as there is already a Notice of Proposed Rule Making (NPRM) in the USA and the proposed procedure in Europe, it seems a little difficult.

Keith Seyer: Just to follow on from the different dummies in side impact, the rules are also different: regulations in the USA are different from the European regulations in a number of important areas such as the mass of the trolley, the deformation face and the injury criteria. Could you comment on how difficult it might be for manufacturers to comply with both?

Dominique Cesari: I'm not sure it's so difficult to comply with both. Probably in the future, it will be which one is more difficult to comply with, but I don't think there is really any problem. Some of the difference can be explained quite easily. For example, the car population in the USA is not the same as in Europe and you could imagine that resulting from that, the mass of the trolley, the mass of the barrier, or even the stiffness of the barrier should be different in Europe from in the USA.

For the dummy and injury criteria, it's really a question which should not occur any more. It's more difficult to understand the differences. For injury criteria one may need a compromise between what we consider the best in terms of biomechanics or, for the dummy too, what is reproducible and sufficiently sensitive to the input charge. So maybe we can have a different approach on that, but scientists should at the same time agree on the common basis.

Donald Simpson: I was interested in what you said about fractures of the pelvis. For the elderly, of course, it's what one would expect from clinical experience. Do you have many cadaver data on skull fractures in the elderly, because one has the impression that osteoporosis is not such an important factor in the skull as it is in the pelvis or the long bones?

Dominique Cesari: I don't have any data with me. I know there have been some tests in which at least a disc of the skull has been removed and carefully analysed. These tests were done by Peugeot Renault a few years ago in an investigation of brain injuries which looked for any correlation between skull characteristics and brain injury. Even if they did not do so, it would be possible to relate the characteristics of the skull bone to the age of the cadaver in terms of, for example, density of the bone or thickness of the bone and so on. But as far as I know this has not been done.

Jack McLean: Earlier, Ken Digges made a remark in relation to the development of the 3-point belt by Nils Bohlin of Volvo, that this was one of the major public health advances. Sometimes I think we don't appreciate just how important these measures are. Some years ago I tried to calculate very roughly the number of lives saved by the introduction of longitudinal restraint in door latch design in the 1956 model year in the United States. This change resulted from the work of Automotive Crash Injury Research (ACIR) at Cornell University Medical College. This was the first research group to identify ejection from the car as a leading cause of death and of greatly increased risk of severe injury. The numbers surprised me and I think it is fair to claim that that change in vehicle design has saved more lives and prevented more disability than the introduction of the polio vaccine. And yet, while Sabin and Salk are household names, I suspect there are probably few people even in this room who have ever heard of John Moore and Boris Tourin, who directed the work of ACIR; ~~in~~ some of you have heard of Hugh de Haven who established the research program. The case is even more compelling, I think, for the introduction of the 3-point belt. Those of us who are aware of these things should start maybe to talk more to the general public about them, because there really is, as the Deputy Vice Chancellor said last night, a modern epidemic of trauma and the consequences of it are every bit as serious as those of the more traditional diseases. By the same token, some of the successes already have been very impressive.

NEED FOR THE STUDY OF CLARIFYING INJURY MECHANISMS BASED ON ACTUAL AUTOMOBILE ACCIDENTS
(Perspective on Crash Injury Biomechanics)

Koshiro Ono

The determination of the human impact tolerance is indispensable to the improvement of automobile safety devices against crash accidents. The tolerance of the human being against such impacts may be determined based on the following steps:

- clarification of the injury generation process by finding answers to the question of what types of injury would occur by determining the bio-mechanical responses (pathological, biochemical and/or neurophysiological responses);

- determination of details and severity of injury based on the human anatomical structure by finding answers to the question of what mechanisms cause an injury or injury pattern;

- determination of physical quantities and degrees of severity of impacts by putting the details and patterns of injuries in a proper perspective and identifying physical quantities of crash impacts corresponding to individual injuries.

Impacts on human bodies in actual automobile accidents are, however, apt to occur in various types of accidents and diversified forms of impacts. It is also necessary to take full account of the fact that various difficulties lie ahead against efforts to clarify the injury mechanisms when carrying out actual experimental studies.

In this regard, the subject of this paper will be presented by focusing on the actual situation involving automobile accidents and details of their injuries as a specific example of efforts being made at the Japan Automobile Research Institute (JARI) for the clarification of crash injury biomechanics. Some other relevant material will also be introduced. This discussion will be focused on issues related directly to the current problems concerning crash injury biomechanics. It is hoped that there will be forebearance if the presentation does not cover future perspectives.

Recognition of Evaluating Injury Severity by Pathological Changes

The results of traditional research concerning the clarification of injury mechanisms have been obtained from the relation between the injury based on a visual pathological survey and the physical quantity based on impact tests using cadavers which do not realistically indicate the changes in vital functions. At present, this method is considered as the best possible way to predict injuries on living bodies.

In experiments using living primates, injuries such as subdural haemorrhage can be studied, but pathological changes in vital functions such as this are not seen in cadaver impacts. In the case of actual accidents and in research on clarification of injury mechanisms by using live primate subjects, however, it has been shown that effects on the vital functions of a live body may often be worse. These effects are not only due to pathological change but also to synergism which results in the impairment of the central nervous system and the respiratory-circulatory system, causing a more serious injury.

The lower part of Figure 1 shows an anatomical diagram of the heart and aortic arch. The upper part shows the diagrammatic changes between the chest wall and the heart upon impact. In the case of chest impact to the steering assembly or the instrument panel in the car, serious injuries are frequently observed in the chest; these include rupture of large vessels of the aortic or pulmonary systems. The occurrence of injuries like the rupture of large vessels depends on the pressure conditions in the vessels. Of course, the injury severity also depends upon the degree of impact. However, the heart beat and the condition of the organ of respiration is important in the occurrence of injuries.

Whenever injury prediction is made using a cadaver, therefore, it is necessary to state clearly that a difference does exist between the response of the cadaver and the living body.

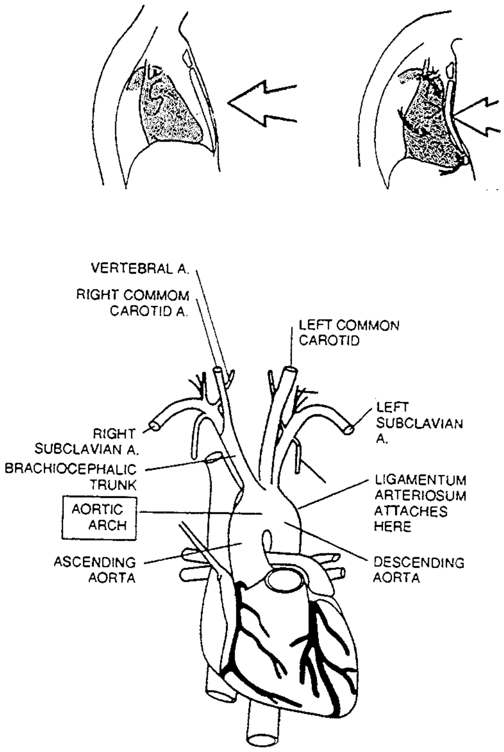


Figure 1: Diagram of heart and aortic arch (below) and of the relationship between the chest wall and the heart upon impact (top)

The Difference in the Tolerance Threshold Depending upon Age and Sex

The probability of serious injury by age and crash severity, where crash severity means the change in velocity (ΔV) during the crash, is shown in Figure 2. This data comes from 'Transportation in an Ageing Society' published by the Transportation Research Board in the USA, 1988. It can be seen that, for 20 years old or less, the probability of serious injury at 30 mph ΔV is around 42% whereas for 60 years old or more, the probability is 75%.

There was a similar tendency in the in-depth case study of accidents. In the distribution of driver/passenger(s) according to age, accidents that result in deaths or serious injuries often came from the groups of younger people (20 to 40 years old) and aged people of 60 years old or more.

From research on bone characteristics, substantial differences exist in the composition and strength of bones depending upon age, sex or the degree of skeletal morphogenesis and mineralisation. Figure 3 shows the relationship between age and the strength of the anterior cruciate ligament in the knee joint. The upper figure shows the relationship between the elastic modulus of the ligament, and age. The lower figure shows the relationship between the maximum stress (strength) of the ligament and age. In both cases, the decline in strength with advancing age is clear.

Therefore, it is desirable that the clarification of injury mechanisms of pedestrians and occupants who have suffered from accidents should elucidate the marginal conditions of fracture occurrence of the skull, ribs, femurs, lower legs and so on, depending on their age, sex, or degree of skeletal morphogenesis and mineralisation.

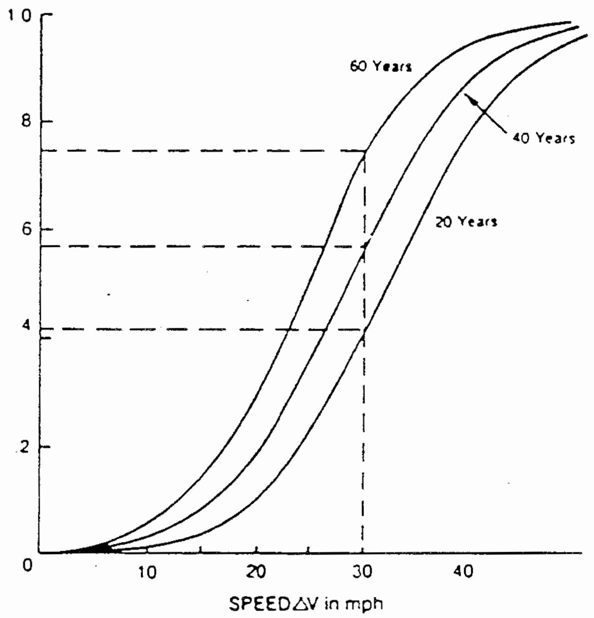


Figure 2: Probability of serious injury by age and crash severity

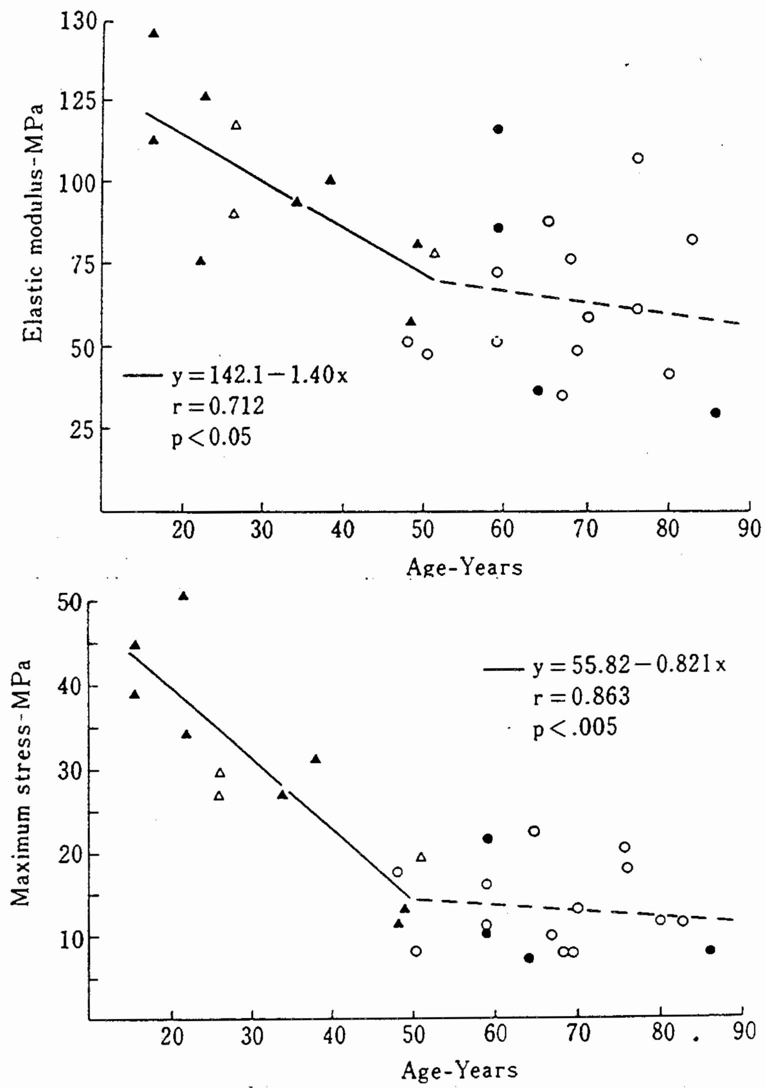


Figure 3: The relationship between the age and strength of the anterior cruciate ligament

Necessity of Providing Scales for Describing the Degree of Impairment and Disability in Addition to AIS

A flow chart of the necessity for providing scales for impairment and disability is shown in Figure 4. The Abbreviated Injury Scale (AIS) does not measure impairment or disability. Therefore the need for a scale that would complement the AIS and provide a link between injury severity and societal cost is fundamental.

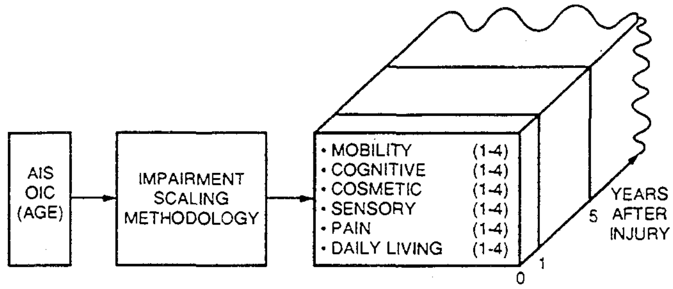


Figure 4: Impairment scaling methodology

In various publications concerning automobile accidents, the relationship between body regions and the degree of injuries shows significant differences in the rate of recovery even though such injuries are classified as being of the same severity by AIS. This applies particularly in the case of injuries to the knee ligaments peculiar to pedestrian accidents, from which the injured person's walking mechanism never does fully recover. Thus, the occurrence rate of sequelae is rather higher. This indicates that the functional loss of the human body is a quite different matter even if the degree of injury in terms of AIS is the same as injuries in other regions of the body.

The impairment scale consists of six parameters, namely mobility, cognitive, cosmetic, sensory, pain and daily living. These injury consequences depend on the passing of time. A framework for constructing an impairment scale has recently been proposed by John States and work on its development has been undertaken by the AAAM Committee on Injury Scaling. Therefore, considering the occurring conditions of the injury with sequelae, it is desirable to establish more accurate and appropriate scales to describe the degree of injury in addition to AIS. When the AAAM scale on long term consequences is available, it should be applied to the injury studies.

Regarding the Influence on Human Head Tolerance Threshold under Different Impact Conditions

During pedestrian accidents involving automobiles, the head can be struck against various parts of a car such as the windshield, A-pillar, wiper spindle, cowl top, fender, front end of hood, etc. Children, being short, have head contact points closer to the front of the hood than do adults. The head contacts, together with their respective areas of contact, shape, rigidity, etc., combine in such a number of ways to produce a variety of injuries. In cases that involved severe or fatal injuries, the majority were caused not by a flat surface, but by a part having a comparatively small radius of curvature and a relatively small area of contact.

At JARI the concussion level in primates has been studied in different impact conditions. There were 3 groups of impact conditions. Group A had pure translational impact over wide contact areas, Group B had rotational impact and Group C, direct impact combined with translational and rotational impacts. More detail on this will be given in a later talk on head injuries (p. 160). Averaged acceleration by duration for concussion levels in these groups of primates is illustrated in Figure 5. The primates' average acceleration is seen to fall on almost the same velocity line in the case of both Groups B and C. The duration of the former is relatively short. Further, a comparison of concussion levels showed that Group B animals required 3 times longer for recovery of the corneal reflex and about 5 times longer for recovery of respiration than those in Group C. Also, brain contusions were found in 70% of all fatal cases and 20% of all survivors.

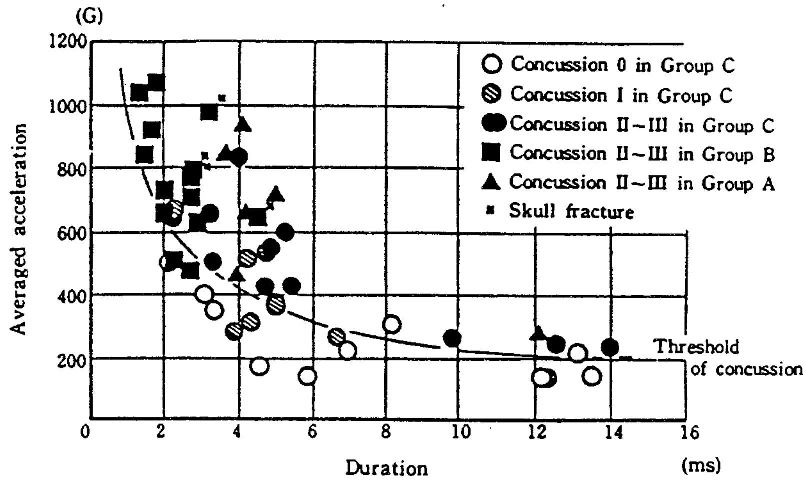


Figure 5: Comparison between averaged acceleration-duration of the monkey head and grades of concussion in occipital impact among 3 groups (A, B, C)

These findings on the cause of concussion and occurrence of brain contusions suggest that not only the pathological findings in the brain but also the neurophysiological findings must be taken into consideration as indications of head impact tolerance threshold under different impact conditions. In such cases, the degree of injuries to the head (brain concussion, brain contusion without or with skull fracture, etc.) also depends upon differences in impact velocity. Furthermore, in the case of a collision involving extensive rotational motion of the head, the conditions of occurrence of internal haemorrhage of the brain (subarachnoid haemorrhage, brain contusion, etc) due to rotational acceleration are naturally different from those due to translational acceleration of the head. A brief summary of possible occurrence of different types of brain injuries caused by severe mechanical input to the head is given in Figure 6.

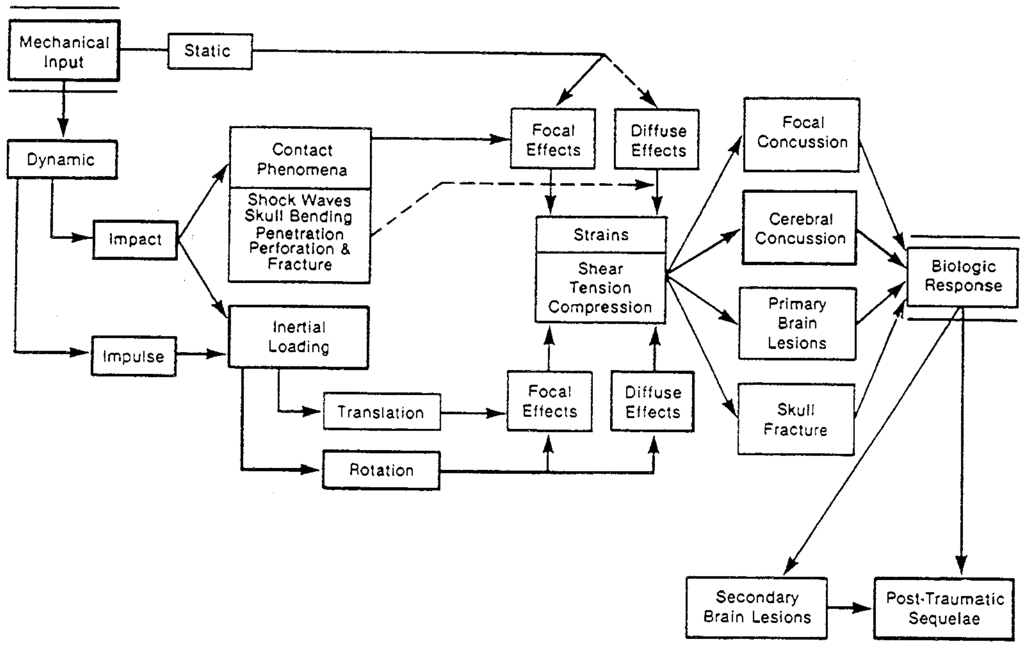


Figure 6: Head injury mechanics

Mechanical input was classified into 2 different types of impact, static and dynamic. For dynamic inputs there are 2 different kinds of loading impacts which lead to contact phenomena or inertia loading. Inertia loading is classified further into 2 parameters, translational and rotational. For these, focal effects and diffuse effects can be found and these different effects cause different injuries such as focal concussion, cerebral concussion, primary brain lesions and skull fractures. It is desirable, therefore, to explore and elucidate the mechanisms on the differences in head injuries related to such different impact conditions.

Clarification of Injury Occurrence Mechanisms of the Neck

A comparatively high percentage of fatal pedestrian accidents involve cervical injuries. Most pedestrians involved in fatal accidents die of injuries to the head and neck. In the case of an adult, assuming a normal behavioural pattern of a pedestrian during impact, the primary impact location is the lower body below the chest or the torso, causing an extreme whiplash motion of the neck. This is equivalent to a substantial impact force or torque to the cervical part of the body. In such a case, when the head is struck against the bonnet of a car, normally the shoulder is struck first and this causes an excessive whipping phenomenon wherein a substantial impact force or torque is transferred onto the neck.

The relationship of neck bending moment and head rotational angle for human cadavers and human volunteers in extension is shown in Figure 7. This is for a car occupant, not a pedestrian. The extension corridor for the human neck is clearly much greater than that for the cadaver example shown which is based upon the extension corridor. The 3 curves at the left hand side show the relationship of neck bending moment and head rotational angle in the different impact conditions according to the presence and height of the headrest. These results show the difference between the volunteer and cadaver with respect to motions of the head/neck system. The muscle tone of the volunteers must be taken into account.

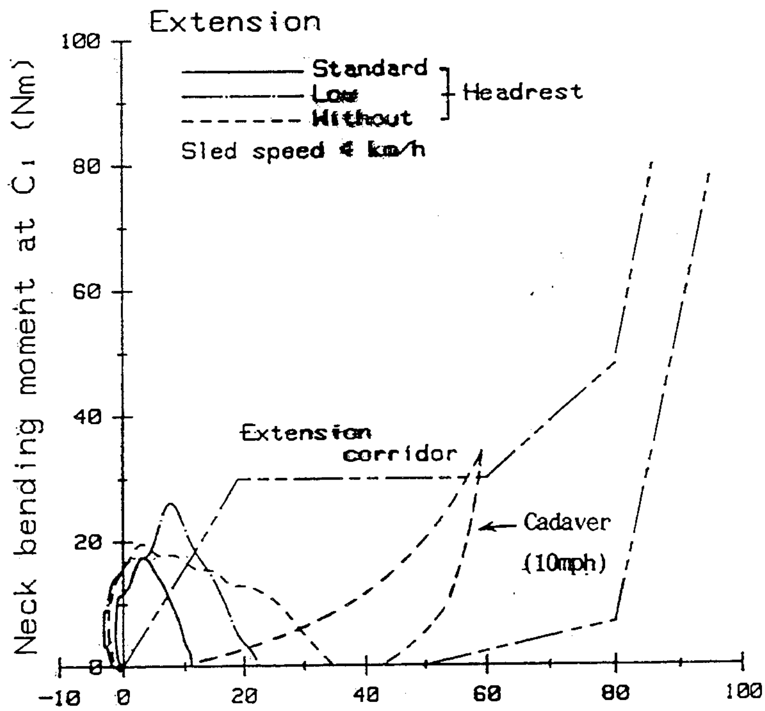


Figure 7: The relationship of neck bending moment and head rotational angle for human cadavers and human volunteers in extension

In the study of cervical injury mechanisms, the marginal conditions of occurrence of cervical injury are reported in relation to bending torque both in flexion and extension, and the subsequent bending angle of the neck. The positions of head and neck which can lead to cervical spine injuries are illustrated in Figure 8. However, further study is needed on marginal conditions of the occurrence of cervical injury even in lateral bending and/or in a twisting direction, including the shear force and total force.

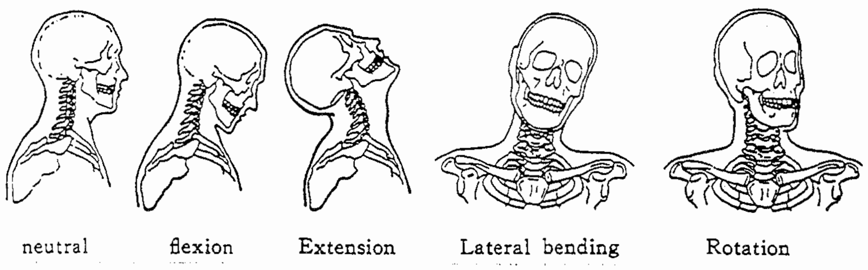


Figure 8: Positions of head and neck which can lead to cervical spine injury

Clarification of Injury Mechanisms of the Limb

Figure 9 shows the lower limb injury patterns caused by pedestrian/vehicle impacts for children and adults separately. The different impact positions depend on the different statures of children and adults.

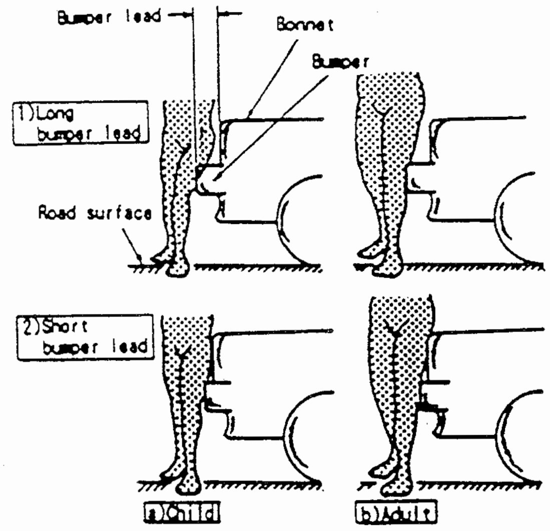
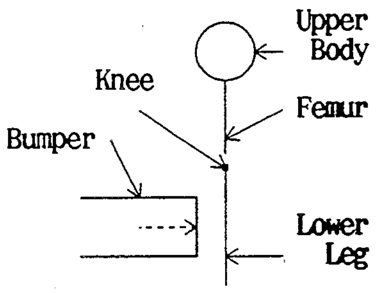
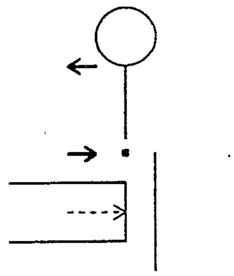
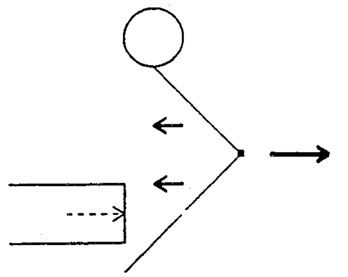


Figure 9: Lower limb injury patterns caused by pedestrian/vehicle impacts for children and adults separately

Injuries to the lower limbs are particularly notable in pedestrian accidents. Injury mechanisms of the lower leg and knee are summarised briefly in Figure 10. This is from work begun recently. Different phases after impact have been distinguished. The first phase covers contact injuries which include damage to soft tissue and nerves, and contact injuries to bones of the lower leg. In the next phase, shearing injuries to ligaments and fracture of the tibial intercondylar eminence occur. During the next step, the bending injury phase, there are ruptures or avulsion of ligaments of the knee joint and fractures of condyles of the femur or tibia.

It is said that injuries to the lower limbs are not fatal unless they are accompanied by profuse bleeding. However, there are many cases where the injured person, after recovery, suffers from sequelae at the knee joint which cause impaired functioning of the walking motion. From such a viewpoint, it is desirable that some scaling be established on the degree of injury depending on the sequelae in addition to AIS as discussed previously.

Phase	Contact Injury	Shearing Injury	Bending Injury
Mode			 Influence of Upper Mass?
Injury	• Soft tissue (Vessels) • Nerves • Contact Injuries (Fracture of the head of fibula and of the lateral tibial condyle, etc.) • Etra-articular injuries (Fractures of the diaphysis of the tibia, etc.)	• Ligament Itra-articular injuries (Rupture or avulsion of ACL, or MCL) • Fracture of tibial intercondylar eminence femoral cartilage injury	• Ligament (Rupture or avulsion of LCL, ACL, or PCL • Condyles fracture of Femur or Tibia
Force measured	• Contact Point • Contact Force • Contact Pressure	• Shearing Force and/or Deformation • Bending Moment and Deformation	• Bending Moment Transfered Through Knee Joint
	• 3 ～ 4 KN (Chalmer's ; Ref.5) • 4 KN (Porshe ; Ref.7) • 150 g (Ref. ?) • 7.5 kN (Ref. 2)	• 1 ～ 4 KN (Chalmer's ; Ref.5) • 5 mm/ 3KN (Ref.10)	• 70～100 Nm (Chalmer's) Ref.5) • 200 Nm (TRRL ; Ref.1) • 6 degree (TRRL ; Ref. 1) • 15 degree (INRETS ; Ref. 8) • 280 Nm (Ref. 3) • 320 Nm (Ref. 3)

MCL : Medial Collateral Ligamnet
LCL : Lateral Collateral Ligament
ACL : Anterior Cruciate Ligament
PCL : Posterior Cruciate Ligament

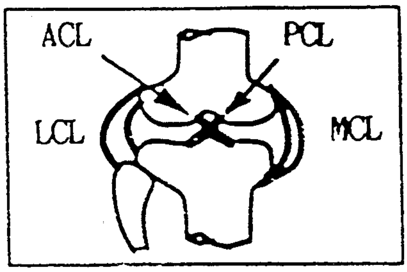
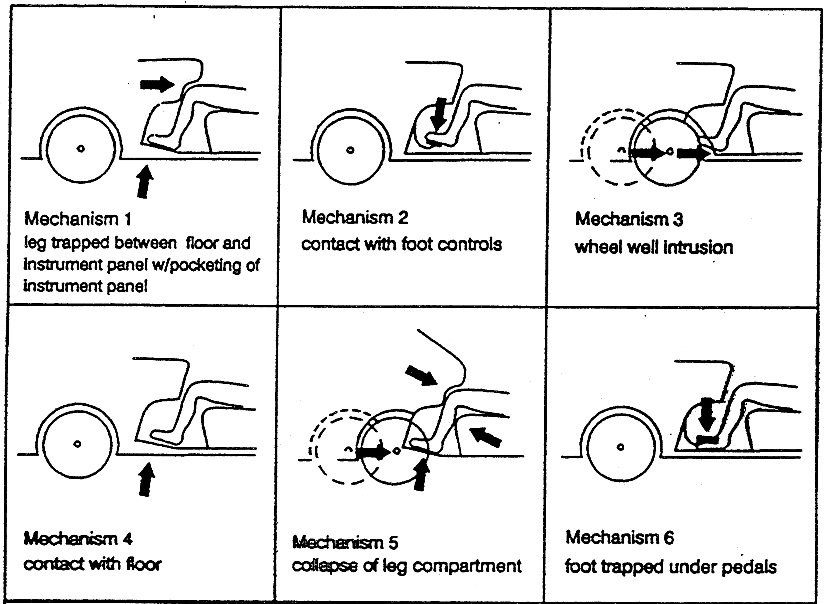


Figure 10: Injury mechanisms of lower leg and knee

The types of the injuries to the lower limbs of pedestrians involved in accidents are very different from those sustained by the driver/passenger(s) of an automobile. The six injury mechanisms associated with lower limb injuries to car occupants are illustrated in Figure 11. There are not many studies on the clarification of injury mechanisms involved in the conditions of the impact in pedestrian accidents. These include fractures of the lower leg or femur due to impact from lateral directions, joint dislocations and ligament injury from bending moments and shearing forces applied at the knee, and fracture and dislocation of joints at the collum femoris due to impact from lateral directions to the lower limb. Under such situations, it is desirable that the mechanism for lower limb injuries in automobile accidents be clarified. Differences in injury types have been found between accident studies and experimental work; it is important to understand the reasons for these differences. Injury type will depend on the position and stiffness of an area of impact; there is a need to investigate that problem.



	Mech 1	Mech 2	Mech 3	Mech 4	Mech 5	Mech 6	UNKNOWN	Total
Driver	7%.	43%	4%	12%	2%	1%	6%	75%
Passenger	5%	0%	1%	12%	1%	0%	6%	25%
Total	12%	43%	5%	24%	3%	1%	12%	100%

OCCUPANTS WITH ANKLE INJURY

	Mech 1	Mech 2	Mech 3	Mech 4	Mech 5	Mech 6	UNKNOWN	Total
Driver	4%	47%	7%	12%	4%	1%	4%	79%
Passenger	3%	0%	1%	12%	1%	0%	4%	21%
Total	7%	47%	8%	24%	5%	1%	8%	100%

OCCUPANTS WITH FOOT INJURY

Figure 11: Lower leg injury mechanisms associated with front seat car occupants

QUESTIONS/COMMENTS:

John Lane: Going back to the work you mentioned on primates, how was concussion defined in that study?

Koshiro Ono: In the case of the primate experiments, it was a little difficult to define concussion, but we can find the level of apnoea, the loss of respiration, and also the pattern of bradycardia. From this data, we can classify different degrees of concussion.

Jack McLean: I noticed in your slide of the distribution of head impact points for pedestrians, that most of the adult head impacts were around the plenum area, the windshield and the roof, whereas the head impacts for children were on the bonnet. This is the same distribution that we have observed in Adelaide for fatal pedestrian accidents, and I think it's of particular interest that the moves in the United States towards rule making for pedestrian head impact protection are dealing with the central area of the bonnet of the car which, with vehicles of today's dimensions, doesn't seem to be a risk area for the adult pedestrian at least. I wondered if you have any comment on that - perhaps Rolf Eppinger also?

Koshiro Ono: When you look at the impact speed for pedestrian head impacts on the bonnet, windscreen or A pillars, according to our in-depth studies, we have found that under 40 km/h, we can find the impact on the bonnet, but over 40 km/h it's a very severe impact, on the A pillar and the windscreen, the hard parts of the vehicle bodies. We have much argument about how to increase pedestrian safety by some improvement of the impact situation. This argument is now beginning at the ISO meeting (International Standards Organisation working group on pedestrian impact test devices) and also the EEC (European Economic Community) and in the United States. At ISO, the first steps focussed on the reduction of injury to the lower limb. And if we take into account the degree of reduction of injury for the lower limbs, we need longer lengths for the bumpers to produce more 'friendly bumping on the bonnet' for the pedestrian. In that case, probably when we focus on the head impact velocities, there would be small increases in the head impact speeds. So it's a competition, which one we select, which one is the more stable situation. So it's a very big argument.

Rolf Eppinger: I'd like to add a few things. The way we looked at pedestrian accident cases, we could predict quite well, within a range of about 5 to 10 inches, where the head would impact if we calculated the wrap-around distance, using the height of the pedestrian and starting with the foot at the base of the bumper. The pedestrian appears to wrap himself around exactly like a piece of string the same length, and to have a slight amount of motion relative to the surface of the vehicle, so his impact point would be slightly beyond what this predicted wrap distance would be. The wrap-around distance was considered in the process when we attempted to develop a proposal for hood impact testing. When we calculated benefits, those in the population that would exceed that were not considered. So yes, I think if you have very short cars, the wrap-around distance would predict that you'd have involvement with the windshield and A pillars and the headers. But when we considered the entire fleet - and maybe the American fleet is slightly longer, I have no idea, between the various populations - we still thought we had substantial benefits to be accrued by controlling impact response to the hood. We took the entire population at risk into consideration.

BIOMECHANICS RESEARCH AT NHTSA AND PRIORITIES FOR THE FUTURE

Rolf Eppinger

This talk will give an overview of NHTSA's biomechanical research directions for the 1990s. While there are a variety of Standards either in place or, like the side-impact Standard, in place and progressing towards being implemented, this talk will focus on the directions that the research should take after these Standards are in place, and what kind of improvements should be sought. Current Agency practice will be reviewed in terms of rationale and currently perceived deficiencies and how the current research efforts will address these issues.

NHTSA's biomechanics mission is seen as developing an understanding of mechanisms of injury for all significant body areas and developing measurement techniques to detect and assess the occurrence, severity and extent of these injuries — and I must emphasise that all three injury characterisations are important. Also, the relationships between the injury, mortality and morbidity must be developed. All these efforts are pursued to gain a better understanding of the types and significance of any alterations that are made in the safety performance of vehicles and, ultimately, allow NHTSA to rationally promulgate safety regulations.

To start with some simple definitions: *injury* is a physical harm or damage to a person; *trauma* is a bodily injury, wound or shock caused by application of energy in excess of physical capacities; and *mechanisms* are the physical processes by which a result is produced. Really, what is being sought when injury mechanisms are discussed are cause and effect relationships. These are illustrated by the equation:

$$\text{Injury}\begin{bmatrix}\text{Occurrence}\\\text{Extent or}\\\text{Severity}\end{bmatrix} = \begin{bmatrix}\text{Measurable}\\\text{Engineering}\\\text{Parameters}\end{bmatrix}$$

where injury, which can be defined as either the occurrence of a particular kind of injury, the extent of a particular injury, or the severity (risk-to-life or degree of impairment) of the injury is on one side of the equation and measurable engineering parameters, such as stress, strain, pressure, force, or acceleration, are on the other side.

To develop these relationships, the traditional approach has been to conduct experiments and observe both the outcome in terms of injury and all the corresponding mechanical measurements, such as time, force, distance and velocity, so that the types of relationships which are normally referred to as the injury criteria can be established.

The measurement techniques are developed because the physical test device resembling a component of the body is the crucial thing in this process. It could be a head, a chest, a leg or a complete occupant — that is, a dummy — that accurately duplicates the forces and motions of a human interacting with the safety system and measures the informative engineering parameters. Maintenance of that thread of logic between what injury is found and what engineering variables associate well with the occurrence of injury is needed in order to be able to predict or reproduce those variables within the physical test devices.

The reason for doing this is that the use of physical testing devices is currently the only acceptable method for requiring and verifying certain levels of safety performance for either regulatory or evaluation purposes.

For the evaluation and regulation of automobile safety NHTSA has a vast array of specified test devices. This includes the Hybrid II, Hybrid III and SID dummies, a Blak Tuffy for steering wheel impacts, motorcycle headforms to evaluate the performance of motorcycle helmets, and even a school bus headform which is tested by hitting the back of school bus seats to evaluate the performance. There is quite a melange of criteria used and, as the following list shows, they are not even consistent within a particular body area:

HIC $\leq$ 1000 (FMVSS 208, 213, 222)
Head g's $\leq$ 80 g (FMVSS 201 - interior)
Head g's < 400g, < 200 g $\vee$ t > 2 msec, < 150 g's $\vee$ t > 4 msec (FMVSS 218, helmet)
Chest g's $\leq$ 60 g $\vee$ t > 3 msec
Chest deflection $\leq$ 3 inches (Hybrid III)
Chest force $\leq$ 2500 lbf (FMVSS 203)
Chest TTI $\leq$ 85,90 (FMVSS 214)
Femur compressive load $\leq$ 2200 lbf
Pelvic acceleration $\leq$ 135 g (FMVSS 214)

Thus, in evaluating the head, there are criteria such as a HIC of 1000 in Standards 208, 213 and 222, which are for frontal protection, child restraints and school buses, respectively, but for Standard 201, which is another impacting device that goes into the instrument panel, the criterion requires that acceleration of that device be less than or equal to 80 g. Then in evaluating a motorcycle helmet, in Standard 218 there are actually three criteria, which set an upper limit of 400 g overall and lesser values for particular time periods. Clearly NHTSA's regulations have been developed in an evolutionary process and the criteria that were most appropriate at the time were used. There has been no unification of all the regulations. In the case of the chest, less than 60 g are required for any time greater than 3 msec; there is also a criterion instituted with the use of the Hybrid III, that deflection at a single point on the dummy sternum should be less than 3 inches for any time period; and within the 203 Blak Tuffy test, there is a requirement of less than or equal to 2500 lbs. For the side-impact Standard the Thoracic Trauma Index (TTI) now applies. That is somewhat explainable because it seems to be a different mechanism, but all the other criteria are frontal mechanisms, or frontal impact type tests, and there is again a collection of various criteria. Progressing down the body, the compressive load in the femur should be less than 2200 lbs, currently, and in the side-impact test there is a pelvic acceleration tolerance level of less than or equal to 135 g. It is desirable to bring a bit more coherence to this process and have more uniform types of performance standards that go across the entire board. Making protection criteria for the head, for example, requires a consistent evaluation.

The Thorax. In the case of the thorax, the rationale for controlling chest g's is Newton's Second Law which relates force, mass and acceleration.

$$F = \text{Mass} * \text{Acceleration}.$$

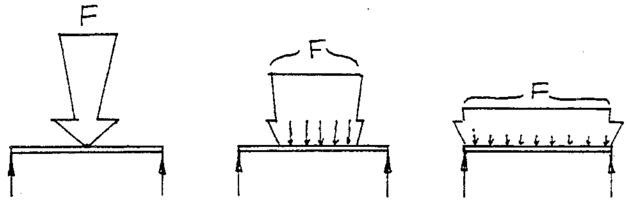
With lower accelerations, lower forces are predicted and if there is a crash condition or a restraint environment where the force distribution is the same, the structure will experience lower stresses and strains and a lower probability of failure. The justification for the G criterion was some experimental evidence from human volunteer tests and some cadaver tests. However, there is also contradictory information: in some pendulum tests where the chest was impacted with a 50 lb pendulum, fairly massive injuries were produced with very low spinal acceleration. So the criterion is not totally foolproof, but seems to be very specific in its application.

An example of the deficiency of the G criterion can be given by examining a very simple engineering structure, a supported beam, to which a force can be applied in various ways. The same total force can be applied as a concentrated force on the beam, as a semi-distributed load over part of the beam, or spread over the total beam (Fig. 1). In all 3 conditions, if the same load is applied, the implication is that the body will experience the same acceleration in each case. But the stress that is produced varies. Because of the symmetry, simple beam theory can be applied to show that the stresses that are produced at the mid-span are reduced as shown by the re-distribution of the load over half or all of the area. Just by modification of the area, the same readings are obtained for the G criterion, but there is a markedly different threat to the structure itself. So the same g produces markedly different stresses and therefore, in the design or evaluation of restraint systems, the benefits of force distribution are not seen if g is the only controlling parameter.

Deficiencies

$G \propto F$

(but no control of area of force application)



Gm = F   G2 = F   G3 = F

Maximum stress produced

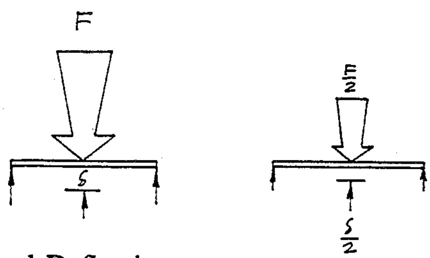
Sm = 1   S2 = 0.75 Sm   S3 = .5 Sm

Figure 1: Deficiency of the G criterion illustrated by an example of simple beam theory

With the introduction of the Hybrid III, an additional rationale, which was fairly well founded, was introduced. It was to monitor a deflection on the thorax.

Rationale: Deflection $\propto$ strain $\propto$ stress $\propto$ failure

Using again the simple beam model, if application of a force gives deflection and associated stress as shown in Figure 2, it can be said that the deflection tracks well with the stress. However, if the extremes of a concentrated load versus a distributed load are considered, when the load is distributed, the deflection is only 62% of that for a concentrated load whereas the stress is reduced by 50% (Fig. 3, top). So the deflection detects some but not all the benefits of force distribution a safety process offers: in this case it predicts a 37% drop whereas the actual drop is 50%.



Measured Deflection

Defl = 1   Defl = 0.5

Maximum Stress Produced

Stress = 1   Stress = 0.5

Figure 2: Effect of different forces applied at the same point

Another potential deficiency of deflection measurement can be seen in terms of the Hybrid III. Here the deflection is measured only at one point, and logically, it is on the midline. But in that case, if the same force is moved laterally, one half the distance off to the side, the deflection drops to 56 % of what it was but the stress drops to 75 % (Fig. 3, bottom). Thus, by using a single deflection criteria, the benefits appear greater than they actually are. One can almost hide potentially injurious loads on a device that only measures deflection at one point.

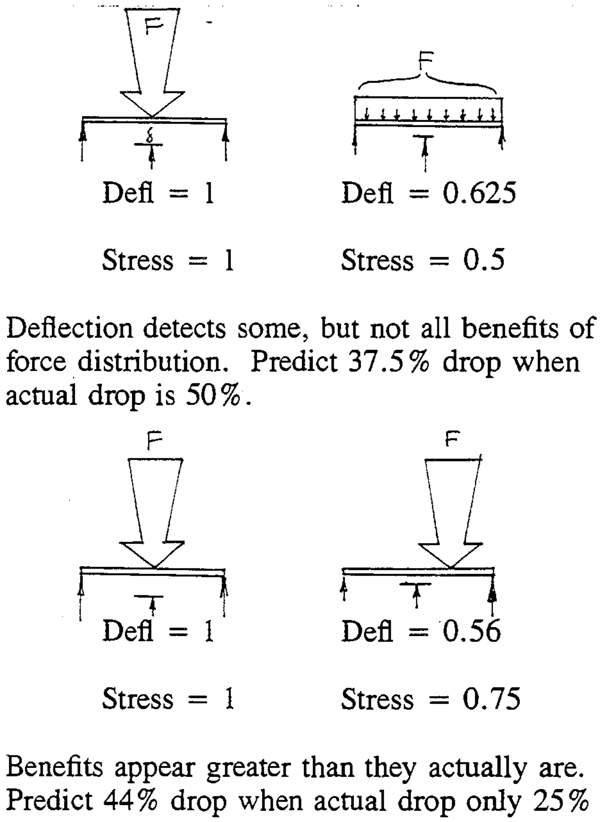


Figure 3: Deficiencies of deflection measurements — effect of concentrated versus distributed loads (top) and of different positions of force application (bottom)

Figure 4 shows the results of an exercise which used some of the cadaver data in NHTSA's biomechanics data base. Chest acceleration was restricted to a 3 msec clip, and results were compared for tests that used 3-point belt systems and airbags at 2 different injury levels — AIS 3 and greater, and AIS 4 and greater. Using a maximum likelihood process, a probability of injury as a function of g was developed for each loading condition. The results show that with the airbag, a significantly higher g environment can be sustained than with a belt system. This substantiates the notion that was discussed using the simple beam as an example — that is, distribution of load has an effect on the severity of chest trauma and it appears that current criteria are not sensitive to those changes.

Conclusions from this work are: that by setting a value on restraint performance with current criteria (60 g and 3 inches of deflection at one point), the injury mitigating benefits of distributing restraint forces over a larger area are not sensed; and localised loads, if applied away from the single deflection sensing device, appear to provide greater than actual benefits. Therefore, for a restraint designer who is under the pressure of designing a system or just trying to maximise the performance of the device being designed to current criteria, there can actually be some confusion and deviation from the real issue which is providing protection for the human.

NHTSA's current research objective in this area is to develop a universally applicable Thoracic

Injury Criterion that is capable of evaluating bags, belts and combination systems in frontal impact with a single type of process. In other words, NHTSA wants people to have airbags in cars and to use a seat-belt system, but also wants the designers to be able to design the best combination system to provide the maximum benefit to the occupant in this process. The direction of the Agency's research is to attempt to accomplish that.

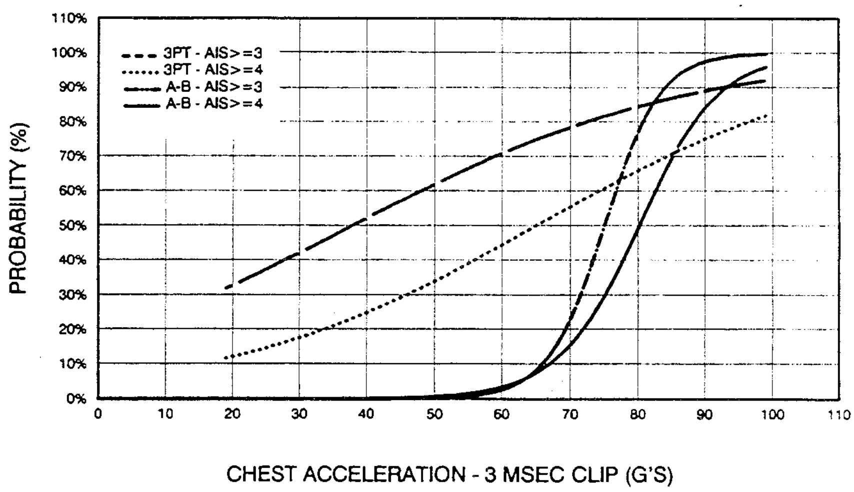


Figure 4: Cadaver tests with three point belts (3PT) or airbags (A-B)

The proposed methodology is based on the consideration that injuries to the bony thorax— fractures — can be described as failures which, from a theoretical point of view, are proportional to the local stresses and ultimately to the total geometry through the sequence shown in Figure 5. This is just the science of mechanics of materials, and failure of materials. Consider again the 3 different conditions: force being applied to mid span, force being applied off the mid span, and the distributed load. Curvature in this structure can be monitored along the beam, and if, for example, with a central load the curvature is normalised to 1, the curvature would be 75% at a position half way between the middle and the end of the beam, and it would be 50% if the force were distributed along the entire beam. The stress in the system would change in a similar manner. If a method giving such relationships were developed, it would have the capability of producing a criterion that, independent of the loading condition, would always track stress correctly by monitoring the curvature parameter.

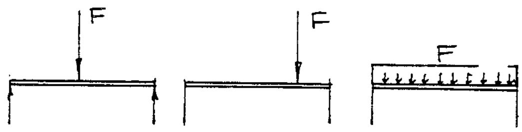
Failure $\propto$ Local Stress

Local Stress $\propto$ Local Strain

Local Strain $\propto$ Local Geometry

Local Geometry $\propto$ Total Geometry

How?



C = 1   C = 0.75   C = 0.5

S = 1   S = 0.75   S = 0.5

Figure 5: Effect of total geometry on failure in a system such as the bony thorax

For this work a chest band has been devised. This is a physical testing instrument which is a band that is placed around the subject and at a variety of points along that band, around the periphery, measurements of curvature are made (Fig. 6). In an impact event, for every instrumented location, a time history of the curvature is obtained. Then, using all of the individual measurements at a point in time the cross-sectional shape of the thorax at that point in time is calculated through an analytical process. By repeating at different time intervals, a sequence through time of change in the geometric form of the cross section is produced. An example of such results is shown in Figure 7; these are from actual cadaver tests, one in a 3-point belt system and one in an airbag system, with measurements at time zero, and then at 65 msec, 75 msec and 85 msec into the event. The diagrams show the progression of the deformation shape as a result of the belt, and it is obvious that because most of the load is applied by the belt, there is little force distribution and a very localised bending of the structure is produced. With the air bag system, which has been tested in exactly the same 30 mph impact condition, there is a significantly different type of deformation pattern due to the distributed loading it offers.

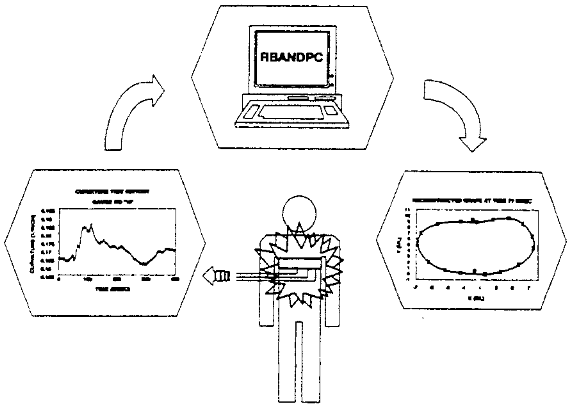


Figure 6: Methodology for evaluation of thoracic shape in frontal impact

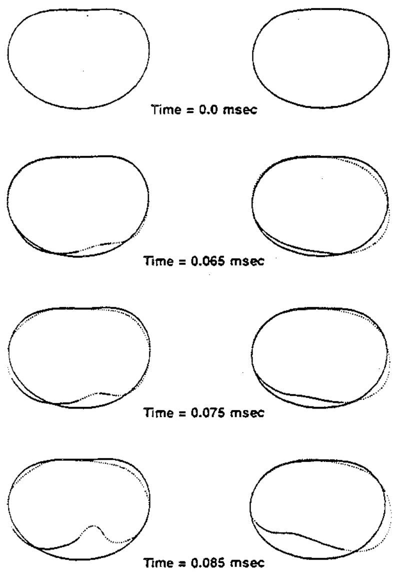


Figure 7: Thoracic shape reconstruction in cadaver tests

Thus, sensing the differences that various restraint systems can apply to the thorax is now possible. Typically, most NHTSA research tests are currently being run with two chest bands, one as high as possible on the thorax, and one somewhat lower that spans a little bit of the abdomen. Regional differences in a vertical direction can also be seen, so that a belt system that has a diagonal belt will produce a lot more penetration high on the thorax. If there is a 2-point belt system which comes down low and tends to push in the lower portion of the rib cage, it also can be sensed.

The uses to which this capability can be put may now be explored. One proposition is that the probability of failure of a rib at any point along its course around the body would be a function of the change of local curvature. Simple beam theory can be applied to deformation changes in the curvature around the body at a given spine location, passing from the spine to the sternum and back around to the spine. The type of curvature distribution around the periphery of the thorax while the body was being deformed, for a subject with a chest circumference of approximately 41 inches, is illustrated in Figure 8. Thus the current hypothesis is that the failure of the beam — the rib — is going to be proportional to the change in curvature from its original stress free condition, which is its normal shape, to that under deformation.

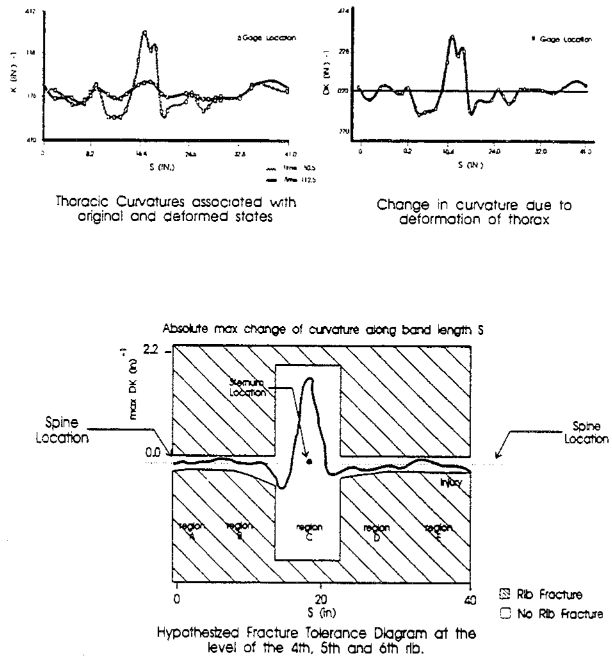


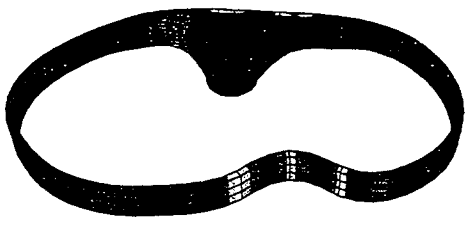
Figure 8: Curvature distribution results

Now the impact event is a dynamic event and so at each time at which measurements are made there is a different curve. Current work is aimed at establishing the maximum change in curvature at every station along the rib. Because the characteristics of the ribs change along their length — they are shaped differently at the back compared to the front — and there are cartilaginous attachments to the sternum, it is possible that the actual threshold for failure might be different at different points along the rib. Therefore a series of cadaveric tests with a variety of restraint systems is being conducted to look for the maximum changes in curvature and to plot them, in order to understand where and when rib failure has occurred. It is hoped that it will then be possible to determine a critical level of curvature change for rib failure at any point along the body. Eventually the critical threshold level for failure around the body should be determined, indicating where rib fracture should occur if change in curvature were exceeded. This experimental work is being carried out at three institutions: the University of Virginia, the University of Heidelberg, and the Medical College of Wisconsin. A series of about 80 cadaveric tests is planned, all run with 2-point or 3-point belt systems or airbag plus lap belt, at a variety of crash intensities. The age and sex of the cadavers is being varied as much as possible.

Analysis of the conditions which produced fractures and no fractures at certain sections of the chest should then produce a curve for probability of fracture as a function of change in curvature. Once these relationships are established, the more complicated process can be pursued using a band on a dummy, for prediction of where rib fractures would occur. The bands put a fairly severe instrumentation load onto the experimental environment because each band requires a minimum of 24 channels of high speed data collection, so they are unlikely to be used in a regulation. Therefore, given that there is a dummy and given that it deforms as cadavers do, it will be necessary to establish the minimum number of measurements necessary to detect and make an approximation of the mode shape of deformation that could then be related to the curvature process.

Alternatively, other forms of analysis of deformation data have been arranged. One method is called the slice model. This uses finite element technology to predict rib fractures and internal injuries. A cross sectional model of the thorax having representations of the spine, the ribs and internal viscera is created. Then a boundary value concept is used by forcing the model to undergo, in real time, the deformation observed by the chest band. The stresses produced by deforming the model could be calculated. It should be possible to explain both the bony and the soft tissue injuries using this technique.

Two examples, again of the airbag and belt test, using this method are shown in Figure 9. In this Figure, models of stress in the rib cage and strain in the soft tissues are shown separately. In the case of the rib cage the contrast between the 3-point belt and the airbag is clear in the original coloured results for these models: because the airbag deformation pattern is slightly different, the false colours that show stress are concentrated where deformation occurs, but are not as intense as in the case of the belt system. This model thus predicts substantial fractures in the deformation area, but with a lower probability than would be expected in a belted condition. The localised nature of these stresses is particularly notable. The loads applied by belt restraints are much more localised across the chest than those with an airbag. This is important for the future. If restraint systems using both bags and belts are to be optimised, which is of special interest in provision of protection for the aged population that has an osteoporotic condition, it will be necessary to harmonise the characteristics of the restraint systems, to get an optimal performance. If a test is run with an extremely stiff belt system and a powder puff type airbag, and 3 inches of deflection occur on the dummy, it would look just like a belt system because the bag would not get involved in this process — there would be none of the benefits that an area of expanding airbag provides. Therefore a test device that shows that this is not being done is needed.

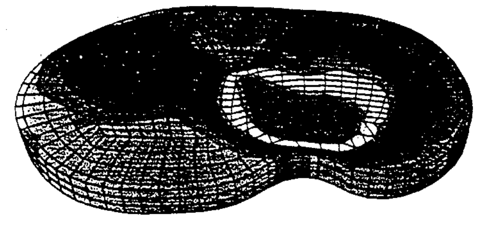


3-Point Belt Case

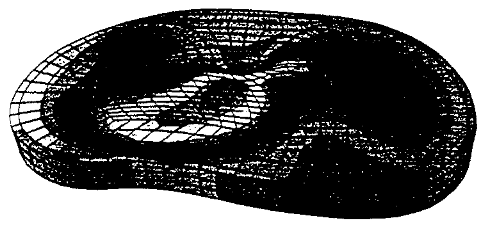
Airbag Case

Figure 9: Finite element modelling of internal rib stress

For a model which includes the internal tissues as well as the skeleton, the initial plan is to represent a uniform homogeneous, visco-elastic internal mass, not distinguishing discrete organs (Fig. 10). It is hoped that by using this uncomplicated method, the areas of high strain in the rib cage can be associated with the internal injuries that are seen. Then the empirical process will be carried out, of trying to link identified lesions in the cadaver with the stresses and strains that these models produce.



3-Point Belt Case



Airbag Case

Figure 10: Finite element modelling of soft tissue strain

A further stage is being attempted with a more totally predictive model. This has nothing to do with the chest band, but involves the making of a full 3-dimensional anatomical thorax model using the same finite element technology, the Dyna 3-D code, which is a marvellous invention: The majority of its use in the industry is currently for modelling crashing vehicles and it is now proving quite satisfactory in modelling human anatomy. As before, one of the purposes is to predict both the surface and internal injuries and to predict the interaction of the thorax with an actual restraint system, using a model of a belt system or an air bag, or with side door padding, or with any other feature of interest. The need is to have an analytical representation of human structure that can give as much information or even more, possibly, than is obtained experimentally.

To do this, a 3-dimensional model representing a thorax has been created. It is anatomically and geometrically a duplicate of a human thorax and is to be integrated into a restraint environment. There have been a lot of developments in relation to the airbag and it is possible now to model an inflating airbag interacting with a model of either a human or a dummy. So the technology is advancing at an extremely rapid rate. NHTSA is investing in the human model side of it because it is not necessary to contend with model changes, leaving the industry to develop its models of vehicles. An example output of this is shown in Figure 11. This modelling is based on the pendulum impacts done at General Motors many years ago, and the aim is to ensure that this model responds just as the cadavers did in those tests. The impactor and the ribs are obvious. The rib cage is made up of multiple structures: there is a complete skeletal structure with a spine, ribs, cartilage and intercostal muscles. The inside is filled up with homogeneous visco-elastic material as mentioned earlier. By appropriate computer manipulation, the skeletal structure can be exposed as in Figure 12 and is seen to have the same sloping ribs and overall geometry as the human thorax.

In practice, the modelling process is begun by placing the thoracic model in front of the pendulum: an initial velocity is specified for the pendulum, and the model is allowed to run by itself; it calculates everything that is going on in the process, such as all the interface forces and pressures, so that it can be watched as it goes dynamically through time (Fig. 13). Developments to allow exposure of this model to a lap belt and to interface this with existing airbag models are hoped for. When sufficient computer time and money are available, both belts and bags will be modelled together. Hopefully this will be an analytical adjunct to the experimental effort, so that interactions between the two can be made, because these models show a lot that can't be measured in the experimental world.

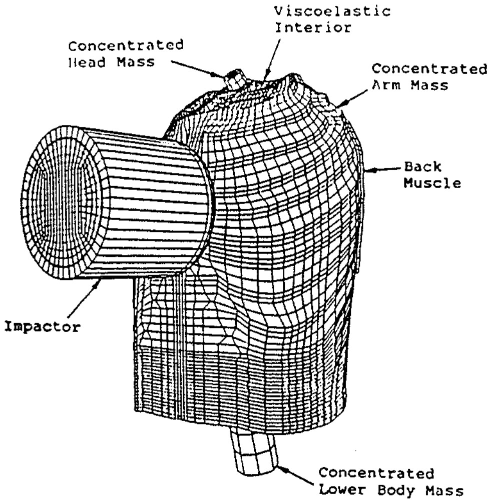


Figure 11: Rib baseline thorax model

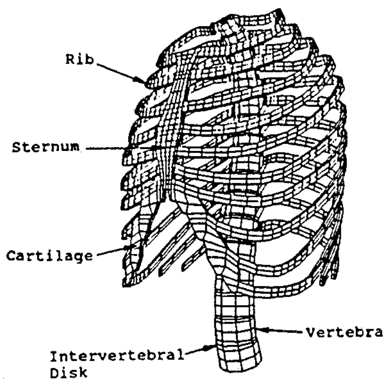


Figure 12: Skeletal portion of 12-rib baseline model

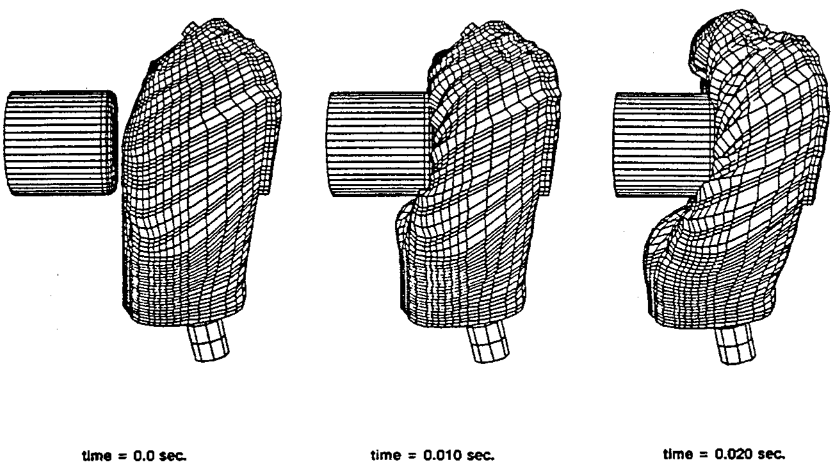


Figure 13: Side view of 12-rib baseline model under impact

The Head. In the case of head injury, HIC is used fairly ubiquitously except in 2 other standards at NHTSA. The formulation for HIC is given in Figure 14 and, as the sequence in this Figure shows, it can be related to change in kinetic energy, to how fast the change is accomplished and to the average acceleration during the change.

$$\begin{aligned} HIC &= \Delta t \, [\, 1/\Delta t \int^{\Delta t} A \, dt \,]^{2.5} \\ &= \Delta t \, [\, \Delta V / \Delta t \,]^{2.5} = \Delta V^{2.5} / \Delta t^{1.5} \\ &= (\Delta V^2 / \Delta t) * (\Delta V / \Delta t)^{0.5} \\ &= (\Delta KE / \Delta t) * \bar{A}^{0.5} \end{aligned}$$

Figure 14: Formulation of HIC

However HIC cannot determine the differences between various types of loading conditions, from very concentrated loads to very distributed loads, because suitable information is not built into it. The empirical association between HIC and skull fracture does not have the sensitivities to give the understanding of skull fracture and how to modify skull fracture that will be needed in the future. The same process discussed above for the thorax will have to be used. All the arguments made for the chest in terms of force distribution and so forth apply to a structure like the skull as well as they would to the chest. Therefore it will be necessary to create a way of measuring either the force distribution or the stress distribution over a structure like the skull, and monitoring its effects. The HIC does have some justification: there was an ISO U.S. position paper that Mertz-Prasad put together to show that there was a relationship between HIC and the occurrence of skull fracture. It's not a very tight fit but is used today only because it is the best thing available. But one of the deficiencies in the Mertz-Prasad paper is that it showed no relationship between HIC and cadaveric brain injuries. Therefore, even if a designer were to increase the distribution of the load over the skull to over 3 times the previous area, there would be no benefit for it unless the HIC was reduced at the same time. But if the load was distributed and HIC remained the same, the probability of fracture would decrease, but the designer would get no credit for it. So in terms of standards motivating designs, if there is no benefit from a change, what is the motivation for making it? That is the type of situation that needs to be changed.

In addition, a lot of study has shown that rotational motions have an effect on the occurrence and severity of brain injury. However, only in a very remote way can it be said that HIC is sensitive to rotational motions, and that is only because of the possible involvement of an $\omega^2 r$ term, so that very indirectly, changes in a rotational environment actually manifest themselves in the HIC formation.

To investigate the possibility of a more effective brain criterion, NHTSA has already done some work at the University of Pennsylvania with Thibault and his group, which has shown that neural dysfunction is proportional to some mechanical strain that is imposed upon an axon and that that strain is a function of the geometry and the motion that the skull and brain receive during an impact. This really implies that it is necessary at least to measure the complete motion of the head during an impact, and that includes the three translational and the three rotational components because they all have a profound effect on what goes on in the brain mechanically. These measurements can then be used as an input to a finite element brain model. So basically the brain model is stimulated exactly according to events during the impact and the stresses and strains within the model are then predicted. A skull fracture criterion is also desirable to understand the effects of force-time-area in the initiation of skull fracture.

Figure 15 is an output of NHTSA's current finite element brain model. The rigid skull, while used in the solution, is not shown. The brain model is of a uniform material that does not have all the detailed architecture of the brain in terms of ventricles and convolutions and so forth. As a first attempt at making something that is anatomically similar to the brain, the model does have a falx in it, but it does not go below the tentorium and there is no cerebellum at present. Gradations in colour represent fringes of pressure at an instant in time, after undergoing a prescribed motion, and are the types of pressures and the distributions of pressures that would be seen on the surface of the brain (the black and white reproduction here does not give an optimal view of these results). The model can be sliced vertically in order to observe the pressure distribution throughout the entire brain. The model also produces a variety of other engineering parameters such as stresses, strains and principal strains. So from the large menu, many aspects can be examined. Figure 16 is an example of a slice through the midline, again looking at fringes of pressure under an acceleration process, from right to left, and it can be seen how the distribution goes through the marked colours. But again, since a fairly reasonable geometry is now being incorporated, various parts of the brain are readily distinguished and it is clear that there is a fairly complex pattern of pressure distributions in the model.

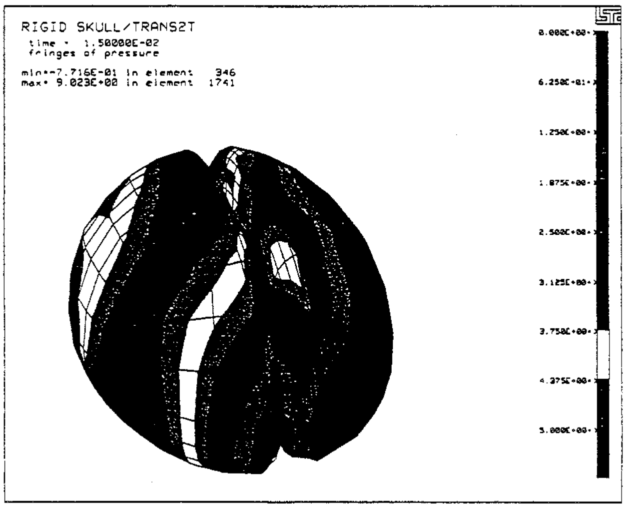


Figure 15: Finite element brain model

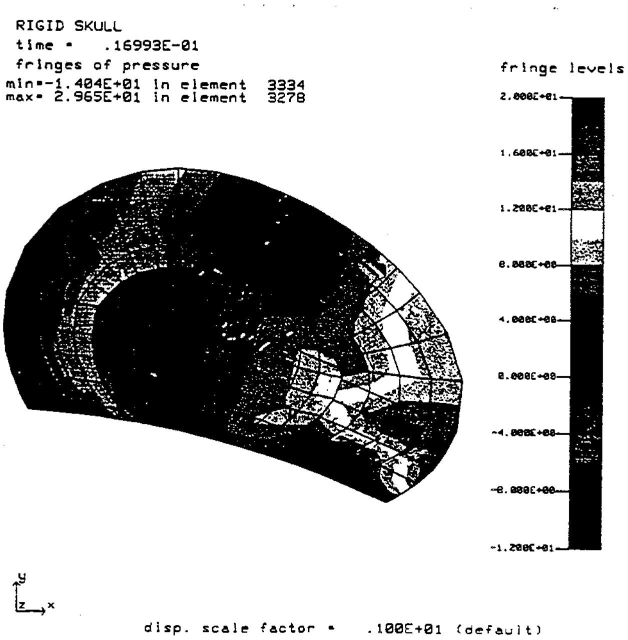


Figure 16: Midline slice through finite element brain model

Work with the developer of the finite element code is being carried out because of NHTSA's interest in the strain hypothesis of neural dysfunction. The new process monitors the entire brain to assess what percentage of the total volume of the brain exceeds different levels of strain. The results in Figure 17 show the relationship between the volume fractions of failed elements and time over the time period of the impact, for strain levels ranging between 5% and 14%. Identification of the lines within the brain model that have exceeded these different strain levels then leads to an understanding of both severity, which is the degree of stretch put into the brain material, and extent, which is how much volume of the brain is affected by these things. It is not known at present if this detail of the model is appropriate and it may prove necessary to weight certain areas of the brain if they are more critical than others. Those details still have to be learnt in the process.

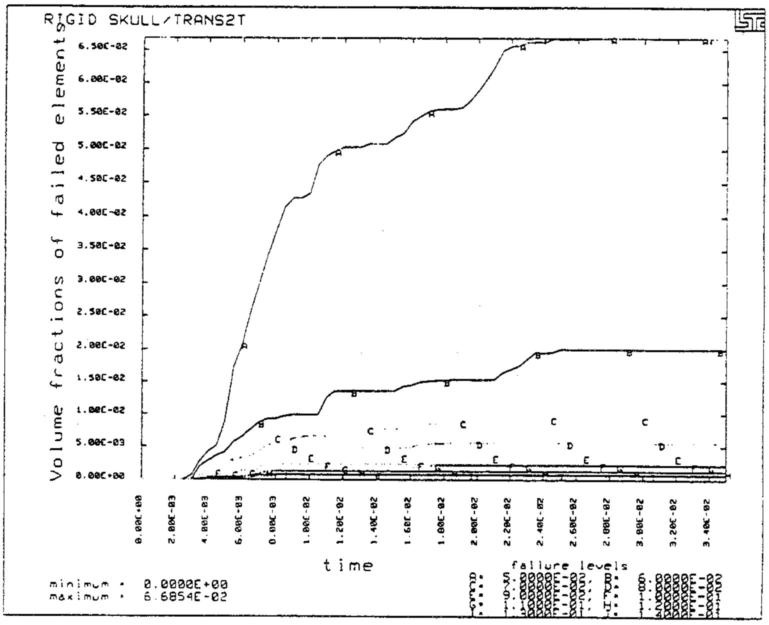


Figure 17: Relationship between volume fractions of failed elements and time during an impact

NHTSA's current research objectives arising from this work are as follows:

- to continue the development of the finite element model of the brain, to include the most appropriate geometry and material properties of the model;
- to use the current injury hypothesis to determine the extent and severity of brain injury as a function of rotation and translation;
- to initiate and conduct experimental efforts to support and validate the analytical efforts.
  This means animal testing and a solicitation by the Agency for proposals to do this has been made. Currently, animal testing is a very controversial process in the USA, and the matter of awarding these research efforts is being reviewed by the Secretary of Transportation. However, NHTSA considers such work to be absolutely essential for a proper understanding of the relationship between mechanical insult and physiological dysfunction and injury to the brain.

Neck injury. Currently, NHTSA is not engaged in any experimental work in this area, but there is some early planning underway. A neck evaluation procedure is certainly needed because there is a lot of neck loading. Proposals on the modification and padding of A pillars have already been announced and the issue of whether or not a new set of problems might arise as a result of head contact with the soft padding should be addressed. Mention has also been made about the potential of inflation hazards with unusually positioned occupants, and that certain forms of airbag restraints, if they're not designed correctly, could put substantial loads on the neck. These are reasons to create the technology to allow the restraint designer to create a device and a restraint system which do not have unexpected consequences that would not be uncovered until after about two years of road use. Clearly the neck is an anatomical structure that is at significant risk in that area, especially in rollover situations.

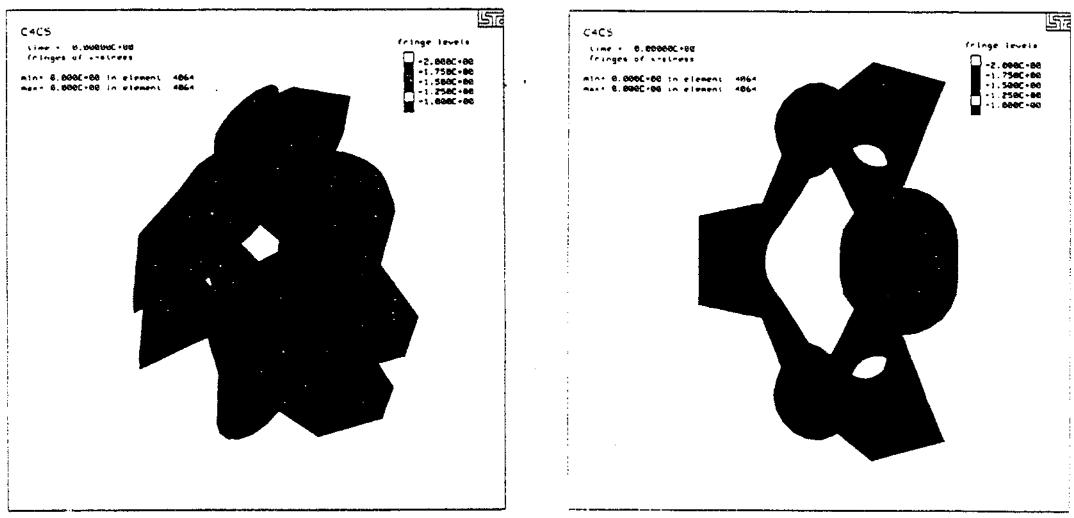
Such plans are at the formative stages. The literature shows that it is an extremely difficult area to work in. The variety of loading conditions that might be expected on the neck, and their sources, are as follows:

- an extension-tension from either the airbag or rear-impact;
- an extension-compression from rollover or A pillars;
- a flexion-tension from airbag or belt;
- a flexion-compression in rollover;
- a lateral bending in side impact; and
- rotation in complex type loadings.

In order to study these, a priority list will be established and appropriate investigations carried out .

To achieve this, parallel analytical and experimental work is advocated at NHTSA. The two systems have to challenge each other. The interplay between analyst and experimentalist forms a very synergistic process and should speed development of prototype necks and criteria that can be used with dummies.

One of the early attempts to produce a cervical spine model is shown in Figure 18. There are actually two vertebrae in Figure 18a and the spinous processes, articular facets, the main vertebral body and the neural canal can be seen in the original, though not clear in this reproduction. Figure 18b gives the view from above. The aim is to incorporate as much detail as possible in the model.



18a 18b

Figure 18: Early stage of cervical spine model under development

Crash Test Dummies. Current NHTSA crash test dummy development includes work on a variety of subsystems. One effort is in the thorax/abdominal area, and this is probably the most advanced. Others concern the neck area and the lower extremities, and eventually the head and face will be addressed. Also, instrumentation is being developed, notably a magneto hydrodynamic rotational acceleration measurement device that gives rotational velocity as an output, from which integration or differentiation then gives either rotational displacements or rotational accelerations respectively. The chest band has been developed. This was a significant breakthrough in terms of understanding the mechanics of chest injuries and these processes. It and its supporting software have been provided to a variety of research organisations. So the creation and maintenance of such unique items is one of NHTSA's roles. Other types of new instrumentation to go inside the new thorax have been devised. As the channel capacity requirements within the dummy are increased in the future, it will be desirable to avoid having a huge umbilicus with 85 channels which weighs 3 times the weight of the dummy, so concepts for onboard instrumentation will be sought that can acquire the data onboard the dummy and either record it directly and keep it onboard, or connect it to external recorders by small coaxial cables to reduce interference. So there is a demand to carry much more instrumentation on the device than is done at present.

The dummy development efforts were initiated about 7 years ago in a cooperative program between NHTSA and the University of Michigan, which aimed to determine the anthropometric shape and position of the seated driver. Nobody had done that before. Three average human forms resulted: a small female, a mid-male and a large male. The seated contour was also defined. The forms have now all been laser scanned and digitised and are the exterior shell to which the new dummy is being built.

Figure 19 shows the new thoracic and abdominal area of the new dummy and the various features being incorporated in this device. The standard Hybrid III neck and head are still being used but there is a completely new shoulder system in the form of a 4-bar linkage. It has a clavicle and the shoulder can move back and forth, similar to the way the scapula slides on the back of the thorax because of the 4-bar mechanisms. Evaluation of 2-point restraint systems that tend to intrude in the low part of the torso requires a chest that basically is anatomically similar to the real human thorax. So the ribs of the new chest are sloped and reach to the lower levels as seen in the human. The stiffnesses and other aspects of the device have been redesigned in an attempt to mimic regional loading of the thorax, according to some limited data from Wayne State University where the measurements were made. The thoracic spine is now a 2-piece assembly which has a rubber element in the middle to provide additional flexibility. The shoulder is like a wing with a pivot shaft and the clavicle is actually attached to the sternum of the dummy. So when the belt system is involved and there are large belt loads, the belt load actually pushes down on the clavicle and it, in turn, on the sternum. This contrasts with the Hybrid III which has a very large aluminium shoulder casting that shunts belt loads directly into the spine, which is very unhumanlike. With this new dummy, loads being applied to the clavicle will be directed either into the shoulder or into the sternum, so that the forces are beamed out correctly.

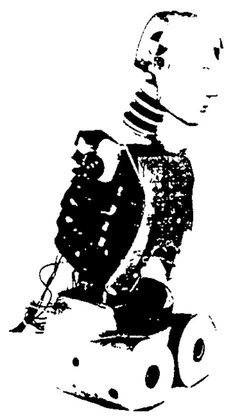


Figure 19: Thoracic and abdominal areas of new dummy

In the lower part of the thorax is a joy stick measurement device sensing deformation, details of which are shown in Figure 20. This is actually a high speed stringpot located between the spine and the base of the sternum. The string is within the telescopic structure and is attached so that it measures changes in the length. The telescopic structure is mounted on a doubly gimballed axis that monitors both azimuth and elevation of the deformation in a spherical co-ordinate system. Four points on the frontal chest surface are tracked x,y,z at present. It may be that 4 measurements are not adequate, but the system is a proof of the concept for this measurement technique. These devices have proved very satisfactory in that the stringpots are able to pull in the string as fast as the impact drives the chest wall, thus allowing measurement of localised loading that different types of restraints might impose upon them.

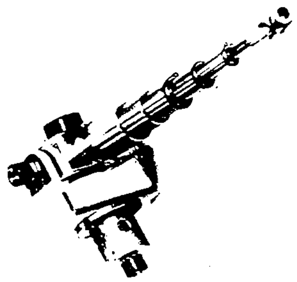


Figure 20: Joystick measurement device

Other Objectives. NHTSA has other research efforts underway in its Highway Traffic Injury Studies. These are detailed accident investigations where a hospital is aligned together with an accident investigation team and studies are made of particular subjects on which early information is desired, such as the performance of airbag and passive restraints, or child restraint systems. The time taken to accumulate a substantial amount of data from the normal NASS system is quite long and it is hoped to accelerate that a little with these efforts.

QUESTIONS/COMMENTS

Tony Ryan: I was very impressed with the work you displayed both on head injury and thoracic injury. I wondered, in that very elaborate model of the thorax being struck by the pendulum impactor, whether the movements of the ribs are similar to what happens in real life, and does it predict where fractures occur?

Rolf Eppinger: We have not run bands on cadavers so I don't know what the deformation pattern under the pendulum condition is. When we have run the bands on the cadavers, we will have to develop the belt model to apply to that thoracic model before any comparison can be made. But it is our intention to do that, and is considered part of the validation process. All of the points of validation will have to be checked. So rib motions, internal motions and locations of injury are not shown on the model at present. Also, the model does not simulate fracture in detail, but, given that the best constitutive relationships for the material are entered, it predicts the deformation. Now we've been having very good results in that it does match the force-deflection histories that we have seen in cadavers — that's why we did this pendulum test — which suggests that even though there are some fractures, they are not substantial enough to significantly alter the overall structural integrity of the chest. In other words, the trauma produced and a certain amount of consequence is visible, but the consequence is not a total destruction which alters the mechanical characterisation of the chest. Eventually we could get into that, but at least in the finite element model, to predict fracture and actually have the rib fracture is very computationally intensive and we hope we won't have to go that far.

Ken Digges: Would you comment on the basis for your clavicle design. Is there a set of human data or cadaver data that examines the characteristics of the clavicle and would you attempt to build it in? Also, how might it be desirable to change the loading on the spine?

Rolf Eppinger: We have the regional stiffness measurements derived from a 2 inch by 4 inch load applicator that was applied to cadaveric specimens either at the mid-line or on each side at 6 points on the thorax. Our concept for clavicle design was that, because we didn't see a lot of clavicular fractures occur, we wanted to make sure that the load was transmitted to the thorax correctly. Therefore if we built the chest that had the correct compliance and attached the clavicle in a manner geometrically similar to the human, we believe that the whole process would work together. We have tried to avoid having a structure that shunts the force away from the clavicle but rather create a structure that ,when a load was applied mid point on the clavicle, half would go into the sternum and cause compression of the chest. The current dummy doesn't do that because it has a large casting that's directly attached to the spine and shunts the forces away from the chest. You can hide the forces by understanding what the dummy design is.

Laurie Sparke: Would you care to comment on how you see the relative importance of bone fracture and soft tissue injury, particularly in thoracic injuries and head injuries, and finally, would you care to comment on the work that Viano has done on viscous injury criteria?

Rolf Eppinger: The last issue will be dealt with in a later talk (p. 89) In terms of the severity differences between fracture and internal injuries, in general severe internal injuries have an AIS rating of 4 or 5 — a rupture of the liver or some other organ is a very serious problem. Fractures are generally seen before internal injuries, so it is a kind of progressive process. With the continuing aging of the driver population, I would like us to justify and do some work based on fracture for the elderly, to make sure that the restraints that are designed produce a minimum type of fracture with thresholds well below those for the internal injuries. Since the population at risk changes as new protective systems are introduced, it's never certain what injuries will be important in the future — it's an evolving process. So, from the research aspect, I have to anticipate that we must be able to predict both the soft tissue injuries and the bony fracture type of injuries.

Laurie Sparke: I'm surprised that the focus is on bony fractures.

Rolf Eppinger: It is because the threshold for bone fractures is below that for the internal organs. If a situation is well below the fracture threshold then the life threatening injuries to the soft tissues are eliminated, because of that threshold process.

Bryan Knowles: I have 2 questions. Can you tell me what ISAAC is the acronym for? And also, in the work that you're doing with chest bands, which is fascinating stuff, is there a plan down the road to try and maybe work with car manufacturers' programs like MADYMO 3-D and 3-dimensional and total kinematic programs to integrate your models into them in order to study any changes you see in chest injuries and deflections as part of the total model of the vehicle structural changes, column effects, or airbag effects, as a total vehicle effect?

Rolf Eppinger: Yes, ISAAC stands for Injury Sensing Android for Automotive Crashes — which is just a nice play on names and it has a sort of Newtonian ring to it.

What we hope to do with ISAAC and the various components that we're developing is to avoid the difficulties that we had with the side impact dummies. So we're inviting a variety of the European Governments' and the Japanese Government's research organisations into this process. This is still in its formative stages but I believe we would like to have the cooperation of everybody in this process, so that the device meets agreed-to specifications and the desired needs of the various organisations. This, hopefully, will result in a world dummy, at least for the frontal portion. At present we are at the forefront of the design because we have started this process but we're not trying to do it to the exclusion of anybody. The device, currently as it's designed, is available from the dummy manufacturers and we are encouraging other people, if they have the budget and the interest, to look at this device and tell us what they like about it and what they don't like about it. We'd like to get those dialogues started and continuing as we go through this process of upgrading the performance of the device.

Michael Griffiths: At the Experimental Safety Vehicles Conference last year, Transport Canada was promoting some alternative injury criteria. The proposed criteria are simplifications: that a straight g measurement be used for head forces rather than HIC; that the deflection criteria be reduced from 75 to 50 mm in the thorax; and that for the lower or abdominal injuries, more attention be given to seat belt geometry rather than to try to make any transducer measurements. Do you have any comments to make on this?

Rolf Eppinger: Relative to the head, if it's strictly in the frontal-type 208 environment where you're attempting to avoid head contact, I would say that would be a possible criterion to pursue. However I have some difficulty with the proposal because if we have to do an A pillar evaluation, and we want to actually understand the performance of the A pillar during the impact, not just prevent the impact, I think we have to go to this much more detailed process. I think the detail is justified ultimately because we can probably produce more efficient and cheaper restraint systems in the long run if we have a better scientific base in terms of prediction of the injury.

BIOMECHANICS AND ACCIDENT INVESTIGATION IN THE DEVELOPMENT OF CAR SAFETY

Ingrid Planath

One hundred years ago there were not many cars out on the roads but certainly there were a number of other objects with which cars could collide such as trees, and unfortunately this is still a rather common accident today. The fine old cars of that early period did not have any of the safety features that are taken for granted today - there were no bumpers and no safety belts and although there must have been very good air conditioning there were no windshields that could protect people.

Gradually it was realised that in the event of a car accident it was not always the driver who was the one to blame but that many things could be done to increase safety both in the road environment and also in the cars. An interest in car occupant protection and also in biomechanics grew gradually as the number of cars increased. The creation of the National Highway Traffic Safety Bureau in the United States played an important role in that it initiated a set of standards that would control vehicle performance in terms of crashworthiness. In the meantime, car manufacturers had already started to work on safety as a specific design issue and a result of this is the laminated windscreen which could be found on the market already in the 1940s. And even if the old style of testing windscreens with a hammer blow was not a very sophisticated test compared to what can be seen on crash advertisements today, it served its purpose just to show what was good with the laminated design feature.

In the complicated traffic system of today with a mixture of many road users, it is human to make mistakes, but still the driver is not always the one to blame. The car and the road environment should be adapted to suit the abilities and the limitations of the driver and not the other way around. When traffic safety is talked about, 3 parameters must be borne in mind: these are people, vehicles and the traffic environment. The important thing is to make the interplay between these three as smooth and as easy as possible. The three factors are shown diagrammatically in what is usually called the Haddon matrix (Table 1), named after the man who thought it out. The three factors involved in traffic safety are shown horizontally and measures to improve traffic safety are shown vertically, split into three sequences, so that the periods before, during and after the accident can be seen.

Table 1 : The Haddon Matrix

	People	Cars	Traffic environment
Before	Avoiding accidents		
During	Preventing injury		
After	Reducing injury		

In looking at what can be done to prevent the injuries from happening in the first place, people can be considered first. The driver, of course, has to take the responsibility to be fit enough to drive although there are some basic rules that are set up by society — driver education, driver experience and awareness are spoken of — that the driver, for example, should not be drunk when driving.

Regarding the vehicle, it is the responsibility of car manufacturers to develop a combination of steering, braking, suspension qualities - which can be called road manners - to give the car a forgiving attitude if the driver should fail; an example is the anti-lock brake system. This is what is commonly called dynamic or active safety and it is anticipated that this is an area where more progress will be seen in future. The interplay between the driver and the surrounding traffic can also be made easier by, for example, convex side mirrors which widen the rear view for the driver. Also, the driver's local environment is important: a good climate and a comfortable temperature inside the vehicle are important for good driver performance; and also, to have the instrumentation at a convenient distance is essential to have safe control of the car. Regarding the traffic, traffic planning can be applied to avoid direct conflict between road users by separating them as, for example, with pedestrian over passes.

Such factors contribute to accident avoidance, but if accidents still occur, what can be done to prevent injuries? People need to know the safety features of their cars and how to use them — for example, the seat belts. This is a big reason for giving more information to the public. At a recent exhibition where the Volvo car was displayed with an airbag that had been stuffed with paper inside, many people asked for guidelines on how to fold the bag back once it had been used! Restraint usage will not be of high benefit, though, if it is not combined with a car design that will absorb as much as possible of the crash energy in crumple zones, thereby transferring little energy onto the occupants, so the occupants should be contained in some sort of safety cage (Fig. 1). It is also important to have a friendly interior that can cushion the occupant in the event of a contact with the interior. So padding material on contact surfaces, collapsible steering wheels and steering columns and also airbags are examples of this.

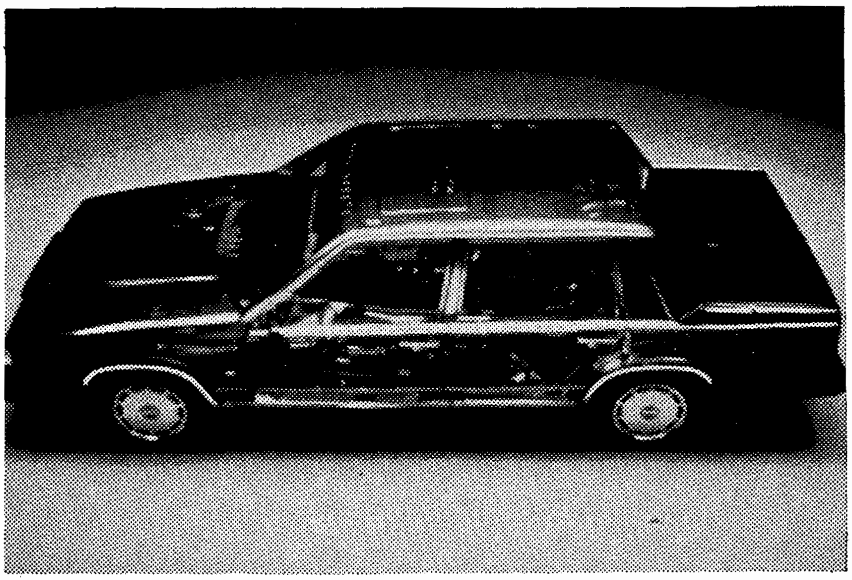


Figure 1: Illustration of energy absorbing skeleton of Volvo car

In the traffic environment, road furniture and the area close to the roads should have a soft energy absorbing design. It would be much better, for example, to deform the power pole than to deform the car.

Finally, what can be done to reduce the consequences of a crash? People can contribute by knowing the basic rules of first aid and practical actions to be taken in the event of a crash. Regarding the car, it should be possible to open at least one door after the accident in order to facilitate quick rescue of injured occupants. Rescue teams with short access time and efficient emergency treatment in trauma centres and in ambulances are other very important factors in the efforts to save lives after a crash.

With this background, how does a car manufacturer proceed to develop a safe car? To have this background is not enough. Efficient means of transferring the knowledge into the design and development work and to convert it into safer car designs, are essential. When Volvo's work is described, reference is usually made to the diagrammatic sequence shown in Figure 2, which both starts and ends in the traffic environment in the real world.

To know where to concentrate in the development of a car, crash data is analysed at Volvo and also suggestions are obtained on what can be modified in the car. Although some legal safety standards have been established to try to cover the complex reality that exists out on the road, some car manufacturers have developed their own standards in addition to the legal requirements. The accident analyses are used as input to the development of these crash test methods.

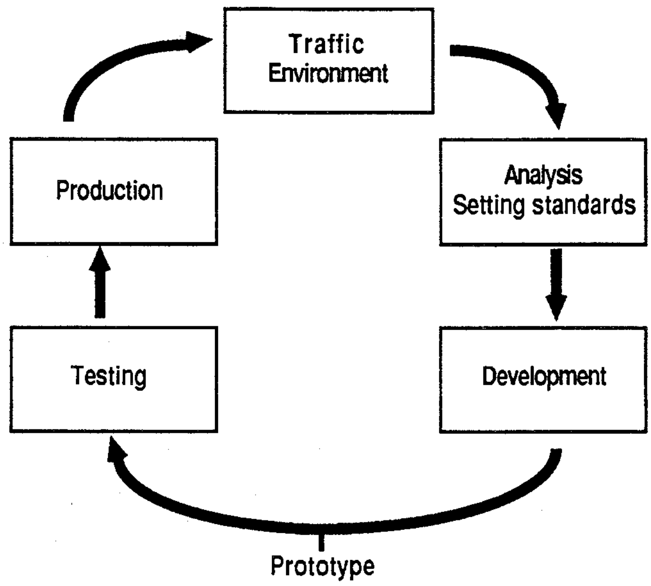


Figure 2: Volvo's production cycle

To know if the car passes or fails the test, criteria will be needed and these are what are called standards in Figure 2. When a new design is developed at Volvo the standards are known to the design engineers. The new design is then subjected to tests to see whether it fulfils the standards that were set up. It can do so the first time, but some of the standards may fail at the first test of the prototype. If so, the crash test engineers will find out why this happened and then suggest alterations in the design. The development department will be informed on what could be changed; a new design will then be developed and tested and maybe several loops will have to be run here before all standards are fulfilled. Processes similar to this one can be found for other properties of the car like noise and vibration or climate comfort. Once all of this is fulfilled, the car is ready to go into production but the crash activities won't stop here. Every now and again a car will be taken directly off the production line and into the crash lab where it will be subjected to tests to check that the safety features of the car were maintained when the prototype was put into mass production. This is because prototypes are usually made singly and with great care, maybe by hand.

The car then goes into the traffic environment and hopefully an accident won't happen, but if it does the accident research teams can investigate whether the safety design that was implemented has performed as intended. Since so much of the work that Volvo does is based on field data, the company's Accident Research Team is important. It was established more than 20 years ago and since then it has been investigating all major Volvo accidents in Sweden. Its purpose is to collect information about what happens to the car and the occupants in real world traffic accidents and this includes both the mechanism behind the accident happening and the outcome of the accident. Another goal of the Research Team is to feed this knowledge back into the design and development work.

The Research Team's work is carried out in three different ways. In-depth studies of accidents are made, statistical data is collected and some special studies are also performed. The in-depth studies are performed in two different geographical zones, local and distant. The offices are in Göteborg on the west coast of Sweden, and the local area is what can be reached in one hour's travel from the office in Göteborg. If someone in a Volvo car has sustained injuries in a crash, the Accident Research Team will be called directly to the scene by the police or by the Alarm Control Centre. The Accident Research Team is on call 24 hours a day, 7 days a week.

At the accident scenes, photos will be taken and if the police are there team members will talk to them and also to the people that were involved in the crash if they are still there. The car will of course be investigated also and a number of measurements will be taken both on the exterior and in the interior of the car. After more than 20 years the engineers know rather well what parameters are important to record in order to be able, maybe in 2 or 3 years from the day that the accident happened, to recreate the impression of the accident. These in-depth studies give a very good and detailed picture of numerous items. Among them are the collision objects, the traffic environment, the road conditions and how the car behaves.

The distant area is primarily the rest of Scandinavia and also if there are accidents of special interest in other European countries, we can travel there too. But obviously in this area it is not possible to reach the accident site while the car is still there. However useful information can still be obtained by visiting the garage where the car has been stored or the workshop where it is repaired and also by looking at the accident scene. In total, more than 100 in-depth studies are carried out in the local and distant areas each year.

The aim of Volvo's special investigations is to try to increase knowledge of vehicle safety aspects in ways that are not covered by the usual data sources. So here, for example, external accident data are analysed and special services are performed. One special investigation has been the neck study that will be described later (p. 71).

Statistical data collection is not as detailed as the in-depth studies but it covers a much larger number of accidents. In Sweden, Volvo has its own insurance company, Volvia, which insures about 500,000 vehicles. Each year about 45,000 of these are involved in some type of accident - it could be anything from scratching the bumper to a fatal accident, but as long as it involves the insurance company it is among the 45,000 accidents. The most severe cases are selected on a repair cost criterion, so that about 2,000 accidents are analysed each year. All of the accidents, including those in the in-depth study, will be followed up by questionnaires. A questionnaire is sent to the owner of the vehicle and this asks for a number of details about the persons who were travelling in the car — their height, their weight and so on — as well as a description by the people involved, why the accident happened, and about the road conditions. One very important question asks for permission to obtain the medical records of injured persons. The vast majority of people agree to this and by doing so they allow access to an excellent data source, because the medical records are usually very detailed in describing location and type of injury. A medical doctor associated with the Accident Research Team takes care of this data. Also police reports can be used in the follow up of an accident.

All of the information obtained is coded and stored in a computer data base. About 200 variables for each accident are recorded. The vehicle damage is coded by the collision deformation code while personal injuries are coded according to the AIS system. It is to be hoped that the disability scale mentioned by Koshiro Ono will be reality soon so that it also can be included in the data base. At present the data base covers over 15,000 severe Volvo crashes with more than 25,000 occupants.

To illustrate some of the statistics that can be extracted from the data base, Table 2 shows the distribution of crash types. It can be seen that a third of serious or fatal crashes are due to frontal impacts, a quarter are due to side impacts and 12% are made up of rollovers. The rest is the combination of other crashes, called multiple crashes. There is a particular category called 'other' and although there are no kangaroos in Sweden, there certainly are moose who like to walk on the street, and they make up a substantial part of this 15%.

Table 2 : Crash types in Volvo's data base

Crash type	Percentage for all crashes	Percentage for serious to fatal crashes
Frontal impact	35	33
Side impact	20	25
Multiple crash	17	14
Rollover	12	12
Rear end impact	7	1
Other	9	15

(Based on 14,700 crashes)

Luckily, most of the injuries that occur in the accidents are of minor threat-to-life risk: about 70% are AIS 1 and 9% are severe to fatal injuries (Table 3).

Table 3 : Distribution of injuries by AIS

AIS level	Percentage
1	70
2	14
3	7
4-6	9

(Based on 26,000 injuries AIS 1-6, sustained by 10,000 occupants)

This accident data can be combined in many ways but only one will be shown here. The distribution of injuries in the different body regions in Figure 3 shows that when all crash types are considered the head is the most frequently injured body part, followed by the chest. In the case of side impacts, that order is reversed: the chest is more frequently injured than the head. Also in side impacts, the abdomen and the pelvic region are more frequently injured than in crashes in general. Findings like these are very important when new test methods are developed to concentrate efforts on the relevant areas of interest.

A variety of barrier tests are carried out at Volvo: the full frontal barrier test; the partial overlap test, where only part of the front will be engaged; rear-end impacts, which can be used, for example, for evaluation of fuel leakage; and the side impact.

While the full-scale crashes give very good information on the vehicle performance, they have the drawback that it 'consumes' one car each time. So there are other ways to evaluate some systems like seat belts, and Figure 4 shows a sled crash simulator on which have been mounted the relevant parts that are needed to get an impression of the behaviour of the seat belt. Mounted are a seat, a steering wheel and the seat belt itself. Also sometimes it is sufficient to just test a subsystem as by mounting the side members and other structurally important parts of the front of a car on a dynamometric wall and impacting with a mobile barrier.

Another excellent tool is to take advantage of the sophisticated computer technology of today. There are a number of models available in both 2 and 3 dimensions, for both the occupants and the structure of the car, but this is a science in itself and will not be covered in this presentation.

In crash barrier tests, dummies (Fig. 5) are used and by measuring the effect on the dummy it is possible to estimate the risk of occupant injury in a real world crash of corresponding severity. The desirability of having a means to correlate the measurements made in the laboratory with the injury risk out on the road has already been mentioned and Volvo has, in fact, developed

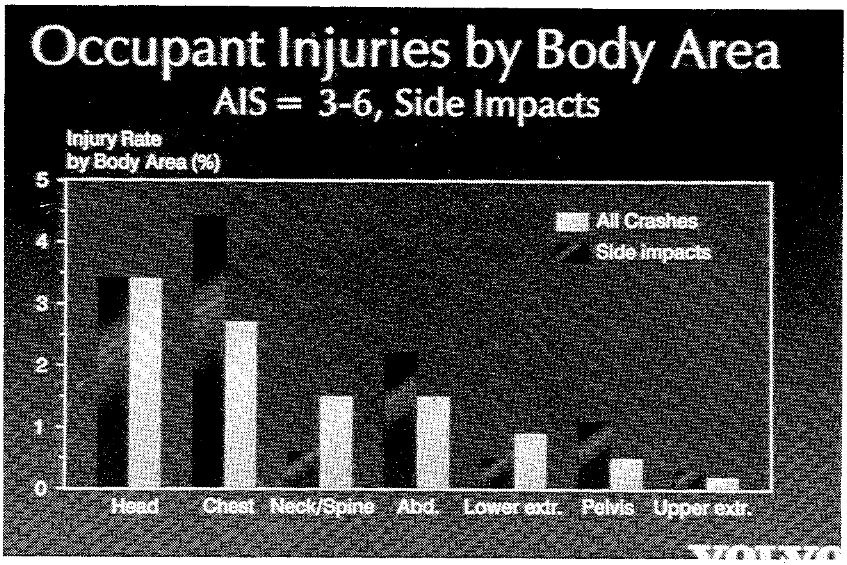


Figure 3: Occupant injuries

Figure 4: Sled crash simulator

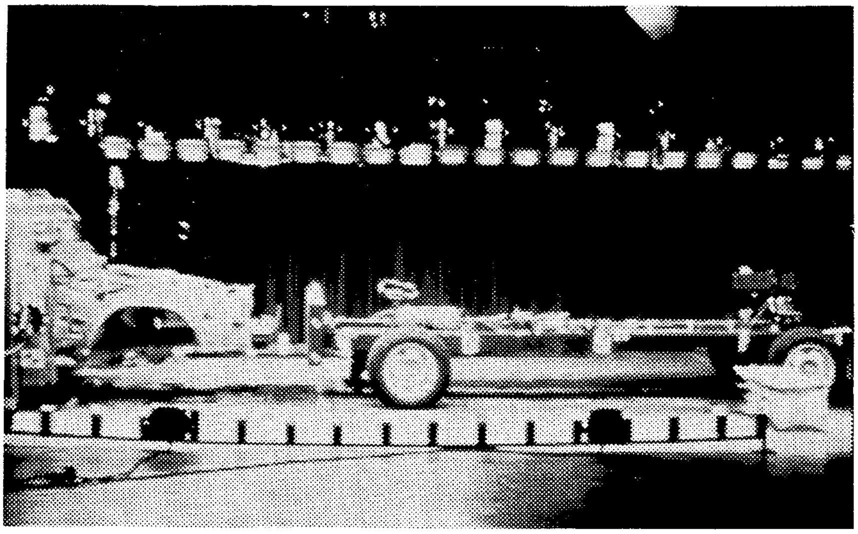


Figure 5: Volvo crash dummies

such a method recently. It is a statistical method which predicts injury reducing effectiveness of a new safety design long before it is out running on the roads. This statistical method depends on the use of accident statistics to describe injury risk as a function of crash severity for a well defined crash configuration. Crash severity could, for example, be expressed in kilometres per hour. For a specific car model, it is possible to fit a curve to accident data as shown in Figure 6; this curve will then describe the specific injury risk for each level of crash severity for that specific car model. If dummies are then placed in this particular car and the crash is reconstructed at different speeds, the dummy response at a number of different crash severities will be obtained. Thus there will be a common link between the dummy response and the injury risk (Fig. 7). Then by appropriate calculation, for a particular crash severity the dummy response and an injury risk will be apparent.

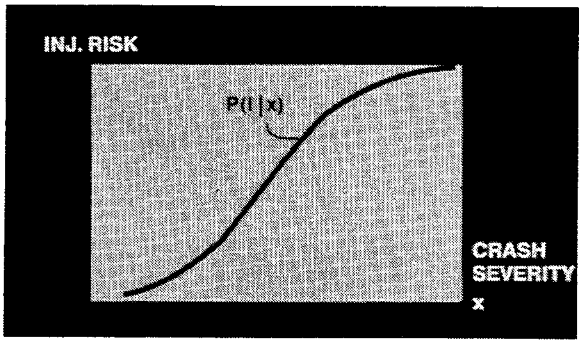


Figure 6: Relationship between injury risk and crash severity

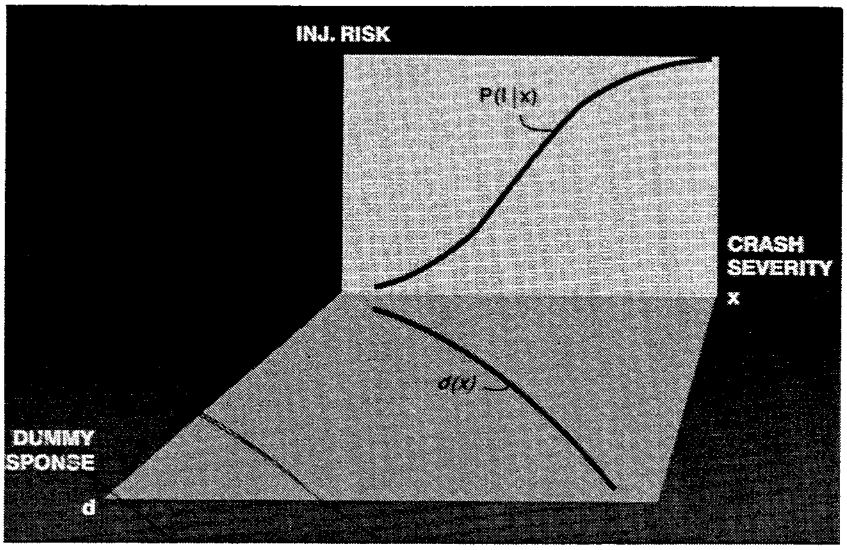


Figure 7: Dummy response linked to injury risk

This data can be combined to obtain a new graph which gives the relation between dummy response and injury risk (Fig. 8). Once this relationship is established, a new design can be tested at different speeds, to give a curve which shows dummy response as a function of crash severity for that design (Fig. 9). By moving backwards compared with the previous case, because the relationship is still the same, a dummy response can be picked and the corresponding injury risk will be seen. Eventually, a relationship for the new design of injury risk as a function of crash severity will be determined. Graphs for the old design and the new design are illustrated in Figure 10.

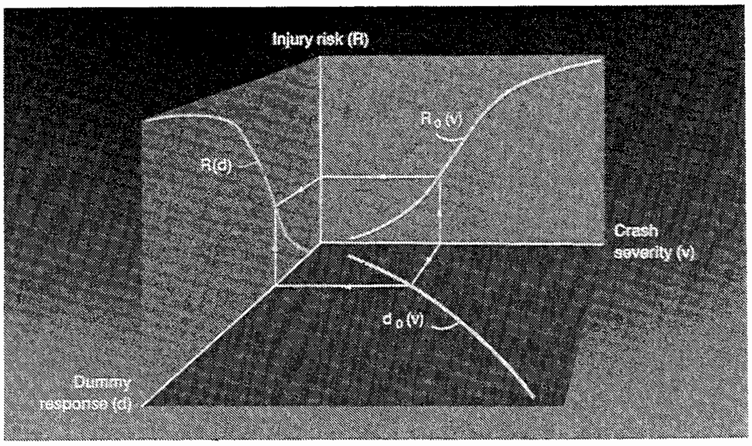


Figure 8: Relationship between dummy response and injury risk

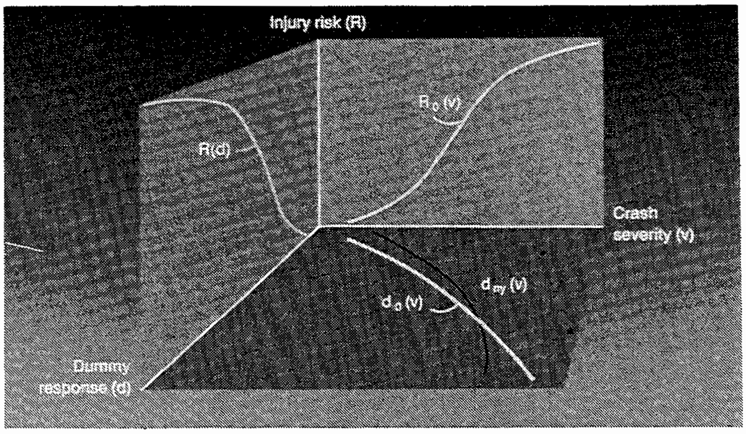


Figure 9: Dummy response as a function of crash severity

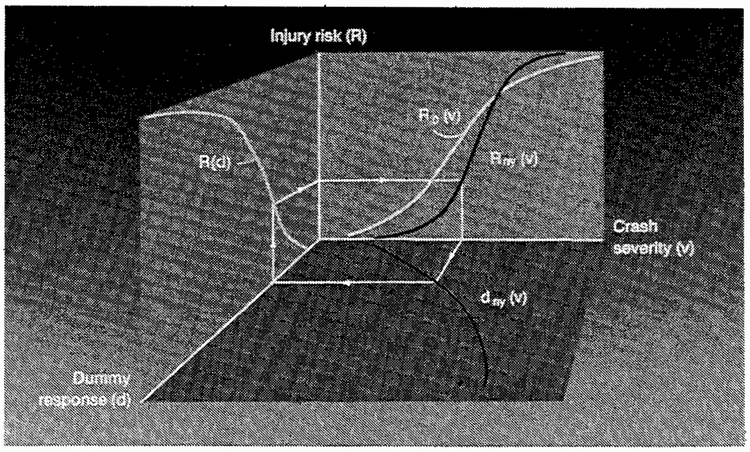


Figure 10: Injury risk as a function of crash severity

This has particular use because the distribution of crash severities is known from the accident data and if it is assumed that this would remain unchanged in the near future, it can be used as a weighting factor when assessing the amount of injury over the whole range of crash severity. So if this data is combined with the curves derived as described above, then the amount of injury over the whole range of crash severity will be the area under the graph corresponding to the design under consideration (Fig. 11). A reduction of injury will then correspond to the smaller area. If the difference between the base line model and the new design shows a reduction of the area by x%, then the injury risk will have been reduced by this percentage.

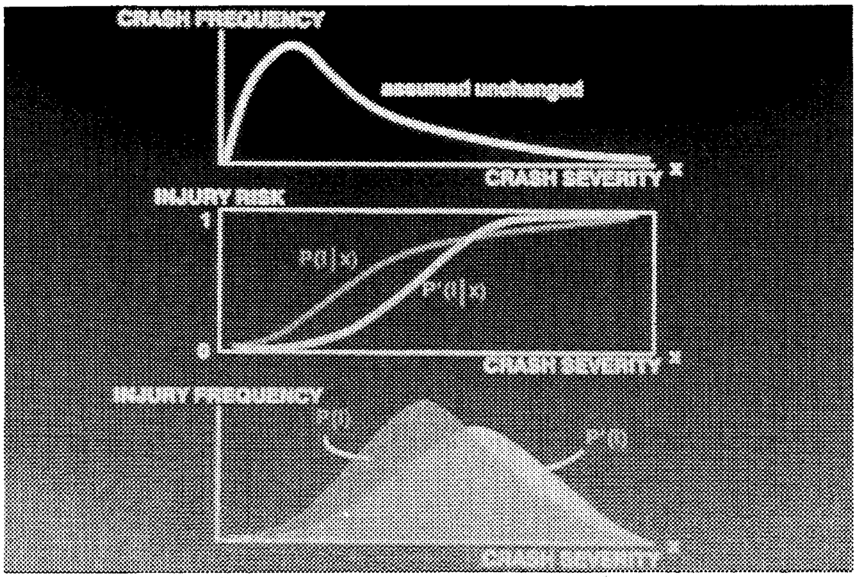


Figure 11: The relationships between crash severity and crash frequency, injury risk and injury frequency, respectively

Some hardware resulting from this research is the side impact protection system which was developed by Volvo using this methodology to set the standards for that impact protection system. For a person sitting in a car, there is a very limited amount of space between the person and the impacting vehicle in the event of a side impact. As this concerns car/car impacts, there is also very little area to put energy absorbing material in, as compared to the frontal impact in which case there is the whole car front to work with. What was decided with this design was to limit the degree of intrusion into the vehicle and Figure 12 is a schematic sketch of what Volvo wanted to achieve. It was decided to reduce the intrusion of the side wall and to keep the speed of this intrusion at a low level. By limiting the intrusion, a good environment for a padding to work in would be created, as padding would be less beneficial if the side wall collapsed. Therefore the lateral strength of the car was increased by putting substantial cross members under the rear seat and also under the front seat, by re-inforcing the B pillar, upgrading the roof rail and strengthening the roof cross member. Lateral tubes were placed under the seat or in the seat, so that when a car impacted they would transfer the energy onto the energy absorbing box mounted on the tunnel between the two front seats. The door panel is a very flat panel with no protruding arm rests and underneath this panel, padding areas have been mounted where the chest and the pelvis would be likely to make contact.

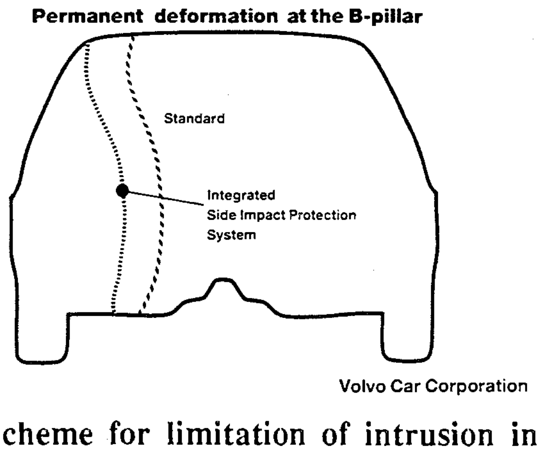


Figure 12: Scheme for limitation of intrusion in side impact

The energy flow that would be seen in the event of a crash is shown in Figure 13. It is clear that the crash energy can be transferred through all these elements and this will reduce the intrusion of the wall.

Referring to the middle graph shown in Figure 11, it was found, based on results of the tests that were performed, that both for the pelvis and for the severe to fatal chest injuries, the new vehicle design had reduced the injury risk by 25% in car/car impacts and this was considered to be a rather good reduction. This reduction might have been greater if very large blocks of padding could have been mounted on to the door, but then there would be a conflict with compartment roominess. There are not many people who would like to sit very tightly against a padding or a padded wall. Another means would be to increase the width of the car but this would be in conflict with fuel consumption and these days fuel consumption is very important.

So what would be ideal would be a padding that would not be seen or sensed when not in use, but which would expand in the event of a crash — in other words, a concept similar to a side airbag. A research prototype of a side airbag has been developed. Volvo designed it for the chest and presented it at the Experimental Safety Vehicle Conference in 1989. There are many manufacturers working on the side airbag and hopefully it will be in production before too much longer.

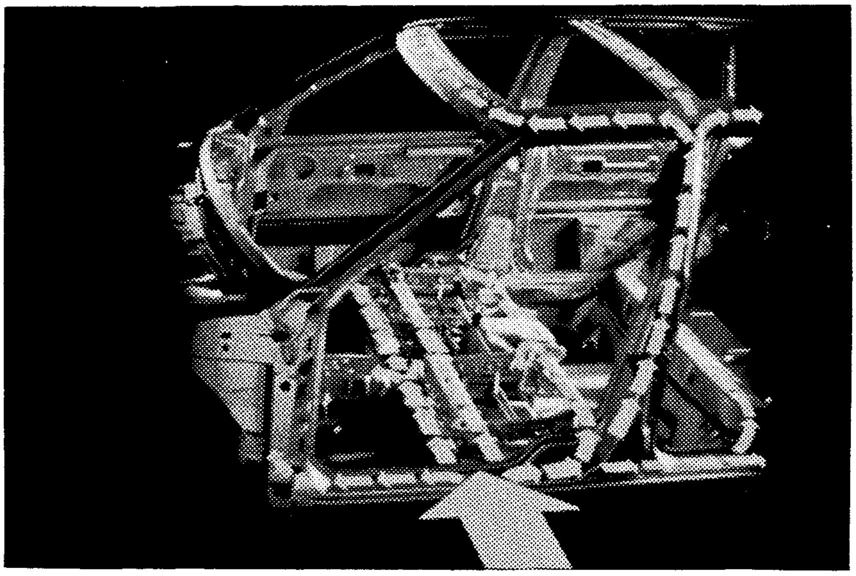


Figure 13: Energy flow in side impacted vehicle

QUESTIONS/COMMENTS:

Ron de Forest: Barrier testing is sometimes criticised for variations in results that are achieved and I was wondering, with the high number of tests that you've obviously done at Volvo, whether you could say how much of that variation could be attributed to the vehicle and how much could be attributed to just the method of barrier testing?

Ingrid Planath: There is a substantial variation in crash testing. Whether it is the vehicle itself is difficult to distinguish because the barrier method itself, I guess, is one of the more repeatable ways of crashing, but the dummy contributes to the variation between different vehicles and I can't distinguish that. There may be people who can do that, but I cannot.

Brian Fildes: I'm particularly interested in the claim of 25% injury reduction for the side impact protection system. I wonder if that applied to all regions, or which particular body regions it applies to, and whether or not you've actually published those results?

Ingrid Planath: We have not published the background to this, but we have done tests that indicate this. The reduction is claimed for the pelvic area and for severe to fatal chest injuries.

Tony Ryan: In relation to that I'd like to ask — referring to the bottom graph in Figure 11— you have an overlap of two distributions. Is it that there's a reduction in one part of the graph and an expansion in another?

Ingrid Planath: You mean that we've made things worse at one speed while improving at another?

Tony Ryan: Perhaps.

Ingrid Planath: This was a schematic sketch and we have not made things worse compared to the baseline model at any speed, but we wanted to achieve a better result compared to the baseline model. A higher degree of improvement at the lower speeds than at the higher speeds was desired because most accidents in fact take place at lower speeds. So we did not want to make it worse anywhere, and concentrated our efforts below the very high velocity range.

Tony Ryan: Could I pursue that a little further. You said you knew the impact velocity of the accidents. Is that because of your continuing accident investigations, that you know what is the crash environment that you're aiming for?

Ingrid Planath: Yes, we have reconstructed several crashes with the purpose of being able to estimate the crash severity. With rear-end impacts we have developed a method so that we can estimate at what relative speed the other vehicle was driven into the car we are looking at.

CRASH HELMET EFFECTIVENESS OVER 50 YEARS

John Lane

The effectiveness of crash helmets in reducing injury in motorcycle crashes is of current interest. This paper presents a review of studies which have estimated the amount of injury reduction due to helmets.

The change in injury patterns relating to helmet wearing can be measured under several circumstances.

First, when helmet wearing laws are introduced or repealed, a step function change in the wearing rate generally results. Under these circumstances, it is not difficult to calculate the risk reduction effect of the helmet on head injuries.

Secondly, the effectiveness of helmets can also be derived from studies of crash data, where estimates of the proportions of head injury among helmeted and unhelmeted riders can be determined.

The most commonly available data are from populations of injured bicyclists and motorcyclists presenting in hospitals as the result of a crash. However, in this situation it is important to be aware of the selection bias that is likely to be introduced, given that the helmet's effectiveness in preventing or reducing injury may decrease the likelihood of a helmeted crash-involved individual requiring hospitalization. The proportion of potential head injury prevented by helmet wearing, among those wearing helmets, can be estimated as follows:

$$\frac{\text{incidence of head injury} \mid \text{no helmet} - \text{incidence of head injury} \mid \text{helmet}}{\text{incidence of head injury} \mid \text{no helmet}}$$

The results of studies examining the effectiveness of helmets in reducing injury are summarized in Figure 1, which shows the per cent reduction in injury by year of the published report. It is important to note that these studies use varying definitions of outcome; they are noted in the Figure. Although most of the studies examined the effectiveness of motorcycle helmets, some of the more recent studies focused on bicycle helmets.

Because of the amount of scatter, the differing methodologies and the use of various outcome measures, it may not be justifiable to draw a regression line to indicate the effectiveness against time of introducing helmets. However, if this is done, it can be seen that over the last 40 years there has been about a 10% improvement, which is perhaps a little disappointing. In fact, most of the protective effect of a helmet may have already been achieved by the relatively primitive helmets which were used in the 1950's and early 1960's, despite the more recent improvements in materials and technology.

The question then arises of why a greater degree of improvement has not been evident. This may be partly explained by the fact that substantial areas of vulnerable cranium are not protected by helmets and that the unprotected area has not been greatly reduced over the years.

It may also be that helmet design is following a wrong track, as suggested by others. For example, some studies have been able to identify individuals who have sustained quite severe head injuries but whose helmets have appeared to be undamaged. In other words, the liner has either not been compressed or, if compressed, has recovered completely. The helmet designers and/or manufacturers are required to meet various standards which are generally similar throughout the world: there are requirements for penetration and for some form of drop test with the criterion being expressed either simply in terms of headform acceleration or acceleration beyond a specified minimum duration. The now well-recognised question is whether these requirements have directed helmet development along a somewhat unproductive path in that the foams used to absorb energy within the liner have been too stiff.

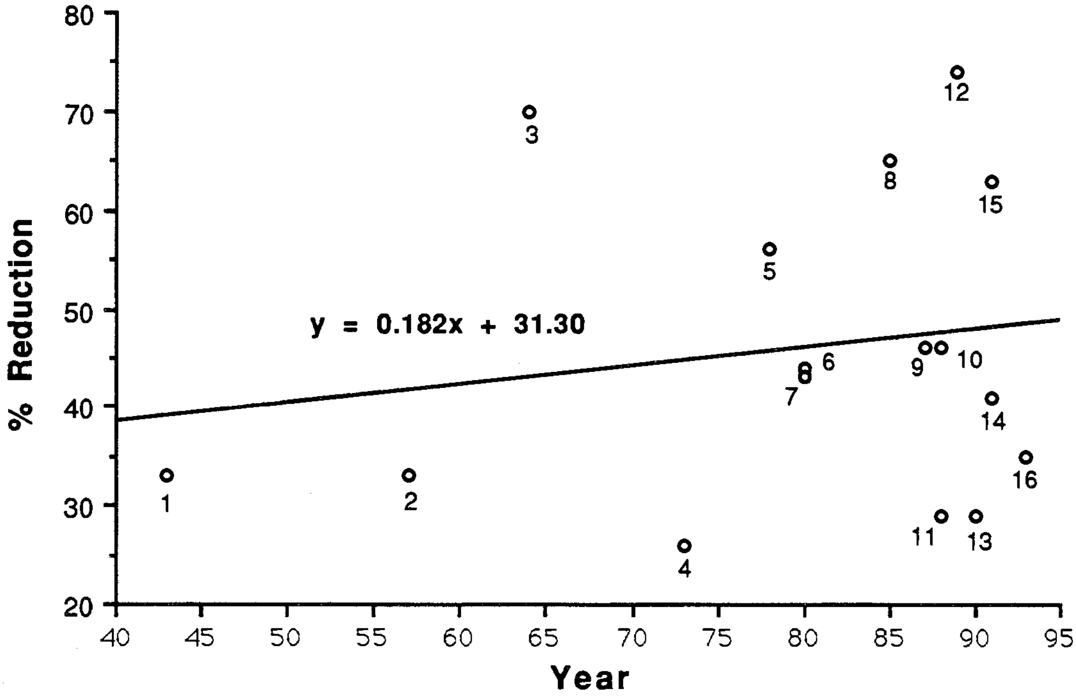


Figure 1: Per cent reduction in injury by year of publication

1. Cairns, H., Holbourn, H.: Br Med J 1:591-598, 1943
2. Chandler, K.N., Thompson, J.K.L.: Operat Res Quart 8:63-71, 1957
3. Foldvary, L.A., Lane, J.C.: Aust Road Res 2:7-24, 1964
4. Jamieson, K.G., Kelly, D.: Med J Aust 2:806-809, 1973
5. Graf, L., Blair, R.W., Weisbuch, J.B., et al : American Public Health Association, 106th Annual Meeting, 1978
6. McSwain, N.E., Lummis, M.: Surg Gynecol Obstet 151:215-224, 1980
7. Watson, G.S., Zador, P.L., Wilks, A.: Am J Public Health 70:579-585, 1980
8. McSwain, N.E., Petrucelli, E.: J Trauma 24:233-236,1984
9. Dorsch, M.W., Woodward, A.J., Somers, R.L.: Acc Anal Prev 19:183-190, 1987
10. Bachulis, B.L., Sangster, W., Gorrell, G.W., et al : Am J Surg 155:708-711, 1988
11. Evans, L., Frick, M.C.: Acc Anal Prev 6:447-458, 1988
12. Thompson, R.S., Rivara, F P., Thompson, D.C.: N Eng J Med 320:1361-1367, 1989
13. Wasserman, R.C., Buccini, R.V.: Am J Sports Med 18: 97-98, 1990
14. Offner, P.J., Rivara, F.P., Maier, R.V.: J Trauma 31:1035, 1991
15. Kelly, P., Sanson, T., Strange, G., Orsay, E.: Ann Emerg Med 20:852-856, 1991
16. McDermott, F.T., Lane, J.C., Brazenor, G.A., Debney, E.A.: J Trauma, In Press 1993

It has also been suggested that the use of the Head Injury Criterion (HIC) instead of a simple acceleration may also not result in much improvement. Given acceptance of the HIC algorithm, one is still left with the issue of the appropriate criterion level. This raises the question of what outcome is being sought: prevention of concussion, prevention of brain injury or prevention of death. Hope and Chinn at TRRL in England have been trying to calibrate HIC against outcome criteria. Table 1, presented at the 1990 IRCOBI Conference, indicates an 8.5% probability of death at the currently used criterion of 1,000 HIC. This suggests that the criterion level is too high.

In summary, the aim of this paper has been simply to raise the question of why helmets have shown less improvement than might have been expected and thus provoke some discussion or controversy over this issue.

Table 1: Probability of death by HIC

HIC	% fatality
4,000	65.0
2,000	31.0
1,000	8.5

Source: Hope and Chinn, 1990

QUESTIONS/COMMENTS:

Michael Henderson: I agree wholeheartedly with what you're leading towards, that we ought to be looking at softer foams for all the helmets, but I live close to the problems faced by the manufacturers in relation to the standards-setting bodies and the difficulties there are with the dummy or any other sort of modelling. I don't see how the circle which the manufacturers, the development people and the standards organisations are in can be broken and I would appreciate your views on this. It is for instance, the Snell Foundation's avowed intention to stay ahead of the other standards-setting organisations. The only way for the manufacturers to meet the Snell Foundation's standards is to build stiffer helmets. There is no choice. No one can even try making a more compliant helmet. How do we break out of that circle?

John Lane: Clearly, the problem really is the Standard. Well, it's not quite as bad as that. One of the constraints on the Standards has been the penetration test, which has mandated the fairly massive hard shell and this means that the shell initially behaves elastically and returns quite a bit of the velocity as rebound velocity — as much as 60% at times, I'm told — so this is clearly not a good result. There is plenty of evidence to show that the helmet very seldom strikes a small hard object and at least the penetration requirement has been removed now from the Australian Standard on bicyclist helmets. This is certainly a step in the right direction. Perhaps the next thing to do is to assess the performance of the newest helmets, which comply with the Australian Standard which became effective in about April of this year. In turn, the problem really does revolve around the better modelling of the head injury process and getting a better figure of merit than the scale that HIC is used against — the points that were raised earlier today. I'm sure that at least getting rid of the requirement for a very heavy hard shell is a step in the right direction.

Michael Griffiths: A problem with Standards and the motorcyclist helmet Standard, is that a test that might have initially been in there to review, say, the penetration aspects of the helmet, ends up doing something else. For instance, with the motorcycle helmet Standard, people would be loathe to omit the test where the helmet is hit with a very sharp pointed object, because it's also become a de facto test for things like the abrasion qualities of the helmet, the durability qualities of the helmet, etc. If people want to bring about changes in the Standards, then somebody has to make an investment, either manufacturers or government, in finding more appropriate tests for these performance attributes. So if you want to get rid of the penetration test, for instance, you have to ask why the test is not being improved and suggest an abrasion test or a durability test. Then we'll get rid of the old test because it's not really serving a purpose any longer. So an investment is required, in terms of looking at what the test is supposed to be doing and then looking at what set of tests are really needed.

Tony Ryan: I might add that in 1963 and 1964, when Jack McLean and I were looking at crashes, we had a collection of some thirty motorcycle collisions and my recollection is that the only difference between riders wearing helmets and not wearing helmets was that while the incidence of concussion was similar in both wearers and non wearers, those not wearing helmets had lacerations to their scalp. So the only visible protective effect was that the helmet stopped soft tissue injury to the scalp. Just recently, we've had some young engineers trying to calculate the stiffness of a helmet in terms so that we can estimate the acceleration of the head. The stiffness, to my amazement in view of some of the assumptions that we make, is roughly the same as that of the A pillar of a car. That is, the helmet plus the foam is, in effect, the equivalent to the effect of being struck by the A pillar of a car. Now, just intuitively to me, that doesn't seem right in a protective piece of equipment, so maybe there's some work to be done on that.

NCAP: AN AUSTRALIAN NEW CAR ASSESSMENT PROGRAM

Michael Griffiths

This talk is mainly an update of the current state of Crashlab's NCAP program with particular reference to the development of the barrier testing. It is a privilege to be the person making this presentation on behalf of all my colleagues who have conducted this program.

The Australian NCAP program aims are:

- to provide information of the relative performance of cars in frontal impacts in controlled conditions;
- to focus consumer interest on the crash safety performance of cars; and
- to provide consumer incentives for manufacturers to have relatively good crash performance of their vehicles.

The methods that are proposed in the Australian NCAP program are:

- a full frontal impact test at 56 km/h;
- an offset frontal impact test at 56 km/h;
- use of Hybrid III test dummies; and
- measurement of as many injury criteria as possible with the equipment available; these include:
  - Head Injury Criteria
  - Neck Injury Indicators
  - Face Injury Indicators
  - Thoracic Injury Criteria
  - Spinal Injury Indicators
  - Abdominal Injury Indicators
  - making seat belt geometry assessments with the Canadian belt-test device.

The vital tool that was needed to do this was a crash barrier. Crashlab's new barrier consists of a very large concrete block supported on piers set down in the ground, each pier being more than 1.5 m in diameter and more than 8 m deep. The barrier can withstand a 5 tonne vehicle at 100 km/h with little deflection. The specification was for an impact from a 5 tonne vehicle at 80 km/h but the testing that has been done so far indicates that it performs well in excess of that. There is a deflection of 0.25mm from a 2 tonne vehicle at a 100 km/h and from a 5 tonne vehicle at 80 km/h.

The capacity of the drive motor allows it to accelerate a 5 tonne vehicle up to 100 km/h with quite low g's (< 0.3 g). The sophisticated program allows an S-shaped acceleration profile. It's not just a straight ramping effect: there is a gentle initial acceleration phase, a fairly straight acceleration phase and then a tapering off phase before the stabilisation period is entered. The capabilities of the system are that it must maintain program speed to within 5% during the acceleration phase and it must maintain program speed to 0.5% during the stabilisation phase and the final crash phase.

The track length is 145 m. The barrier is undercover and there is an airconditioned moveable preparation chamber which is used to set up the dummies, the vehicle and any other equipment. At the moment the track is open, but if work of a commercially confidential nature is secured, enclosure of the entire track is probable.

The decision to build a crash barrier in New South Wales was made primarily because we wanted to conduct an NCAP program. This meant we needed access to an independent facility in Australia. That needed to be a facility with potential for more flexible use to allow testing of roadside furniture and other aspects of roadside equipment. Also, it was recognised that a local Australian facility would be economically beneficial in removing the necessity for overseas testing: for example, the high cost of the Capri program to the Ford Motor Company was an indication that a local facility might make a small but significant dent in the Australian balance of trade figures.

A video showing a series of commissioning tests on the crash barrier followed, with a range of car impacts from 48 to 106 km/h and a 5 tonne truck at 80 km/h. Also, the car to car capability of the track was illustrated by the inclusion of an outside test, with 2 cars each at 60 km/h giving a closing velocity of 120 km/h.

These tests were done for the contractor to show the capabilities of the barrier. Tests using that range of speeds are not planned in the near future. The video shown gives some idea of the capabilities of the barrier: they are ultimate capabilities, and the main tests envisaged will be at 56 km/h.

The need for some sort of NCAP program was realised more or less simultaneously in various States of Australia and in some of the motoring organisations, so that the New South Wales initiative attracted several expressions of interest. As a result it has now become an AUSTROAD program and is a national program with funding from the State Authorities and motoring organisations. That was very satisfactory, as a national identity was wanted from the start. Also, once the barrier was built, it was realised that there was a need to provide separation of the tester from the NCAP program runner, to enhance the prospect of getting work from the automotive industry. It is important to have that separation, so that is another good reason to have the AUSTROAD involvement. We now have a national NCAP management committee. The convener of that committee is Mr Ian Pettigrew from the Queensland Department of Transport.

The crash barrier will provide local development opportunities for vehicle manufacturers in Australia. The facility allows an independent assessment of the crash test performance of vehicles. The vehicle manufacturers can request, at their cost, the placement of extra transducers on the vehicles during NCAP testing and they will have access to items such as the Canadian belt-test device. Also, there are deformable load-sensing faces which can be put on the Hybrid III dummies. Thus manufacturers will have the chance to get information on their vehicles' performance earlier rather than later. Also Crashlab hopes to have the involvement of the staff of manufacturers and government authorities to allow the facility to perform an additional role in local staff development.

Along with the NCAP program there is potential for other research opportunities which can be conducted as expansions of the NCAP program. This is not part of the program to be used for publication but parts of the program that can be used just as a straight research opportunity. The committee running the NCAP program is looking for and evaluating other opportunities from the program. After vehicles have been crashed, some early work into side impact testing could be done, using the crashed vehicles to do research assessments of side impacts to either the European proposed Standards or the USA Standards. With vehicles that are going to be frontally tested, there is the potential to do low speed rear-impact testing. Also the Ambulance or Fire Brigade or other experienced rescue teams could be involved in doing assessments of the comparative accessibility of vehicles after they have been crash tested. The motoring car clubs could organise repair cost assessments. Structural experts could do manufacturing consistency assessments. Another use of the tests could be for simulation validity assessments: there are a number of joint projects in Australia at the moment, looking at super computer crash simulation.

Those involved in the program are also concerned with what is happening internationally. There has been constant communication with overseas bodies that are interested in doing NCAP-style testing and also with people and manufacturers doing tests, to gain opinions on what sort of testing should be going on, particularly with respect to offset tests because of the current international debate about the actual parameters that might be used. So an international review process of crash test procedures has started, and more may be mentioned about that during this Conference.

As well as review of the injury criteria, there are several factors that need resolution with respect to the frontal offset tests, such as the velocity, the percentage of the offset, any angle on the offset part of the barrier hit, the actual radius of the corner, and the desirability of some kind of addition to the facing material to stop the vehicle sliding off. With respect to NCAP, the camera positions for the United States' NCAP have been set for some time and it is proposed to review those and try to have camera positions that might provide more information. It may be that some of the camera positions which were established initially are no longer necessary due to the improved performance of some of the test dummies. Maybe there is scope for putting a couple of cameras on-board which might give a better picture than some of the views attempted from off-board cameras currently.

The other issue being raised and discussed now is what to aim for in the next generation of NCAP tests. It is known that the United States is reviewing its NCAP program. They are currently looking at side impact NCAP and regulations for rollover. Other areas which may be worth looking at are rear impacts and eventually some sort of assessment of the pedestrian-protective qualities of vehicles is going to come about; maybe they will be subjects for future NCAPs.

QUESTIONS/COMMENTS:

John Lane: Could I ask a bit more about the Canadian Belt Test Device?

Michael Griffiths: Basically it is a replication of the actual shape of the frontal torso and the abdomen of the human being, which is bolted onto the H-point rig and fitted in the vehicle; then seat belts are put across the device and measurements of the seat belt route are scaled up. From that you can get an assessment of the actual geometry of the seat belt — both of the lap parts of the seat belt and the sash part of the seat belt. The Canadian approach is that in the abdominal/pelvic area there are not a lot of readily available injury criteria, but there are some things we know about what constitutes good seat belt geometry and what constitutes bad seat belt geometry. This is an attempt to get an objective measurement of that geometry.

Tony Ryan: How many tests do you plan to run over what period of time?

Michael Griffiths: The program currently allows for approximately 90 tests over a 2 1/2 year period.

Tony Ryan: One every 2 weeks?

Michael Griffiths: Yes, if it's done that way. More likely it will be done in intense bursts which could be of one per week, with current staff resources!

Bryan Knowles: With regard to the NCAP testing procedures that you're planning to do: in the States, we know that NCAP and the manufacturers work very closely together with the results, and we know that in the near future you're going to be crashing a number of cars and publishing the results. In the States when NHTSA have done their tests, if there are results which appear to be significantly at variance to what NHTSA might have expected or think might be embarrassing to the manufacturer, the manufacturer is asked to comment or inspect the data and go through and ensure that the data is real and within expectations of the manufacturer. If there's then nothing that appears to be remiss, then NHTSA go ahead and publish that data. Given that the local manufacturers generally with their cars aren't even testing to (FMVSS) 208 because it's not yet an Australian Design Rule rule and it's something that, within our environment, it's difficult to justify on a cost basis: when you've got your results and you've got your numbers, what recourse does the local manufacturer have to explain the results and not suffer really some PR damage in relation to, perhaps, some of the imported cars that might be exported to the States as well, probably do meet the rules, but could cause quite a bit of damage to the local car industry which is at a disadvantage in this.

Michael Griffiths: The checks on the testing here will be very similar to the checks that existed in the USA, in that manufacturers will be given the opportunity to observe the test, they will have immediate access to their own test results and will have a period of time in which they can comment on the test results. So they'll have the chance at that time to say whether they can explain a test result or think it is not right or whatever else. So that will be the opportunity they'll get.

To date, in terms of working closely with the industry, we have had 2 sessions with the Federal Chamber of Automotive Industries. At the last session, a few weeks ago, there were 26 representatives of the automobile industry and future meetings are planned.

Bryan Knowles: I'm sorry, but I think you're avoiding the point. The point is that the local manufacturers by and large don't have a large body of crash information on their unique Australian cars to know whether the result is right, wrong or indifferent, and furthermore aren't really in a position to do much about it, whereas an importer who sells a car to the States does have that body of information because of the economy of scale he's got and the need to do it for that particular market. Really, the point I'm trying to make is that because those indigenous vehicles are not being developed to any form of 208 Standard, at least not on sale right now developed to a 208 Standard, we are at some disadvantage.

Michael Griffiths: Well, it may be so. If a manufacturer hasn't done any testing with their vehicles at those particular speeds, they may indeed be at a disadvantage. We're getting conflicting messages in that regard from manufacturers as to whether they have access to testing of their vehicles.

Michael Henderson: In the United States, if a manufacturer has a problem with a test and its results, what then happens? Is it not the case that they are published anyway?

Michael Griffiths: Yes, if a manufacturer can't produce information that says there is a problem with the test, then the test results would probably be published anyway. The manufacturer has the option of asking for a re-test, as his cost, which is the same provision as they have in the United States' NCAP system.

Bryan Knowles: Do you mean the second results would be published along with the original ones?

Michael Griffiths: We haven't finalised exactly what is going to happen in that regard, but I would think that if a manufacturer came in and said that they wanted to have a number of tests done at their own expense — well, I shouldn't pre-empt the NCAP committee's decision, but I think that would be taken into account.

FACIAL INJURY BIOMECHANICS

Ingrid Planath

This talk is to be an overview of research done in the areas of facial injuries. Some accident data will be presented as well as what has been done in the biomechanics field and also how biomechanics research has been used in producing assessment techniques for facial injury.

Volvo 1988 accident data shows how common facial injuries are today. Of nearly 6,000 belted drivers in frontal collisions, 44% were injured and of those, 10% sustained some facial injury.

With regard to the severity of these injuries, the data in Table 1 shows that, as was the case with accidents in general, most were of low severity degree: 77% were AIS 1, AIS 2 made up 17% and only 1% were severe or fatal facial injuries. It must be pointed out that this is AIS, which measures threat to life, and this might not be the most appropriate rating of the harm that is sustained by the individual. There might be a lot of psychological harm in having a facial injury, even if it is rated AIS 1.

Table 1: Severity of facial injuries of belted drivers in frontal collisions

AIS level	Percentage
1	77
2	17
3	5
4-6	1

N=268

(Volvo 1988 accident data)

It is also interesting to see which types of injury were seen in this sample (Table 2). Lacerations make up a large part of the injuries. There has been research performed on lacerations, particularly in connection with windshield development. There were difficulties in quantifying these injuries but laceration severity indexes have been developed, for example, by Packard and his colleagues in 1973. Fractures, which usually account for the higher AIS values of facial injuries, can be found in approximately 20% of the cases.

Table 2: Types of facial injuries of belted drivers in frontal collisions

Injury	Percentage
Laceration	44
Fracture	21
Abrasion	13
Contusion	9

If the locations of all facial injuries are considered, the most commonly injured parts are the frontal bone, the nose and the mandible (Fig. 1). Injuries to the left or right of the face have not been distinguished in Figure 1, but are shown as one position. Lacerations are seen to follow a uniform pattern whereas with fractures, the maxilla is also frequently injured.

Thus much information regarding facial injury in the real world can be found, but what is needed in order to develop cars that have a lower potential of facial fracture, is a relevant test tool. Historically, much of dummy development has been devoted to frontal impacts and the development has concentrated on avoiding severe injuries which were found mainly in the head and chest in frontal impacts. So the dummies that have been developed so far are good for measuring brain injuries, but to enable studies of the effect of impacts on the facial region, a test tool that is based on biomechanical data is needed. Some of the biomechanical research that has been performed is reviewed below.

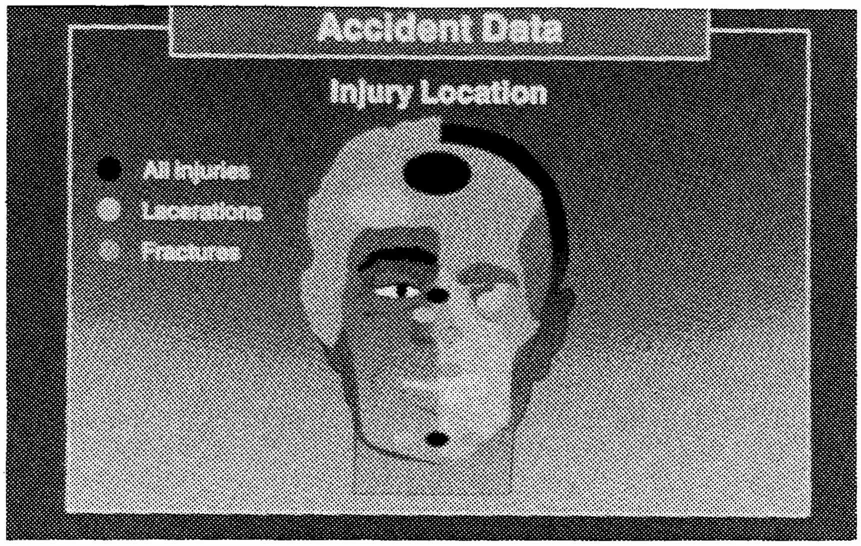


Figure 1: Location of facial injuries in frontal collisions

Of the work selected for discussion, the first is by M. Le Fort who in 1900 produced a classification system of facial fractures that is still in use today. Then during the 1960's in particular, studies on localised loading on individual facial bones were performed and in the 1980's distributed load was studied by a number of researchers.

Le Fort's main contribution was the classification of severe types of facial fractures. He used somewhat crude methods, such as table edges, to produce these fractures. Nevertheless he found interesting information and he classified facial fractures in accordance with the scheme shown in Figure 2. The description of a facial fracture as a Le Fort 1, for example, can still be found today in medical records and this would be a horizontal fracture across the maxilla.

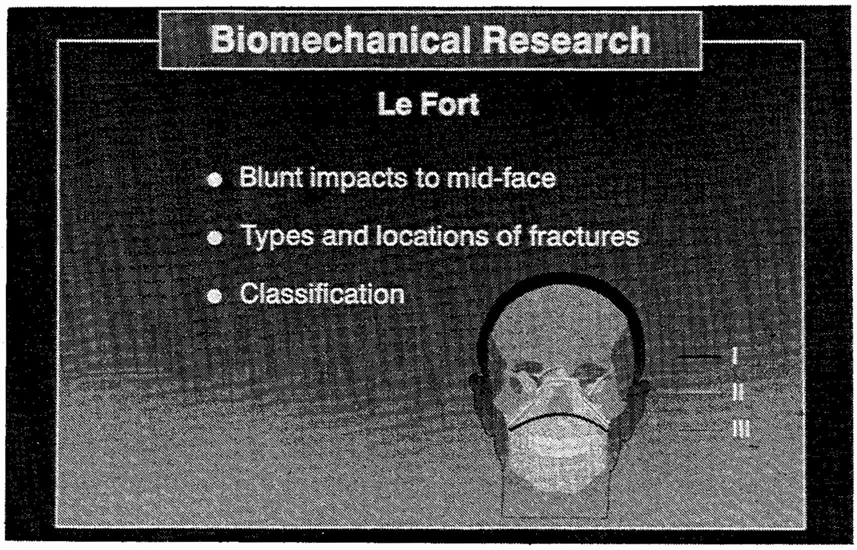


Figure 2: Le Fort classification of facial fractures

Various researchers have studied the individual facial bones - for example Hodgson, who applied localised impact to the zygoma. Different loading conditions were used. Swearingen found that the force required to fracture a facial bone depended on the loaded area. This result was supported by Hodgson who found, too, that the line of action of the force was of importance. Nahum and his colleagues also performed localised loads on a number of different facial bones and they looked at the maximum forces and also proposed some thresholds for fracture, or tolerances.

In the early 1980's, Dr. Tarrière at Association Peugeot Renault in France, expressed concerns that most of the biomechanical data for the face so far had been obtained by applying localised loads, but distributed loads are mainly dealt with in the automotive environment. This research team performed a number of cadaver and dummy tests in parallel and measured forces and accelerations. The outcome of this research was that the standard Hybrid II head, the dummy head that was tested, was found to be stiffer than the human face when subjected to a distributed load (Fig. 3).

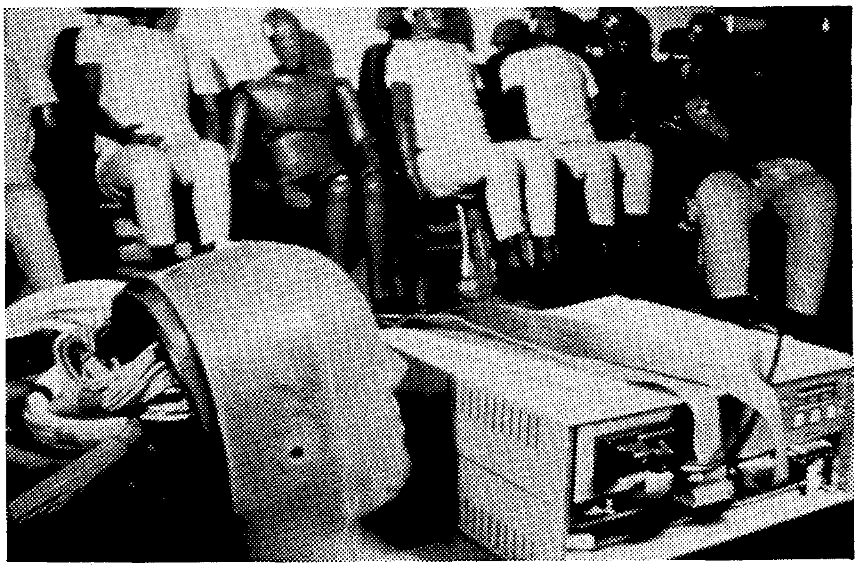


Figure 3: Hybrid II head test apparatus

Nyquist, an American, contributed to the biomechanical knowledge by a publication in 1986 in which were presented the results of impacts with different masses and different velocities onto the nasal bone. Force-deflection curves for this impacted area were one other result from this test. Other findings were that kinetic energy can be used to determine fracture. Some thresholds for force were also suggested.

Allsop and his colleagues also studied the more distributed impact. They developed an impactor that was able to measure force in 8 adjacent load cells (Fig. 4). They also performed tests in the frontal bone area, the zygomatic area and to the maxilla. This was done with cadavers but only one location was used in a test. Acoustic emission traces were also recorded for all impacts and there was a potentiometer attached to the impactor to provide deflection information. In parallel with the cadaver tests, tests on Hybrid III heads were performed in corresponding locations and under the same conditions. Figure 5 is one example of the types of traces that Allsop got from the research. The curve is the force/time history as recorded by the impactor and the 'cloud' is the acoustic emission. They can be correlated in time and what is seen here is that when the acoustic signal starts to oscillate up and down, it indicates a fracture. By combining the two results it is possible to determine a fracture force. Apart from force deflection data, this research also gave information on fracture forces. Figure 6 shows another sample of Allsop's results. This is a comparison of the human and the Hybrid III data. Comparison of the average plot for the different cadaver results with the result for the Hybrid III shows, as Tarrière had already proposed, that the dummy head was much stiffer than the cadaver average. This data was something that Volvo used later in the development of a new dummy face.

Some other attempts to assess facial injuries include the Hybrid II load sensing face which was developed in 1986 by Volvo together with Collision Safety Engineering, an American company (Fig. 7). Onto a standard Hybrid II head 52 piezo electric transducers were glued and each of them would provide the test engineer with pressure/time data. They were distributed over the whole facial area. The head is gold coloured in order to shield the transducers from noise. The

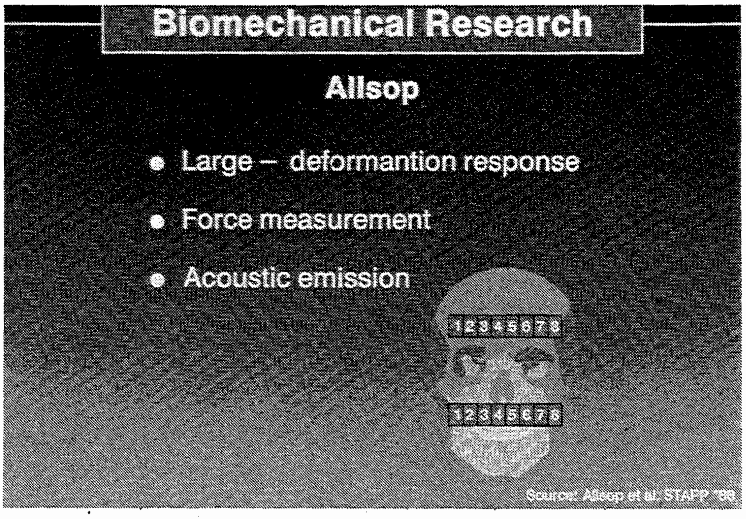


Figure 4: Position of load cells in measurement of distributed loads (Allsop et al, 1988)

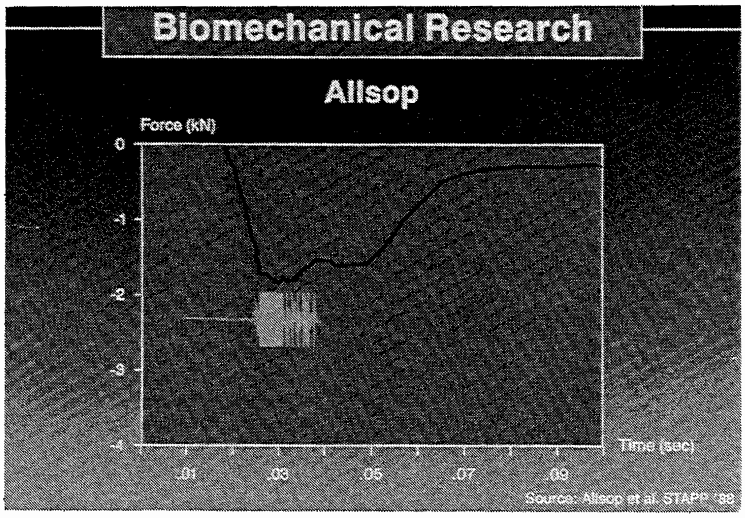


Figure 5: Example of force/time history (Allsop et al, 1988)

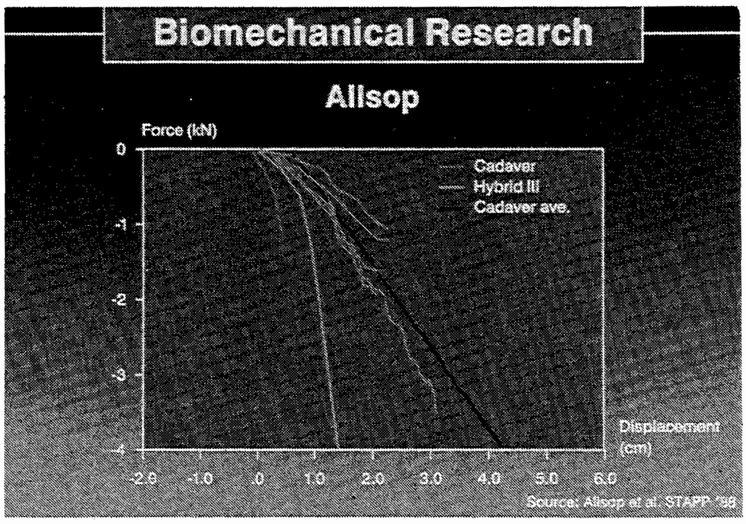


Figure 6: Comparison of results for cadavers and Hybrid III dummies in impactor tests (Allsop et al, 1988)

Hybrid III version (Fig. 8) followed one year later and a similar concept was used here. In this photograph, tape has been placed around the transducer locations so that their location is visible. However the number of sensors here has been reduced to 25 because it was found necessary to decrease the time that it took to evaluate the test. It was felt that this reduction could be made without the loss of any significant information. For both these dummy heads, rubber skin covered the cranium when it was used in a test. Although both heads give useful information about impact locations, they have a drawback in that they are not deformable in any sense.

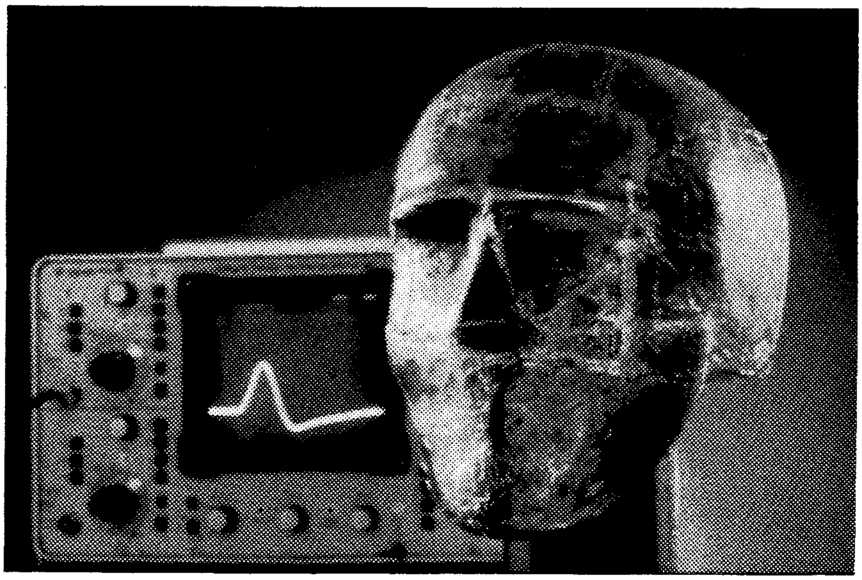


Figure 7: Hybrid II load sensing face

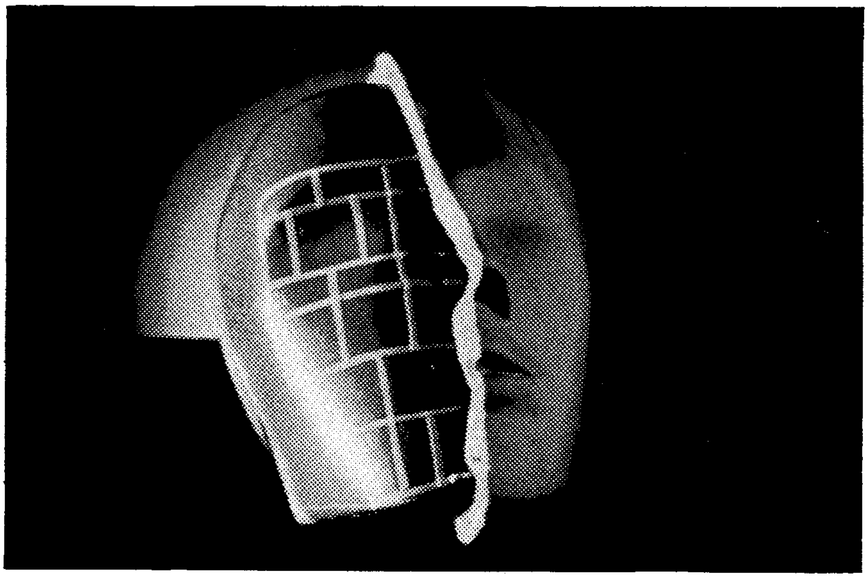


Figure 8: Hybrid III load sensing face

Deformable concepts have been looked at by a number of researchers. In the United Kingdom a flat honeycomb disc has been developed for use in evaluation of steering wheels and in a component drop test. The pass/fail criterion would depend on the depth of the dents in the honeycomb. It is, however, difficult to cut honeycomb in a repeatable way and also some difficulties occurred when it came to measuring the depths of the dents. Work is however still being done on this method.

Transport Canada developed a frangible insert that could be used on a dummy face and it would cover the facial area and have a fracture force that corresponded to a fracture force that had been found in biomechanical research. In France, Peugeot Renault developed a honeycomb insert that simulated the facial skeleton when placed on a Hybrid II dummy head. When tested, it was found that the force deflection was somewhat the same for the cadaver and the Hybrid II with this insert, but there were also significant differences. At General Motors, Melvin used a deformable foam material to provide relevant characteristics of the facial area. The main purpose, initially, was to provide a correct front for the angular acceleration measurements that General Motors are looking into. By using the accelerometer array that is placed inside the head in order to calculate angular acceleration and by combining this with the neck loads that can be obtained with the Hybrid III dummy, force, magnitude and location can be derived.

Volvo and Collision Safety continued their joint work and presented a deformable Hybrid III load sensing head illustrated in Figure 9. The design goals for this head, the third produced by the joint program, were to be able to determine force location and force magnitude. It was also considered important to record the time histories of the force in order to correlate it with other measurements taken during the test and also with the high speed films recorded during a test. Also it was of interest to maintain the standard dummy head measurements so that this new head could be used together with the standard dummy in the usual tests. The basis for this development was the Allsop data referred to earlier. To measure the forces, 18 load cells were mounted onto the head in groups of 3 as shown in Figure 9; e.g. one of these areas corresponds to the location of the facial bone. By using Allsop's data, it was found that the frontal bone of the Hybrid III dummy actually corresponded rather well to the human frontal bone. This was not very surprising because the Hybrid family dummy heads have been developed to perform human-like when impacted in the forehead area. However for the other facial bones, the aluminium dummy head was too stiff, so some adjustment was necessary. Some of the aluminium was cut away in order to position an insert that would be compliant with the biomechanical data. A large number of materials were considered for the insert and a polyurethane material was found which, though too stiff when it came from the manufacturer, proved satisfactory after holes had been drilled in it. So this work is proceeding.

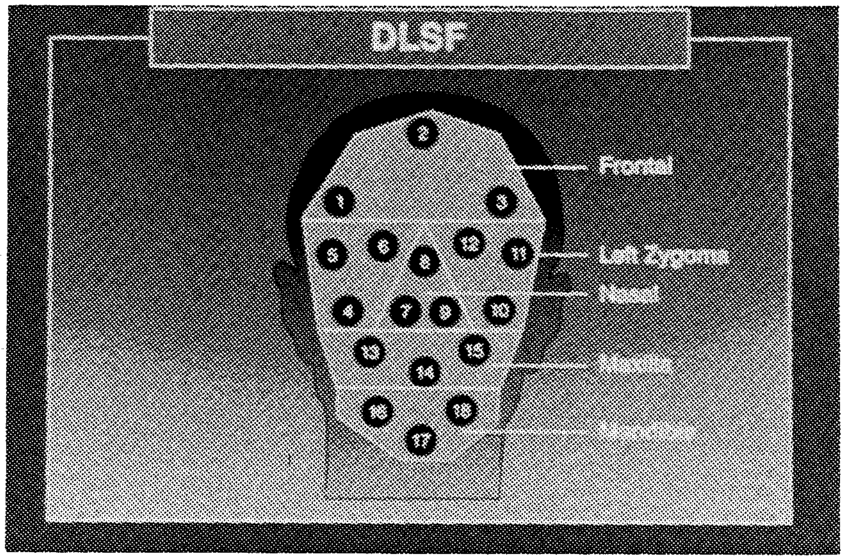


Figure 9: Diagram of deformable load sensing head showing positions of load cells

QUESTIONS/COMMENTS:

Michael Graham: I'd like to ask about the frangible face that you came up with — is that a once-off use? Once you've used it in an impact, is it then re-usable?

Ingrid Planath: Yes, it is, because it's a standard dummy head and the only thing you need to do is exchange the insert with the compliant material. You might not necessarily see anything, but I think it's wise to exchange it.

Ken Digges: Have you used your face device to test any of the proposed steering wheels, for instance the TRRL wheel, and if so, did you find any benefit from the TRRL wheel?

Ingrid Planath: No, we've not done that yet.

Ken Digges: I have a comment on facial injuries. We very frequently discount AIS 1 injuries as not being something that we're worried about, and I think that possibly AIS 1 facial injuries may be an exception in that we very often see AIS 1 facial injuries that can be very devastating, particularly for impacts with the header and sunvisor. In the forehead, the bone is pretty strong but the consequences for the lacerations will be frequently to see the scalp and forehead tissue opened up and three or four or five inches of the skin peeled back and it's a rather bad cosmetic injury.

Laurie Sparks: How do you use your frangible face? Do you use it on a Hybrid III dummy or have you developed a technique to use it separated from the dummy?

Ingrid Planath: One of the advantages with this approach is that you can use it separately, so we have used it in component testing and we have also used it together with a complete Hybrid III dummy.

Keith Seyer: To follow on from the last question, I'd like to comment that we did do some work with New South Wales Crashlab using the deformable load sensing face and in relation to dummy response, with that face fitted to the Hybrid III dummy and the standard Hybrid III dummy face, and it does give a slightly different dummy response. Perhaps Ross Dal Nevo can comment further?

Ross Dal Nevo: Yes, just a brief comment on that. Based on a limited number of tests, the trend that appeared to emerge was that the frangible insert gives a lower peak result in head acceleration because you get the deformation that tends to reduce the g's. But it also spreads the duration of the impact. When that was used to calculate the HIC value we got marginally higher values on some of the results. So it does have an effect, but this is based on about 6 or 8 tests, so we'll be doing more work on that before we draw any major conclusions.

DEVELOPMENT OF AN INFANT IMPACT HEADFORM

Nicholas Shewchenko

This talk will describe some of the work done by Biokinetics and Associates Ltd. in developing an impact headform specifically for the design of infant bicycle helmets. Some of this work had been initiated by one of the manufacturers, Pedal Flight, and has primarily been the result of a lack of current headforms available in North America and Europe.

Some of the specifications for the design of the headform included: looking at what age range of infants would be either riding bicycles or would be passengers on bicycles; looking at differences in anthropometry between males and females; defining the head shape in proper three dimensional co-ordinates; and looking at biomechanical data and existing instrumentation so that the headform could be used as part of the standard testing process.

The first project to look at was basically quantifying the age ranges of infants as passengers. In North America, a minimum age of 6 months is common, as this is basically the minimum age a child can sit up vertically and remain in a seated position. A maximum age was felt to be around 4 years old, when a child preferred to ride a bicycle. In terms of bicycle riders, children are capable of riding tricycles roughly from the age of 3 years old, and at the age of 5 years, 50% of children can ride bicycles. So the age group that was considered ranged from 6 months to in excess of 5 years. When the current headform technology was considered, the smallest headform available was approximately suitable for a 7 year old child so the development needed was for the 6 months to 7 year age range.

To clarify terminology: in this presentation the term *infant* means a range from birth to one year of age; the *toddler* is considered to be from one to 3 years old; and *children* in general range from infancy to puberty. The head circumference growth curves for the 6 month to 7 year old age range in Figure 1 show that in the early stages of growth, head circumference varies considerably and tends to taper off around the 7 year age mark. For this study, 3 headform sizes were chosen (Table 1) to represent the minimum age, the median and the maximum size that would be seen within this age range and these data show that there is roughly a 45 to 53 cm head circumference to be addressed. In this study the first size headform was defined as Size 1, which represents an average 12 month old child. The Size 2 will represent a 2 year old and the Size 3 will represent a larger child. It is hoped that these 3 sizes will cover the whole range for bicycle passengers and riders. The adult population will be addressed with the current headforms, basically available through the ISO specifications and the DOT specifications.

Table 1: Headform size categories

Size	Head circumference	95th percentile	50th percentile	5th percentile
1	450 mm	6 months	12 months	18 months
2	490 mm	1 year	2 years	4-7 years
3	525 mm	2 1/2 years	9 years	--

In looking at available anthropometry, basically two sources are available: 2-dimensional measurements, which are quite common, and 3-dimensional measurements which are available currently primarily as a result of stereophotogrammetric techniques.

The 2-dimensional data is typically represented as head circumferences, width of the head, height of the head — that sort of information. Basically, 3-dimensional data would allow definition, in 3-dimensional space, of the profiles of the head, either in the mid-sagittal plane or the circumference. Figure 2 shows the data that was collected as part of the Consumer Product Safety Commission for quantifying head shapes in 3 dimensions. With this data it was felt that development of the proper definition of infant/child headform shapes would be achieved.

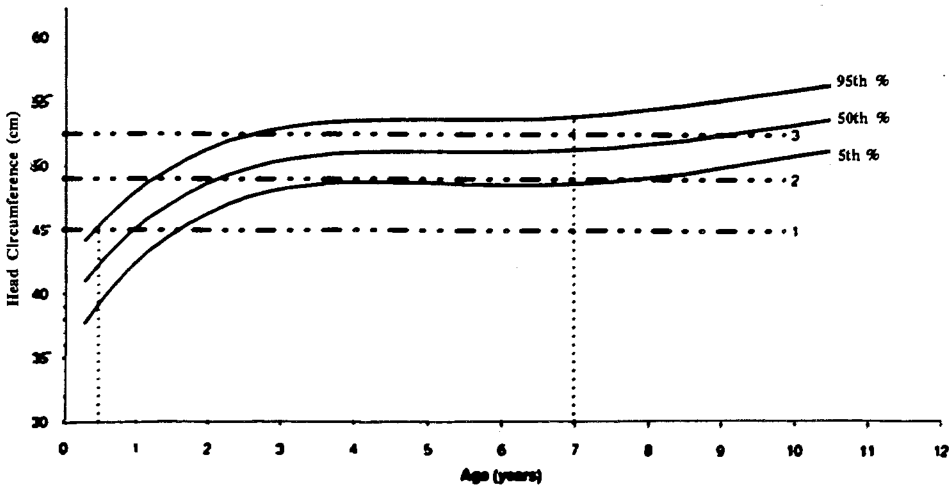


Figure 1: Head Circumference Growth Curves

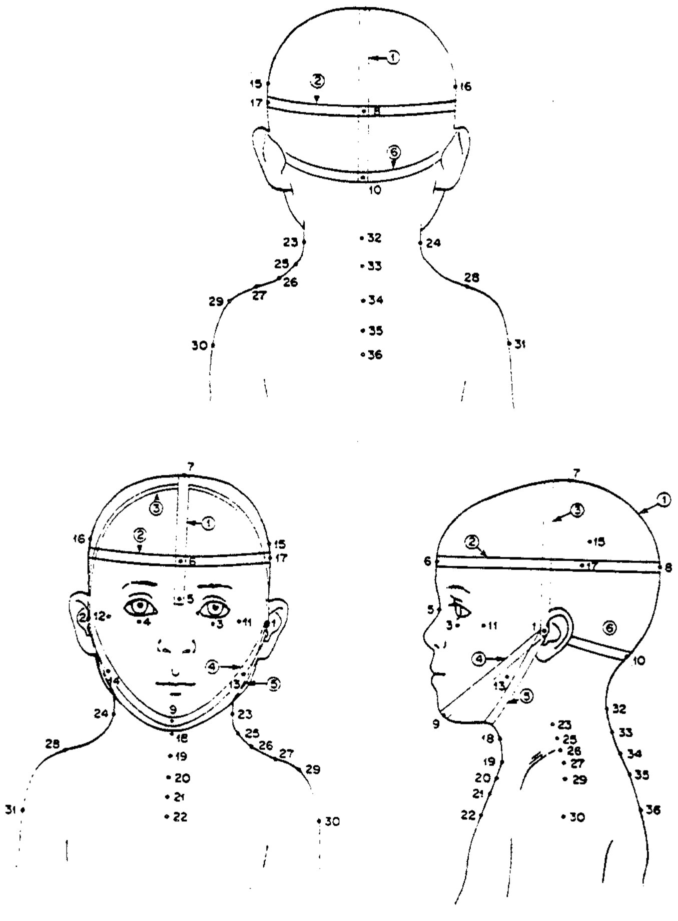


Figure 2: Measurement points used in data collection for quantifying head shapes in 3 dimensions

The process of stereophotogrammetry (Fig. 3) involves 9 cameras, which were used by the Consumer Product Safety Commission, and these photograph the child's head from 9 different view points. With this, 2 image pairs are used to reconstruct 3-dimensional co-ordinates for any point on the child's head. This process allows children's heads to be quantified very quickly, at least in terms of shape, as the basic limitation is the speed of the film, and any shape definition or quantification can be done afterwards. Figure 4 is a typical representation of the 3-dimensional shape information that resulted from the study and Figure 5 is a side view showing the sagittal and coronal planes as well as some of the profiles along the neck and jaw line.

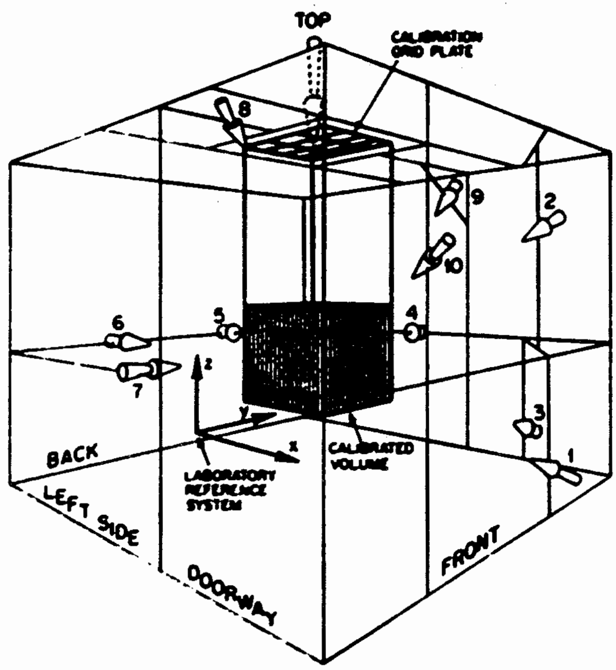


Figure 3: Diagrammatic representation of the process of stereophotogrammetry

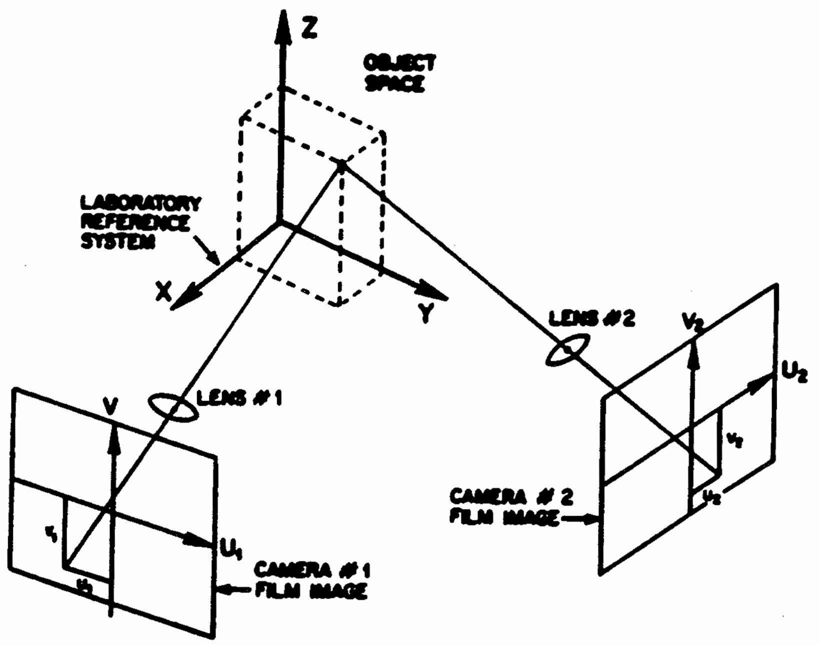


Figure 4: Representation of 3-dimensional shape information from the study

It was felt that one deficiency with this data base was the lack of information in the rear quadrant and for the front temporal areas of the head, so additional information was collected. A very simple system was set up as part of the study at Biokinetics, with 4 cameras viewing the head of the child. The child was fitted with a skull cap (Fig. 6) and profiles were extracted at various points: the children weren't too amused at this process; but this additional data was a useful supplement to the larger data base available to the Consumer Products Safety Commission.

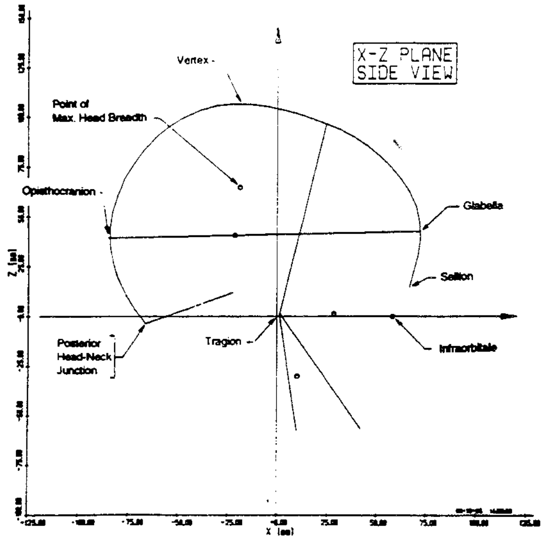


Figure 5: Side view showing sagittal and coronal planes

Figure 6: Measurement of head shape profiles

Now as part of the modelling requirements for the headforms, it was of interest to obtain the proper anthropometric shape and also to define test lines and create a national headform that could be used for helmet testing, as this was the primary goal for developing the headform. In creating the headform from the 3-dimensional data it was decided to use a computer modelling approach, since it is quite easy to manipulate data in 3-D on a computer, these days. The first task was to assimilate both 3-dimensional data bases and this was done by collecting redundant landmark information between the data bases and using a statistical approach to align the headforms. Shape profiles extracted from each data base can supplement each other and be combined to create a brand new headform which would be representative of the children in question.

The profiles available through the Consumer Products Safety Commission were already grouped in 4 age ranges. Figure 7 is just an example of 2 age ranges where the data were grouped in terms of 4 to 6 months and 13 to 18 months. These represent average profiles for those age groups. The Size 1 headform, as an example, lies somewhere between the 2, so the head shape profiles had to be interpolated between the 2 available age ranges. This was done for all 3 head sizes that were proposed. The results in Figure 7 show the mid-sagittal profile. Figure 8 shows the circumferential profile, where profiles were again interpolated and this is the computerised representation of the interpolated data.

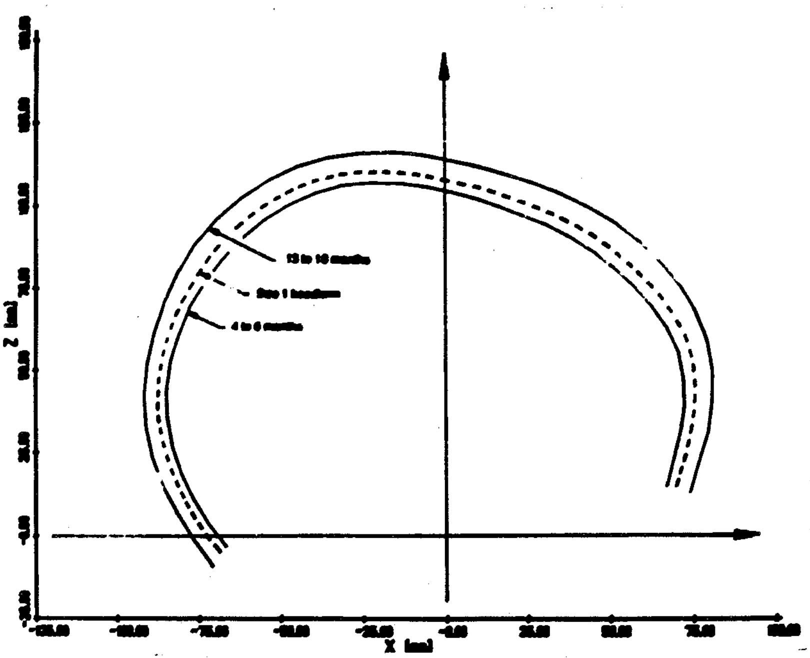


Figure 7: Mid-sagittal profiles for 2 age ranges: 4-6 months (inner line) and 13-18 months (outer line) dotted line = Size 1 headform

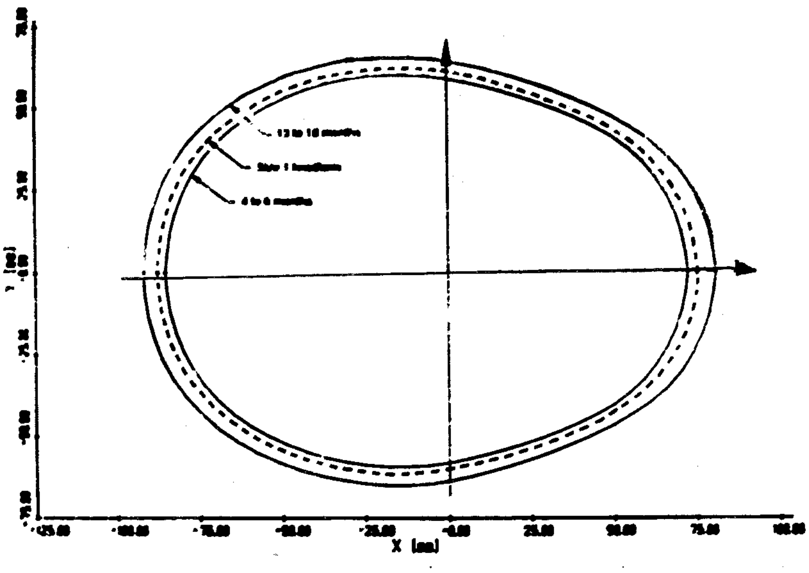


Figure 8: Circumferential profiles for 2 age groups: 4-6 months (inner line) and 13-18 months (outer line) dotted line = Size 1 headform

Profiles had to be extracted using the infant data that was collected to supplement the data base. The first step was to create a surface model (Fig. 9) and then to cut the surface model to extract the appropriate profiles. By combining such results from the 2 data sets, a brand new head shape was derived (Fig. 10), and with this the anthropometry could be quantified quite accurately and the data processed so that a test headform could be developed.

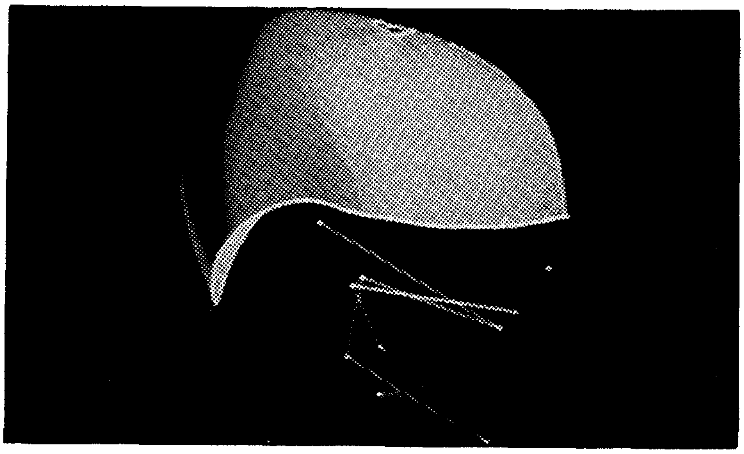


Figure 9: Surface model created from infant data

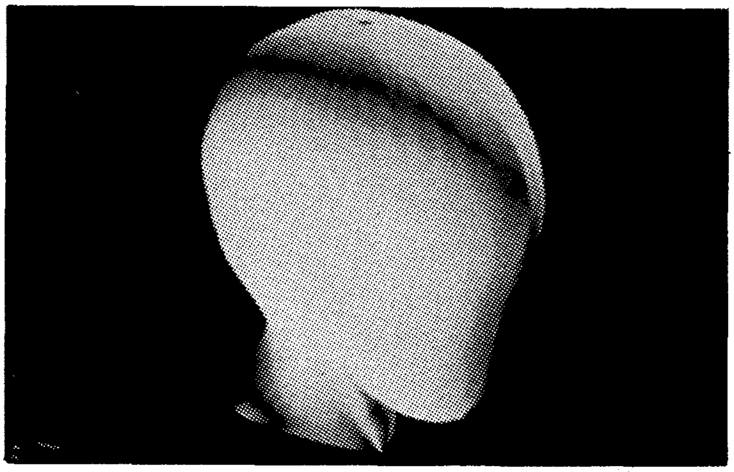


Figure 10: New head shape

Regarding head shape, it should be remembered that the primary reason for doing this project was that children's heads tend to have quite different proportions than those of adults. As can be seen in Figure 11, the cranium is quite bulbous both front and back. The facial features tend to be somewhat receded, with a very small neck circumference. This can be quite important in terms of helmet fit on the child as well as retention capabilities of the helmet.

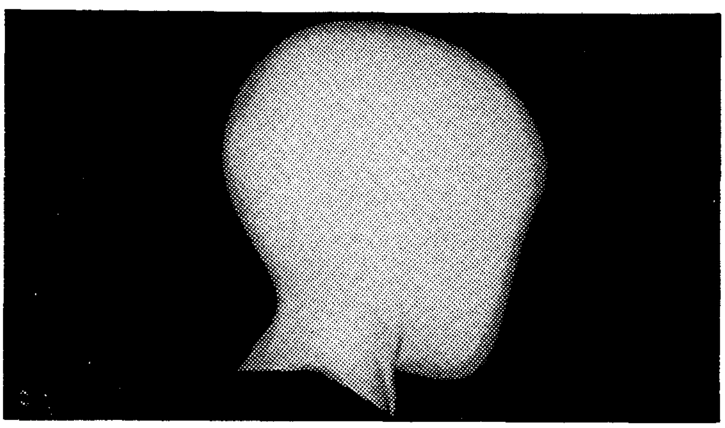
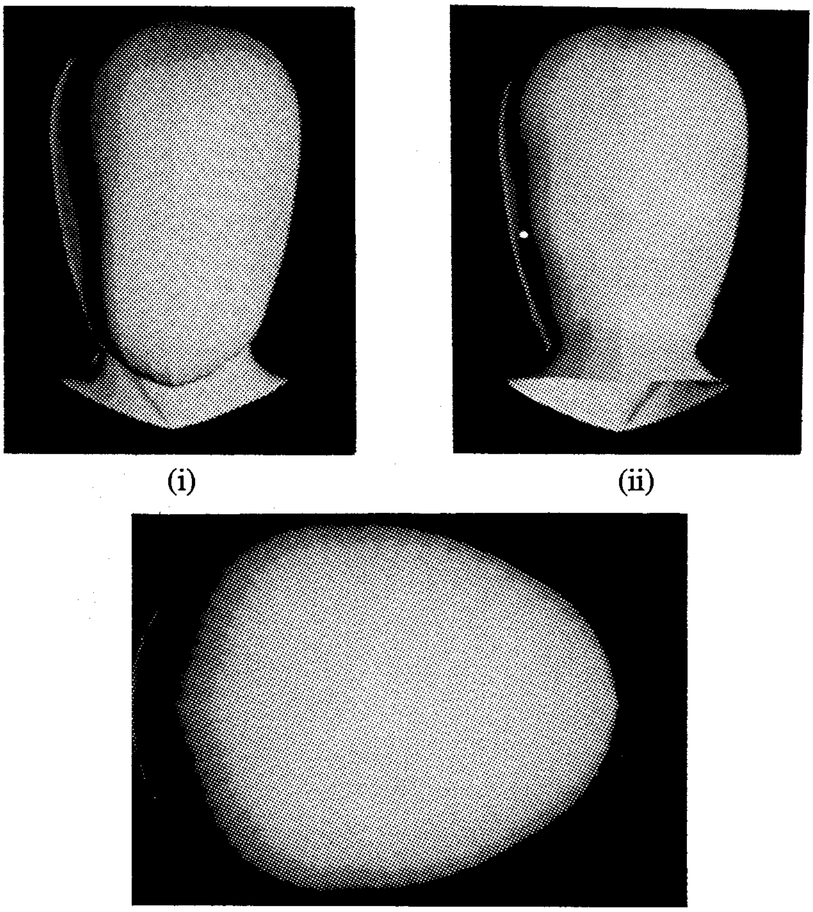


Figure 11: View of head shape showing bulbous nature of child's cranium

Figure 12 displays the front, rear and top views. As can be seen from the view of the top, the front of the head is quite tapered. If the shapes of the IS0 headforms are compared, the profiles tend to be somewhat elliptical and therefore the children's heads tend to bulge in the back and to be quite tapered and pointy towards the forehead region. Figure 13 (i) shows a lateral view of the Size 1 headform which again displays that characteristic head shape; front and rear views are displayed in Figures 13 (ii) and 13 (iii).



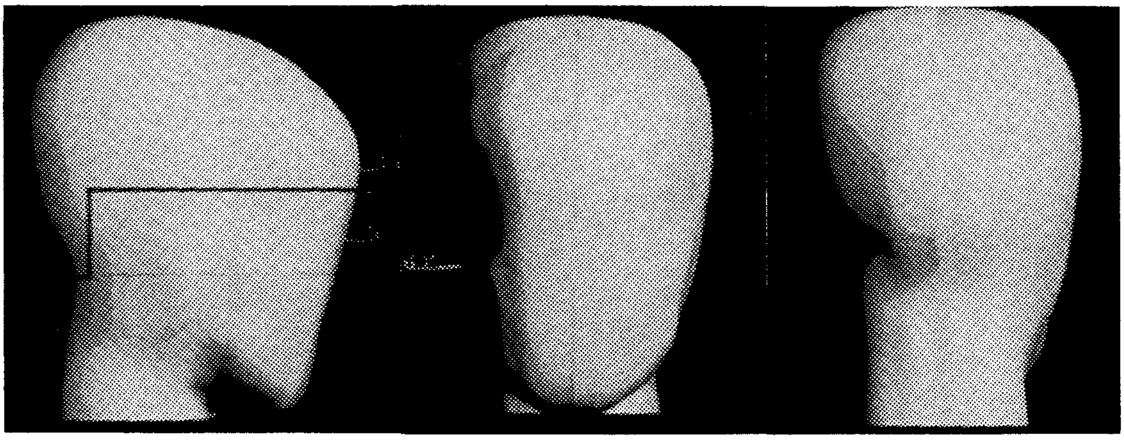
(i) (ii)

(iii)

Figure 12: Front (i), rear (ii) and top (iii) views of headform showing characteristic shape of children's heads

One off-shoot from the computer modelling process is that the test zone for the liner coverage, or the test line, can be defined quite accurately. This can be seen in Figure 13 (i) and also in Figure 14 which shows an example of the test zone on the Size 2 headform. It is placed with respect to anatomical landmarks on the headform. This gives the advantage of choice of landmarks which allows the important parts of the head to be covered in terms of injury prevention.

As discussed earlier, there are some shape differences between the headforms for adults and children. If the head shapes of the ISO A headform and the Size 2 headform are overlayed, it is clear where the differences are between the two headforms (Fig. 15). As mentioned earlier, the forehead tends to protrude a little more and the rear quadrant of the head tends to protrude as well. The opposite side seen in Figure 16 shows that the facial features tend to be quite different from the ISO headforms. Small differences can be seen also in the mid sagittal profile and they tend to range up to approximately 4 mm as a maximum difference. Whether or not these profile differences have an influence on helmet impact performances, experience has shown that the fit of a helmet on a child tends to be greatly dictated by the shape of the helmet.



(i) (ii) (iii)

Figure 13: Lateral (i), front (ii) and rear (iii) views of Size 1 headform

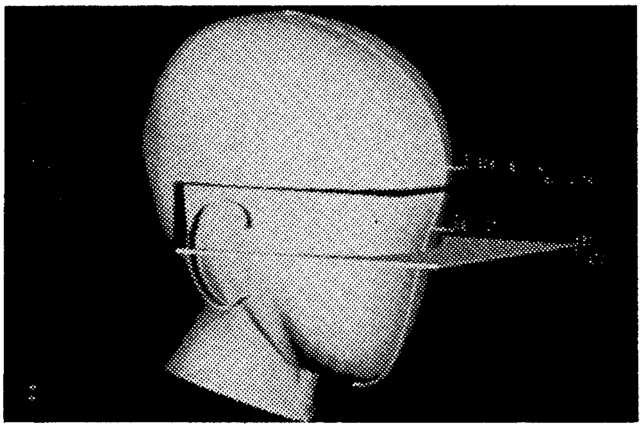


Figure 14: Size 2 headform showing test zone

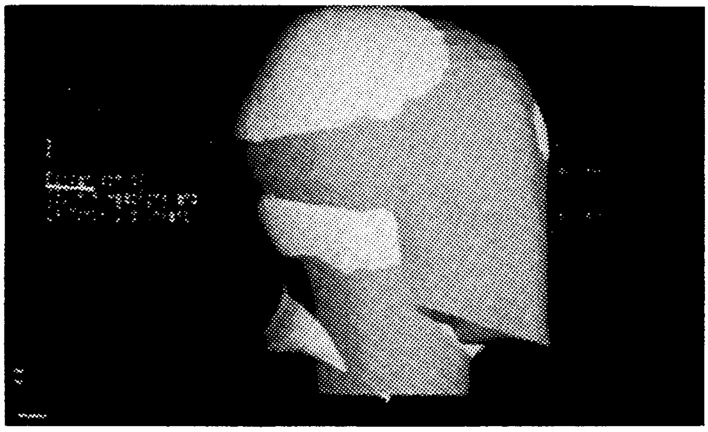


Figure 15: ISO A headform overlayed on new Size 2 headform
(right side view)

Biokinetics and Associates felt that these shape differences justified creating a new headform, and therefore went ahead and developed an actual model of a test headform. As mentioned, in this part of routine testing, a headform is needed which represents not only the anthropometry but also the mass properties, or centre of gravity, essentially, of the headforms, and the configurations should be compatible with existing test equipment.

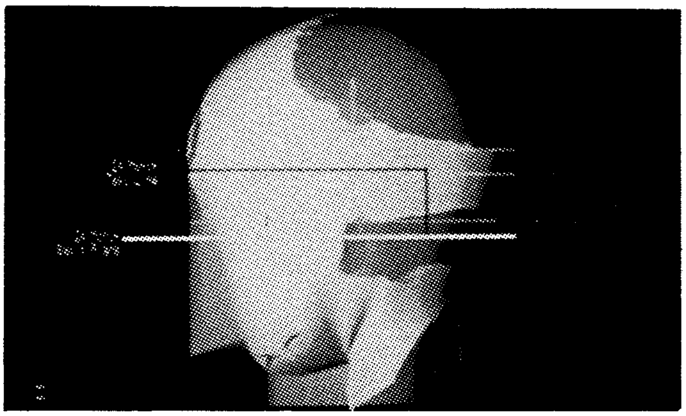


Figure 16: Left side view of ISO A headform overlayed on new headform, showing facial differences

The biomechanical response of the head may also play an important role. The stiffness of a child's head at birth is about 4% of that of an adult; at 6 months it has risen to about 15% and by about 19 years of age, the head has assumed the same stiffness as that of an adult (Fig. 17). So the impact performance of the helmets should really reflect compliance with the child's head as well, and the injury criteria should also reflect this. This still remains to be addressed and it is hoped that it will be a point of future work.

Figure 17: Age-dependent static head stiffness — normalized to adult

In defining the headform shape the standard ball and arm configuration was used (Fig. 18). The solid model of the headform was cut away and the proper centre of gravity location as well as mounting locations for the ball and arm assembly and for the accelerometer were obtained.

These headform specifications are currently being submitted to ISO and it is hoped that they will be included as a supplement to the adult headform population.

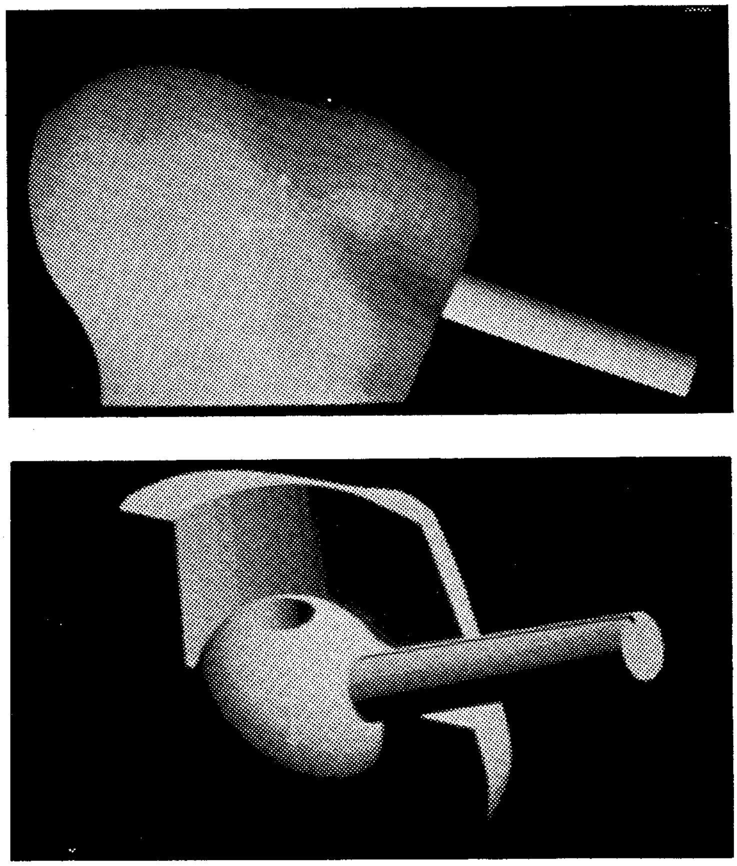


Figure 18: Surface and inside views showing use of ball and arm configuration in definition of the headform shape

QUESTIONS/COMMENTS:

John Lane: I'd like to congratulate Nicholas Shewchenko on that very interesting presentation and ask has the stereophotogrammetric data from the Consumer Products Safety Commission been published?

Nicholas Shewchenko: Yes, it has been published. The data was actually collected by the University of Michigan Transportation Research Institute and it is available through the Institute publications, I believe, as well as detailed 2-dimensional information ranging from birth to 7 years of age or so.

Michael Henderson: Has your own work been published yet?

Nicholas Shewchenko: No, but it has been presented at the ASTM Meeting, I believe, in San Diego as well as in Pittsburg, and the data will be published fairly soon, we hope.

Michael Henderson: Good, because it's very interesting and very relevant here, with infant headforms. You said that one of the objectives of the study was to define where the test line should go, but you rather skipped through that, just showing where it should go. Can you expand on that a little?

Nicholas Shewchenko: The test line that I proposed there was just an example of how one can define test lines. The exact test line is, of course, left up to the Standards Committee developing the specifications, so having the data available, they would be able to justify the placement of the test lines a little better than they have in the past.

Michael Graham: How transportable is the dimensional data you've come up with, in terms of racial differences? I mean, I assume the kids are based on North American children. Do different races have different head shapes, or not?

Nicholas Shewchenko: I'm not too sure of the basis of the data base. I believe that there were 300 children selected as subjects and these were subsamples of a much larger data base and they do represent a cross section of the whole North American market. I'm not aware if there are (racial) differences or not.

THORACIC AND ABDOMINAL INJURIES

Rolf Eppinger

This talk will be a pot pourri of subjects starting with a review of the development of NHTSA's side impact standard. Obviously the usual epidemiology was done, and an understanding of where the injuries occur was sought. The National Accident Sampling System (NASS) data showed that there was a substantial problem in the side impact area (Table 1). When side impact was examined specifically by body region, a variety of involvements of the various anatomical areas was seen, as is shown by the data in Table 2. The head has a significant involvement and the chest has 29%. Therefore, investigation of both head injuries and chest injuries was desirable, but they were pursued through 2 different methodologies.

Table 1: Fatalities by crash mode — car and light truck occupants

Crash type	Percentage of total fatalities
Frontal	43
Side	28
Rollover	23
Rear	3
Unknown	3
Total N	33,067

(Source : 1988 NASS Statistics)

When the injury risk by ΔV was plotted for either AIS 3 or AIS 5 injuries (Fig.1), substantial injuries were seen at 25-30 mph. As an understanding of the problem emerged, some preliminary tests to determine a basic understanding of the mechanisms in side impacts were carried out using some very primitive dummies which were available at that time. The conclusion was reached quickly that side impact was a very drastic event, that the striking car appeared to be almost a rigid device as it struck the very soft side of the impacted car. Accelerometers placed on the interior of the door panels indicated that within 2 inches of displacement they were essentially going at the velocity of the striking vehicle. Basically the distance between the inner panel and the occupant was used up very rapidly, and the occupant was struck by a high velocity, rigid wall. Then a simple repeatable cadaveric test was designed, to run specimens into walls at various velocities, closely simulating what was seen in actual crashes.

Table 2 : Causes of side impact fatalities — passenger car occupants

Body Part	Percentage of total fatalities
Head	47
Chest	29
Neck/Spine	11
Abdomen	8
Unknown	5
Total N	7,676

(Source : FMVSS 214 Notice of Proposed Rule Making)

How and what would be measured on the cadaver, in terms of engineering measurements, was the subject of considerable planning. An array of accelerometers was designed to be placed on the thorax: ventrally, there were 2 on the sternum in the anterior-posterior direction; there were several on the lateral aspects of the upper ribs and the lower ribs; and dorsally, accelerometers were on T1 and eventually on T12 both anterior/posterior and left to right. This was arrived at primarily because the alternative measurements would be through photographic measurements of targets but that was difficult because of the padded environment; good camera coverage is difficult to obtain on the struck side and is impeded by seating so that if there is rotation of the specimen, a reasonable deflection is hard to measure. It was felt that if this global response of the thorax could be captured by these various accelerometers, there was a fairly good chance of developing some sort of relationship between these acceleration parameters and the injuries observed through autopsy.

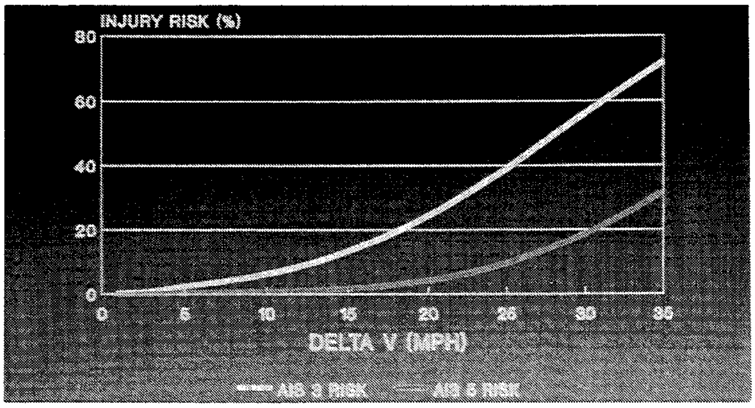
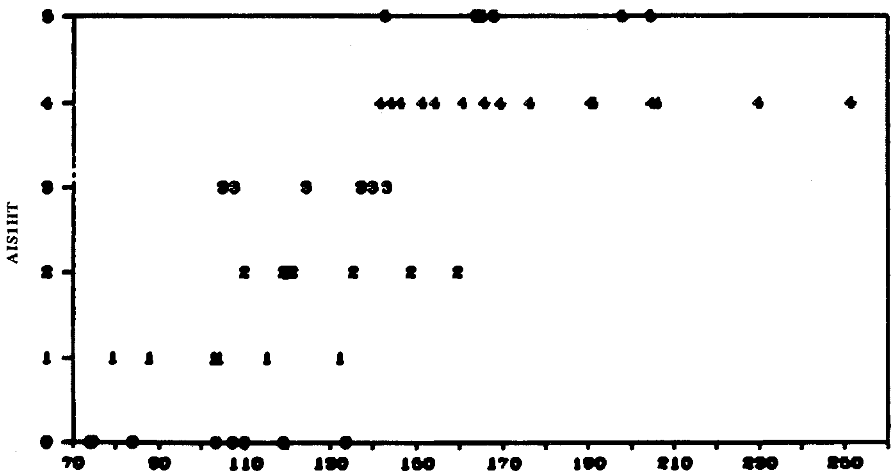


Figure 1: Injury risk vs ΔV
Hard thorax - near side
(Source: 214 PRIA)

After accumulating and analysing the data from over 80 tests of cadavers at a variety of institutions, certain types of peak lateral accelerations were selected, to see how well they would correlate with injury. Through a variety of statistical techniques, the formulation at the bottom of Figure 2 was derived, which uses 2 accelerations (of the lateral T12 spine and the struck-side rib). Using an average of those accelerations, it was found necessary to compensate for the difference in mass of the various subjects, as light specimens seemed to have a higher acceleration tolerance than larger mass subjects. Age also had a very predominant effect on the process. Now, age is a sort of surrogate variable for things such as osteoporotic conditions of elderly people, so really what is happening is that the structure is changing through life but these changes seem to be proportional to age. Therefore age can be used as a substitute. Obviously an alternative way would be to look at certain material properties of the specimens and characterise the tolerance as a function of those. But then it would be necessary to know how these material properties change with the population; again it's a two layered effect ending with a return to age as an independent variable.

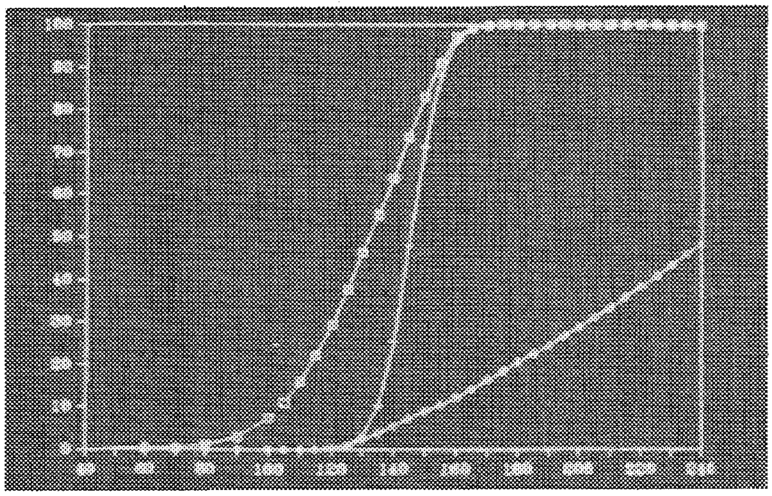


1.4 Age+ 0.5(T12y+ Lury) * Mass/165

Figure 2: Relationship between peak lateral accelerations (x axis) and injury level (y axis)

Figure 2 shows the overall thoracic injury. The normal thoracic AIS evaluations are for injuries above the diaphragm and abdominal injuries are defined as injuries below the diaphragm. It was felt that if the responses of the thoracic rib cage were monitored and controlled, protection could be provided for all organs and systems that lie within the rib cage. That is the reason for the hard thorax definition. Therefore, it was decided to include liver, spleen and kidney injuries that were seen so that the AIS reading was basically the highest hard thorax AIS as a function of the parameters age, acceleration and mass of the subject. The experimental data now gets sorted out in this way. The method shows a monotonically increasing process — in other words, it is consistent with what is wanted in a performance parameter because as it increases, severity increases and there is grouping of the various injury severity levels. A perfect indicator would have each of the AIS levels grouped without overlap. Obviously, with a function like this, not everything is being explained, so that's why statistics are reverted to rather than just a strict relationship between the parameters. However, the method was felt to explain the injury process to some extent.

When probability curves were developed for different levels of AIS as a function of this parameter which has been called the Thoracic Trauma Index (TTI), the distributions shown in Figure 3 were obtained.



TTI (Age + 0.5 * (Lury + T12y) * Mass/165)

Figure 3: Injury risk vs. TTI
AIS1HT ≥ 3 & AIS1HT ≥ 5 FAIRED

In other words, if the two accelerations and the mass and age of the subject were known, then if the TTI was 120, for example, there were be a 30% chance of AIS 3 or greater, and very little chance of injuries in the higher areas. If the TTI was 140 or 150, the possibility for AIS 5 or greater would be about 5%, for AIS 4 and greater about 30%, and for AIS 3 and greater, about 60-65%. The analysis was then repeated using the greatest of the rib accelerations. In other words, when there were accelerometers on the 4th rib and the 8th rib, the maximum peak acceleration that was observed for either one of those two was combined with the lateral spinal acceleration. The result, which is now in the regulation, is as follows:

THORACIC TRAUMA INDEX

<u>TTI PREDICTS PROBABILITY OF INJURY</u>

$$TTI = 1.4 * Age + 0.5 * (Gr + Gs) * Body\ Mass / 165$$

Where:
Gr = Greater of rib acceleration peaks
Gs = Lower spine peak

<u>TTI(d) IS THE REGULATORY CRITERION</u>

$$TTI(d) = 0.5 * (Gr + Gs)$$

Strictly this says that the maximum spine and the greatest of the lateral rib accelerations are monitored on the dummy: TTI(d) stands for 'TTI dummy' and is devoid of the age and mass factors. The probability of a particular AIS injury level occurring given a TTI(d) level can be calculated if the age distribution of the population-at-risk is known as it is with side impacts in the U.S. The result of such calculations are shown in Figure 4. The figure indicates that if all the current population-at-risk were exposed to a TTI(d) = 100, then 5% would experience an AIS 5 or greater and 65% would experience an AIS 3 or greater.

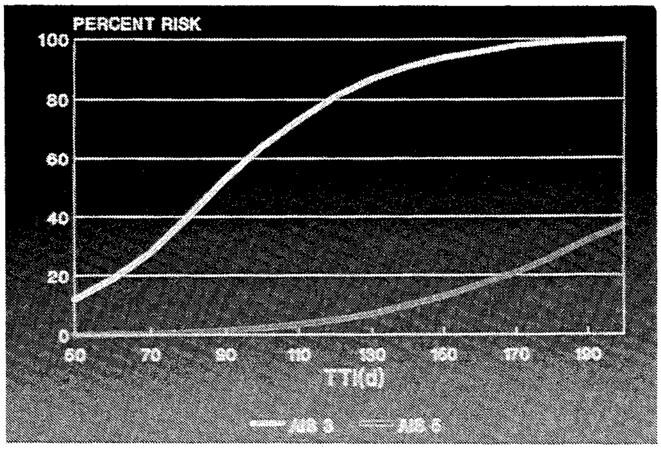


Figure 4: Chest injury vs. TTI

NHTSA then became involved in the development of the Side Impact Dummy (SID) (Fig. 5)

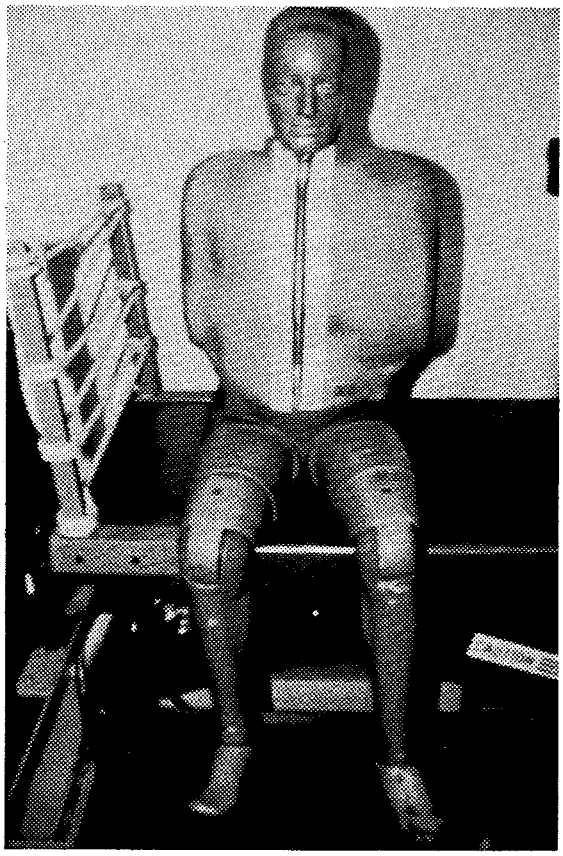


Figure 5: The SID dummy

This is a device that attempts to mimic the acceleration/time histories that were seen on the cadaver runs. NHTSA had 3 different test conditions: a 20 mph rigid wall; a 15 mph rigid wall; and a 20 mph padded wall. Multiple runs of cadavers were performed in each of these conditions, at least 3 or 4, and average acceleration profiles were developed. These became the design specifications for the test device. The dummy reasonably mimics the acceleration/time histories in those regimes. It does deviate slightly from some of the profiles and it deviates the most when it gets into extremely rigid conditions. But in the conditions of the regulation, for example, an event that produces a maximum TTI of 85 in a 33 mph lateral impact into the dummy, the response looks similar to the 20 mph padded wall test where the SID dummy had very good acceleration similitude with the cadaver specifications. Therefore, because the dummy possesses acceleration similitude, the dummy can now be introduced as a surrogate for the cadaver in a test environment, and the criteria that have been developed can be used. NHTSA has used its moving deformable barrier and a SID in a variety of different vehicles and the large variety of performance of different vehicles can be seen in Figure 6, ranging from a (then) American Motors Concord, with a TTI of approximately 70, to a Nissan Sentra with about 150. By calculating the number of injuries produced by baseline vehicles and subtracting the number of injuries that would be produced if all vehicles would have a TTI of 80 or less, the benefits of a regulatory level could be estimated. This process became the basis of NHTSA's benefit analysis for side impact.

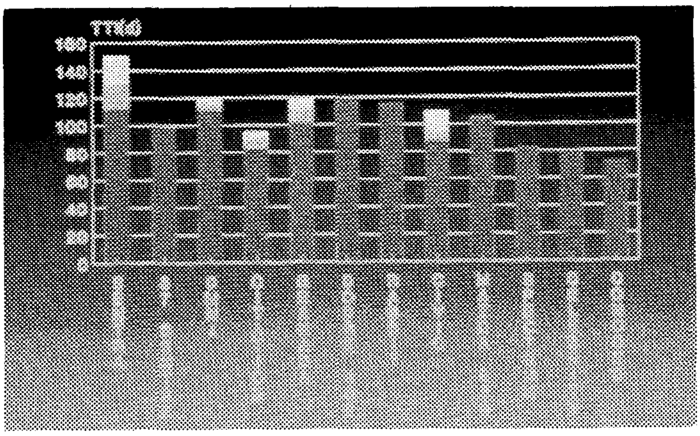


Figure 6: Side impact tests on production vehicles
(Names of car models, from left to right, are:
Sentra, 87 Sentra, Omni, Citation, Granada,
Dodge 400, Rabbit, Civic, Mazda 626,
Spectrum, Celebrity, Concord)

Obviously when the notice for the side impact rule was put out, a variety of comments were received, and the dummy and the injury criteria became fairly controversial topics. Some of the statements that were made about TTI in the responses were as follows:

- NHTSA claimed that it correlated well with thoracic and abdominal AIS;
- others said that it does not correspond to time of injury occurrence. NHTSA considered this immaterial, as long as it predicted the ultimate AIS level. Because it does not give time of injury, TTI is a slightly less refined type of criterion, but it still allows a restraint designer to go ahead and accomplish his task. The actual injury risk of the crash event is produced and this can then be modified. The designer does not have as much guidance as if the criterion actually pin pointed the injury in time, but it was felt that this was not a necessary and sufficient requirement for an injury criterion;
- other comments were that TTI was not specifically associated with any local body phenomenon, either stress or strain;
- it lacked a biomedical basis (what ever that means); and
- accelerations do not reliably describe injury risk.

NHTSA's competitor in terms of injury criteria was the Viscous Criterion (V*C), and various statements and claims made for that were that:

- it was associated with maximum energy dissipated by viscous elements of the torso;
- it related to the actual aetiology of the injury;
- it successfully indicated the time of highest injury risk;
- it was related not to the viscosity of the thorax but to peak (elastic) energy storing rate.

Clearly there were several parallels between V*C and TTI: they were both performance functions that were derived from measurements made on the surface. In other words, the TTI is from acceleration measurements based on surface responses, and the V*C is essentially a relative displacement between two points on the surface. It is not always clear if the displacement is between a struck-side rib and the spine, or the struck-side rib and the far-side rib. Both versions of that have been seen.

But for either criterion, the proposition is really "can surface derived measures such as local accelerations and relative velocities and/or displacements be predictive of local internal material state variables traditionally associated with failure?"

In order to see if something like that could be analysed, it was decided to interrogate a mathematical model as a fundamental approach to this (Fig. 7). It was a very simplistic model but the aim was to configure this to have some sort of impact process, to simulate a wall impact for either a solid mass of 6400 lbs or a pendulum of 64 lbs. There was padding of the wall with a variety of stiffnesses, from 400 to 6400 lbf/in. The model is a distributed, lumped mass model, with a linear spring, and a linear damper between each of the masses. Five of these masses were of 9 lbs and the sixth was 18 lbs, so they had a total of 63 lbs of mass. The stiffness between each chest mass was 1500 lbs/in. The combination of all those springs in series represents closely the overall thoracic stiffness of cadavers. Some operational values for the viscous components were also made. The distance between adjacent masses was 2 inches at time zero (this made a 10 inch wide structure). The model is capable of calculating local state variables such as strain, stress, elastic stress, viscous stress, local strain, energy densities, how much energy gets put into the springs and the damper, the local viscous strain energy density, total absorbed energy, and the viscous absorbed energy for a variety of test conditions.

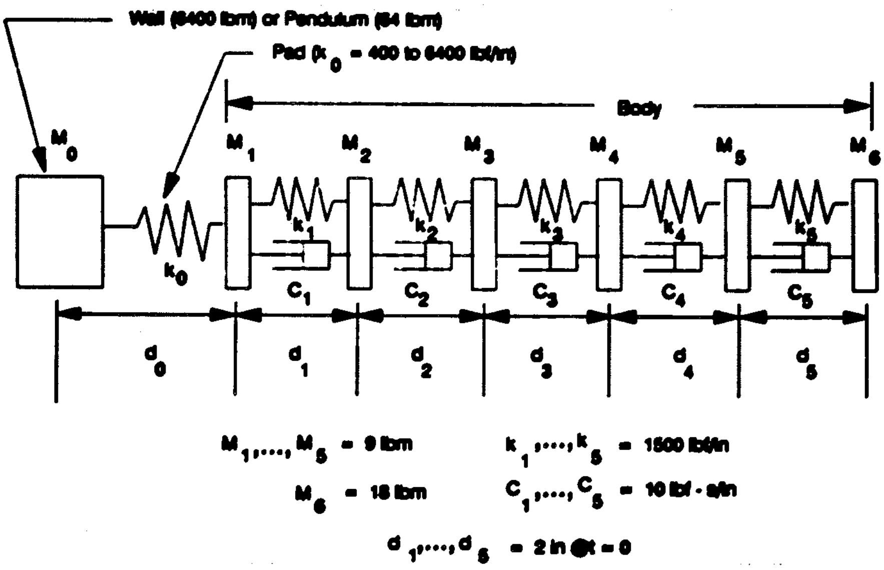
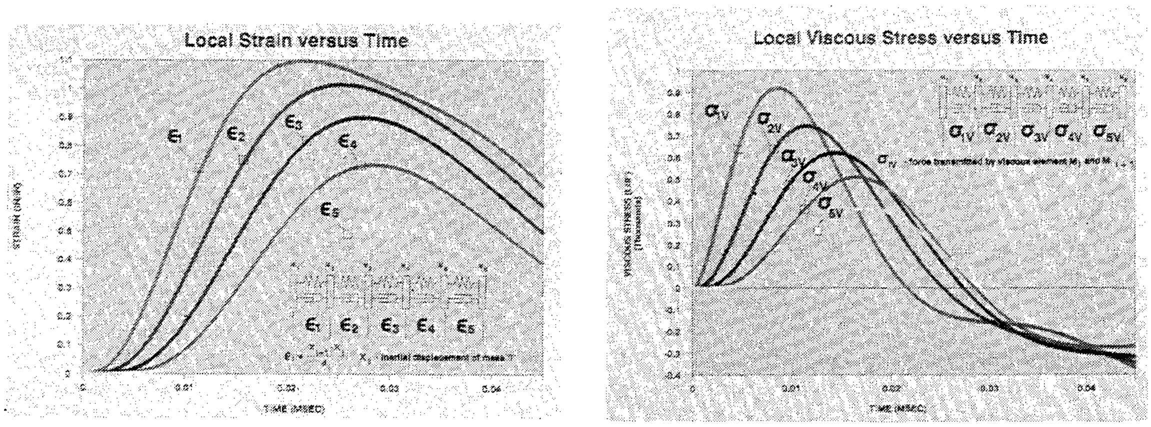


Figure 7: Configuration of analytical model

It was felt that if the various claims were demonstrated on the simple model, then extension to the true anatomical structures of the human being is possibly true. But if the various claims could not be demonstrated on this very simple model, then extension to the anatomical structures was very unlikely. So an analytical test matrix was developed in which both the wall and the pendulum tests were simulated. Five interface stiffnesses between the wall and the pendulum and the model of the thorax were used, and 3 different velocities of impact were investigated.

The first thing noticed was that all things do not occur at the same time. The impact process is a propagating process and, for example, the peak strain at the near-side point occurred at a substantially different time from the peak further into the model (Fig. 8). Therefore, it could be claimed that if strain were the factor causing the injury, there would be a sequence of events in time with injuries occurring as the strain developed at particular places. It could not occur all at once. This suggests that measurement of the maximum risk exactly at the correct time, as was claimed for V*C, would be a false result, because things just don't all happen at the same time. The results for stress included in Figure 8 show the same type of thing occurring — the maximum stress at certain points occurs at different points in time.



(i) (ii)

Figure 8: Local strain (i) and local viscous stress (ii) vs. time

The relationships between peak local stress and TTI, V*C and deflection were then measured at various velocities and with pad stiffnesses ranging from the softest to very stiff. Figure 9 shows the extent to which each of the measures correlated with peak local stress at test velocities of 10, 15 and 20 mph.

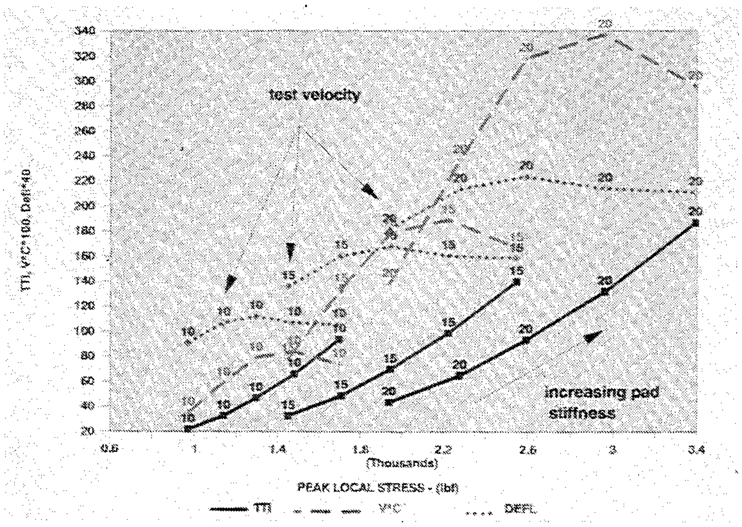
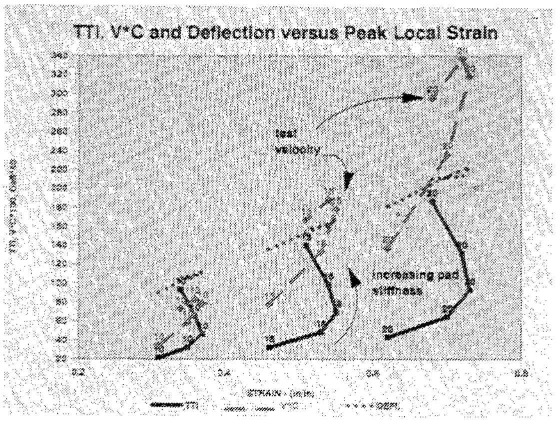
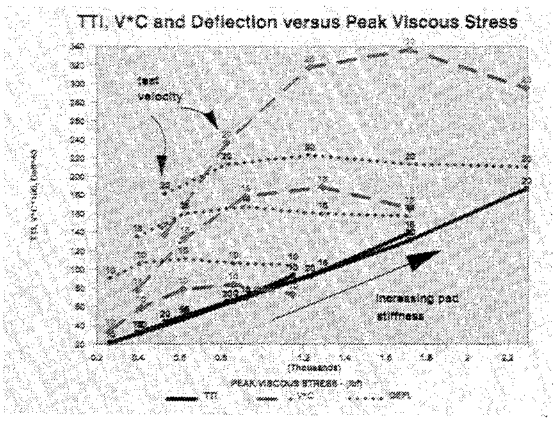


Figure 9: TTI, V*C and deflection vs. peak local stress

For peak local strain, overall displacement correlated very well, whereas both the TTI and the V*C just were not able to predict any local strain very well (Fig. 10, i). With viscous stress, however (Fig. 10, ii), TTI correlated very well but neither deflection nor V*C predicted that particular parameter.



(i)



(ii)

Figure 10: TTI, V*C and deflection vs. peak local strain (i) and peak viscous stress (ii)

These experiments were not attempts to promote any one specific parameter as the vital one, but merely an investigation of how well various external parameters could predict any one of several possible internal parameters. Figure 11 shows that for maximum absorbed energy, which is the sum of viscous and elastic stress, some of the parameters correlate well and some don't. When V*C was tested with peak local stress, the results for pendulum tests differed from those for wall tests, (Fig. 12). However similar tests for TTI versus maximum local viscous stress showed very good agreement between the two types of tests (Fig. 13). Therefore, a particular value of a TTI will always predict the same thing, an ideal characteristic for a criterion.

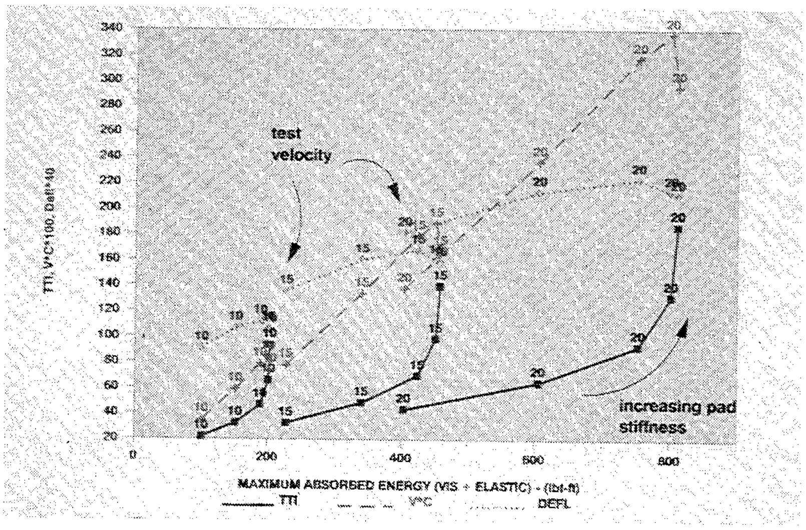


Figure 11: TTI, V*C and deflection vs. maximum total absorbed energy

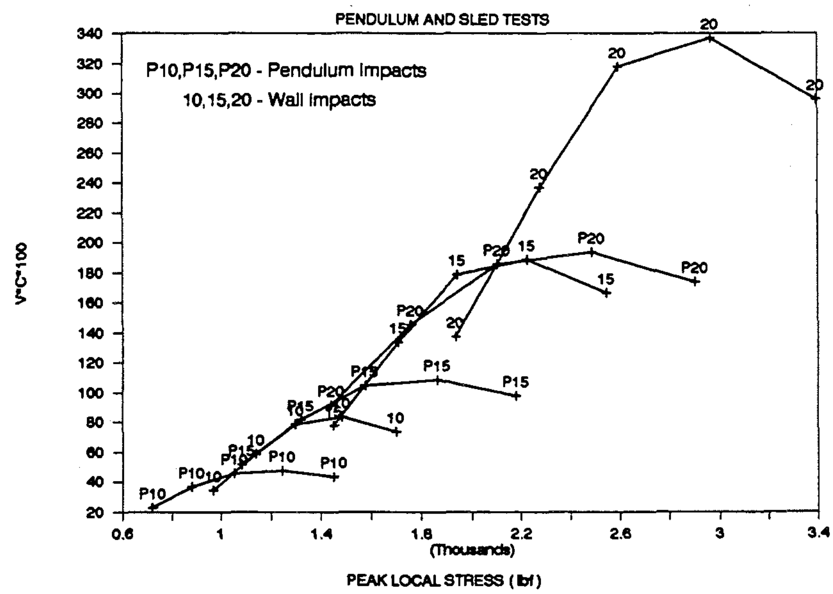


Figure 12: V*C vs. peak local stress

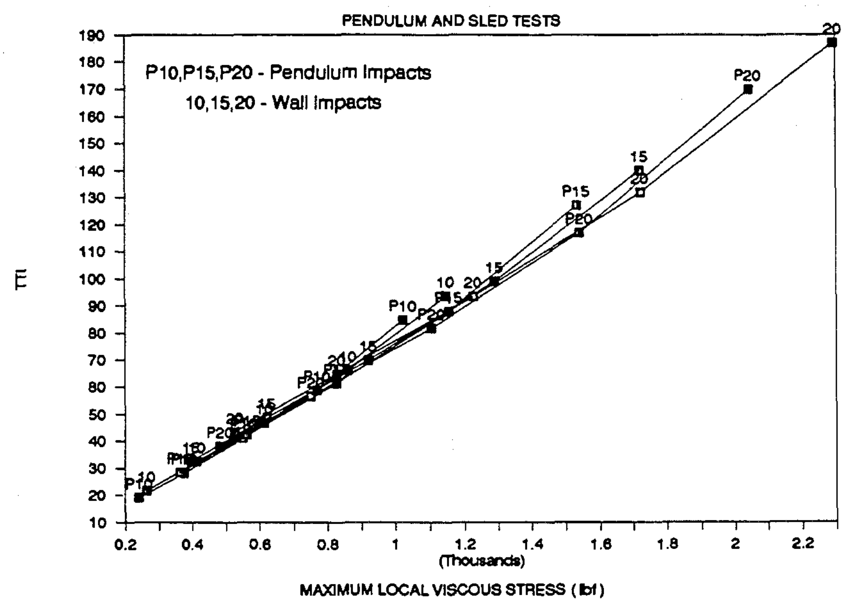


Figure 13: TTI vs. maximum local viscous stress

In summary, it was decided that external indices had different relationships to actual aetiology. TTI correlated well with peak viscous stress, deflection with peak local strain, and V*C did not correlate well with any of the parameters that were examined. Temporal relationships of the indices to failure were not really demonstrated with this model, because for these variables, the peaks occurred actually at a variety of times during the impact event and therefore it was not possible to claim that any one of these parameters actually corresponded to the point in time where peak values occurred. Each one of those parameters, the V*C, deflection and TTI, did correlate with certain types of engineering parameters that historically had been associated with failure, so this exercise did not refute any. The claim that V*C correlated with viscous variables was supported. Claims that TTI lacked biomedical basis and that accelerations did not describe risk were not supportable by this study because certain engineering parameters were found within that model with which TTI correlated very well. In other words, TTI correlated well with several local and overall state variables.

The conclusions from the work were that: all external surface functions were still possible injury criteria candidates; none had a strict theoretical linkage that has been demonstrated; all were derived by empirical processes and therefore the evaluation of the efficacy of any of these functions would be established only by how well it fitted the available experimental data.

Another aspect of the development of NHTSA's side impact standard was the type of dummy used. After SID had been developed, some prototypes of both BIOSID and EUROSID became available and the capabilities of these devices had to be compared, to ensure the SID was still suitable to use. As SID was a well tested device, NHTSA had a long experience with it during the development programs, and it was well documented in terms of drawings and other aspects that would have to be used in order to promulgate it as a test device, there was an inclination to use the SID as the test device of choice at that time. A series of comparative tests was carried out, using the 3 test devices. The test matrix on the sled used 20 mph wall tests with 3 inches of padding and different stiffnesses of that padding with the aim of finding the optimum stiffness that would minimise TTI. The series of sled tests were run with 10 different pads of varying stiffness. Figure 14 shows that the SID and BIOSID devices gave essentially the same result, that padding somewhere between 10-20 psi gave optimum performance. When the EUROSID was tested, a similar conclusion was reached. So there was confidence that, although SID did not have as many features as the newer prototypes, the SID, if used in a regulatory process, would drive design to the same essential optimal point as the other two dummies would.

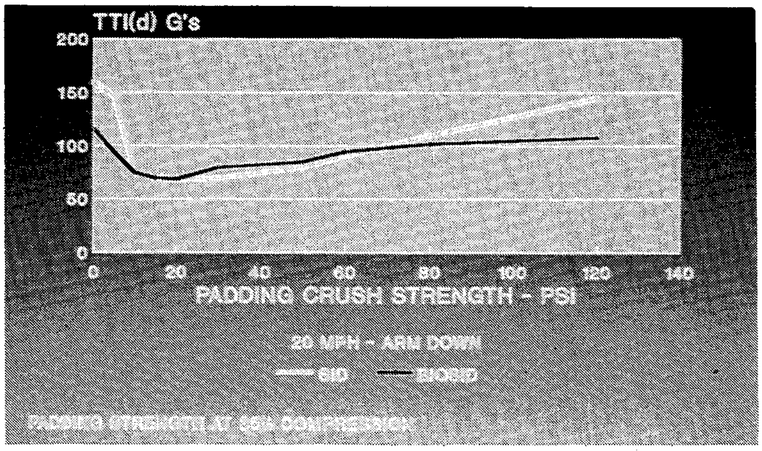


Figure 14: SID/BIOSID comparison
Padding selection for minimum TTI

Ultimately, NHTSA's conditions and criteria that were promulgated for passenger cars are:

NEW SIDE IMPACT STANDARD 214
(passenger cars)

NHTSA BARRIER (moveable, deformable) — 3,000 lbs; crabbed 27 degrees
TEST SPEED — 33.5 mph (15/30 mph struck/striking vehicle)
NHTSA SID DUMMY — front and rear seating positions on the struck side
CHEST CRITERIA — TTI(d) 85 for 4 door cars
90 for 2 door cars
PELVIS CRITERIA — 130 g's
STRUCK DOOR SHALL NOT SEPARATE TOTALLY
OTHER LATCHES AND HINGES SHALL NOT DISENGAGE

A variety of benefit analyses have been carried out and anyone who is really interested in that should refer to the final regulatory analysis on how they were done. Estimates of the fatality reductions and other benefits have been made and the benefits were sufficient for NHTSA to promulgate the rule which is now the standing rule. In about a year the phasing-in of that process will begin.

Some details about how the chest band works and why it was developed may be of interest. Basically when the side impact study was undertaken, an array of accelerometers was used and, in fact, at that time most biomechanical experimentation had been limited to accelerometers, displacements between two points, and photogrammetry. From a theoretical study at NHTSA, it was thought that the concept of defining surface responses to predict internal responses had a very strong basis. It was decided that a good way to define a curve representing a cross section of a surface would be as a function of curvature and distance along the curve rather than in cartesian coordinates as a set of x,y points. Thus, by moving along a closed curve from a point, the origin, and measuring the distance along the curve and defining the curvature at every point along the curve, that particular geometry can be characterised uniquely (Fig. 15).

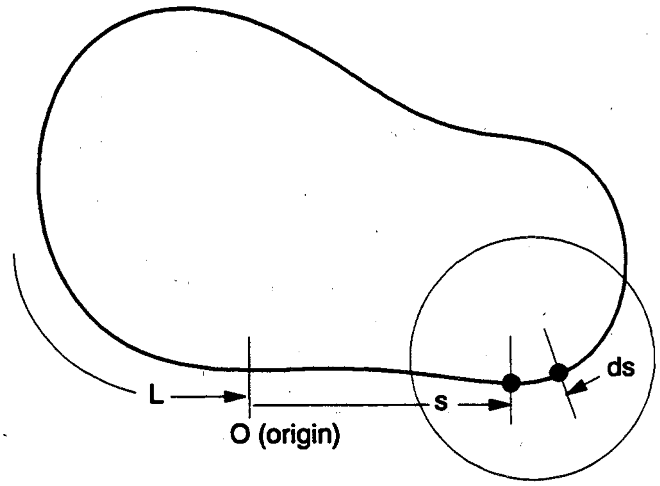


Figure 15: Coordinate system along closed curve of length "L"

Now curvature is defined as the change in direction per unit travel along the curve. Consider a tangent vector at a point *s* on the curve as shown in Figure 16. At a second point, *ds*, slightly down the length of the curve, the direction of the tangent vector has changed. If the change in direction is defined as $d\phi$ and the movement along the curve as *ds*, the curvature is defined as $d\phi/ds$. The reciprocal of this value is called the radius of curvature.

$$k(s) = \frac{d\Phi}{ds}$$

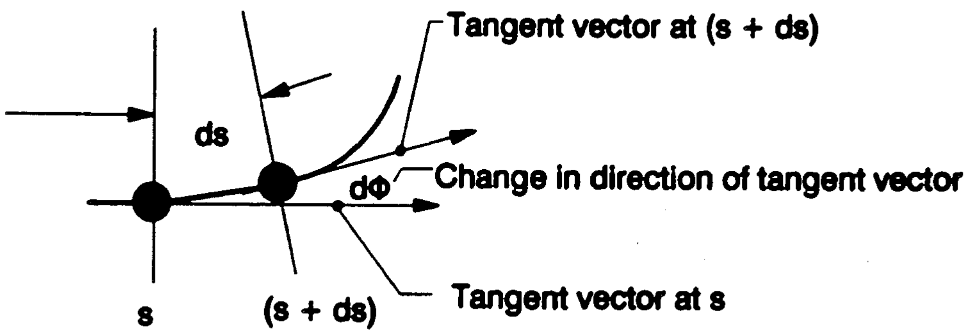


Figure 16: Determination of curvature at location "s"

The radius of curvature for a flat line is infinity, and it keeps getting smaller with increasing tightness of a curve. Curvature, being just the reciprocal of that, is zero for an absolutely straight line and it increases as a curve becomes tighter. When a closed curve is characterised by the *k(s)* process (Fig. 17), the reverse of the process can be used to generate the actual curve. For example, for movement from a point *s* to a new position *(s + ds)*, the area underneath the *k(s)* section of curve, the integral shown in Figure 17, is actually a measure of how much the direction has changed. This can be likened to a race car going around a racecourse: by driving and turning the steering wheel, a path can be followed completely around and the car returns to the same point from which it started. Now if, at every point along the way, recordings are made of distance from the original point and the steering wheel angle (which is proportional to curvature), the shape of the course could be recreated at some other place by just driving along and, at the distances measured, applying the same amount of steering wheel angle as previously recorded. This is the process that NHTSA has used.

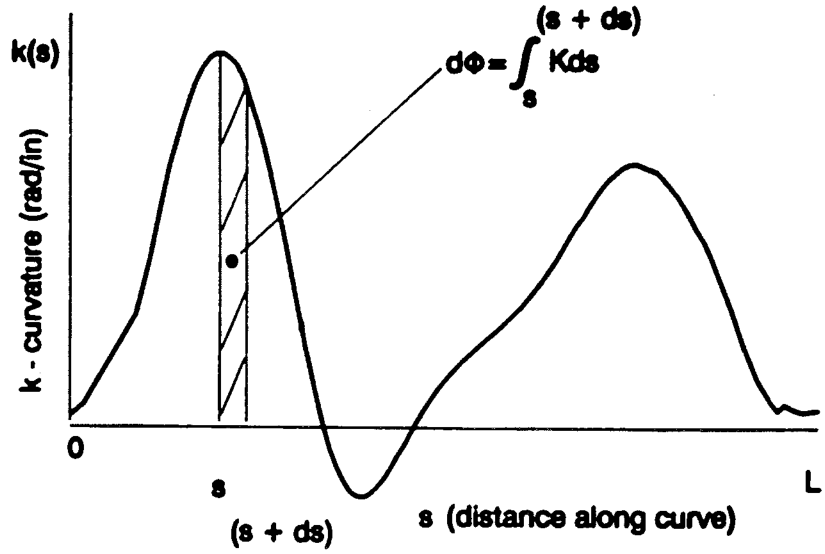


Figure 17: Closed curve characterized by k-s method

A representation of the integral of the *k-(s)* curve is shown in Figure 18. Since this represents a closed curve, if one traverses the length of the curve, the tangent vector must rotate through 2 pi radians of rotation. That is a constraint for a closed curve.

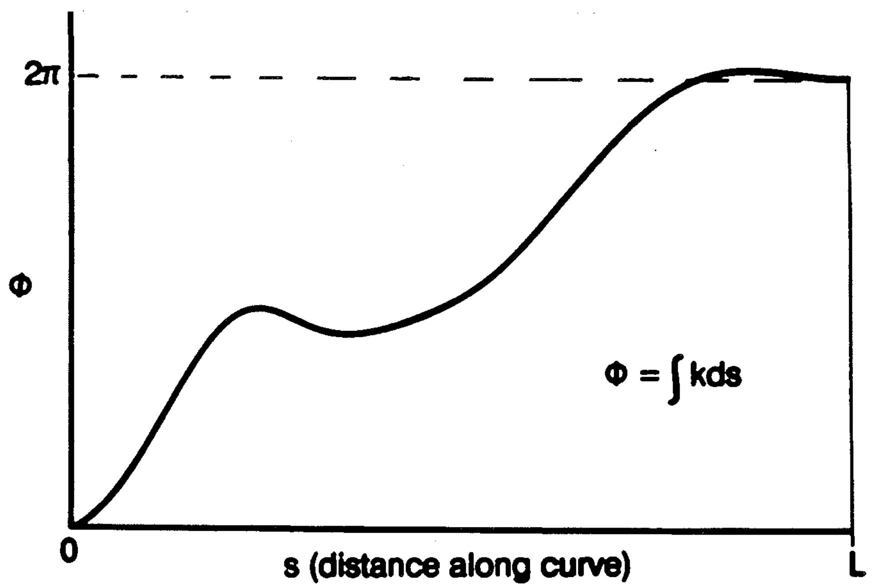


Figure 18: Integral of k-s curve

To implement this theory, a test device was created with a series of gauges along its length. Figure 19 shows an early configuration in which there was 3 inch and 4 inch spacing between the various gauge sites. The band was built of steel shim stock 0.010 inch thick and 0.5 inch wide (Fig. 20). At each bridge site 2 gauges were placed on top and 2 on the bottom, configured in such a pattern that would sense curvature when the upper gauges went into tension and the bottom ones into compression due to bending. Each bridge was calibrated to give a voltage proportional to curvature as an output. The bridge configuration also nullifies any tension effects so the band can be put on and even if there are tensile forces present they do not upset the determination of the curvature.

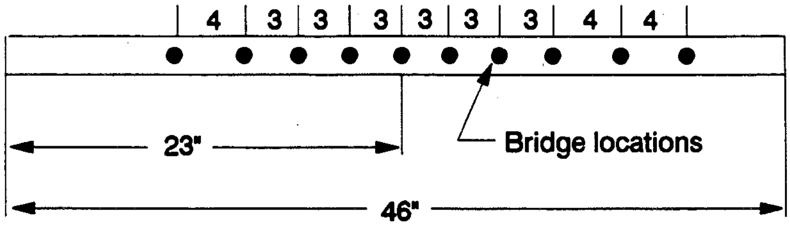


Figure 19: Overall layout of sensing band

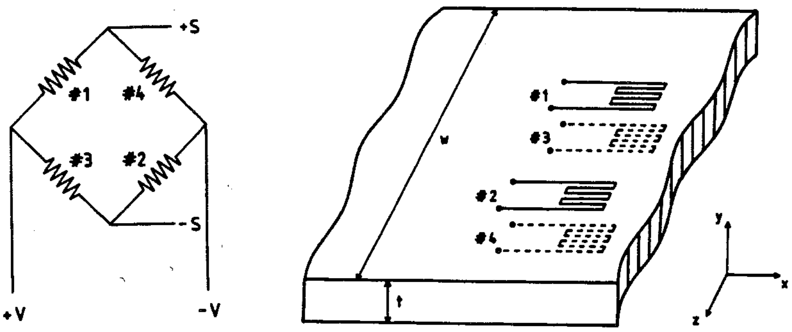


Figure 20: Gauge locations and bridge configuration

Thus, with the band attached to a structure, its curvature is known at every gauge site. So if there are 10 gauge sites, there are 10 measurements of the local curvature. To create a complete *k-(s)* curve, a continuous function is needed. So cubic spline functions are used; these functions have the general form:

$$k_{xy} = a_{xy} + b_{xy}s + c_{xy}s^2 + d_{xy}s^3$$

Each spline defines the curve between 2 adjacent measured points. At the measured points where adjacent splines come together they are smooth through the second derivative. So once the value of curvature for the gauge sites are known, the 10 different splines are calculated. This gives a continuous, approximate description of the real curvature versus length function. Having this, the geometry for the curve can be recreated by the reverse process. Obviously, there is always some element of experimental error in this process, either in the calibration or some other aspect, that would affect accurate generation of the measured closed curve. However, these can be minimised by using several known facts. One is that as the band is a closed curve, the integral of that *k(s)* curve has to be 2 pi. If it's not 2 pi, a small correction factor, *e*, is used as in the following equations:

If $\phi(s) = \int_0^L k(s)ds \neq 2\pi$, then find "*e*" such that

$$\phi(s) = \int_0^L (k(s) + e)ds = 2\pi$$

This guarantees that the beginning of the curve and the end of the curve are pointing exactly in the same direction.

Another problem is that after completing the band, the 2 points would actually not join. To alleviate this error an additional correction factor process has been developed. This process adds two additional terms to the curvature formula, thus making it:

$$(k(s)+e)+k_x\cos(2\pi s/L)+k_y\sin(2\pi s/L)$$

and by an iterative process, which has been implemented on a small PC program that determines the k_x and k_y, generates a closed curve. Once this is done, the shape can be developed in cartesian coordinates.

Figure 21 shows the results of a static test in which the band was placed around a deformable piece of foam and compressed. Curvature readings were taken and the computed shape was reconstructed. Then through a photographic analysis, the actual shape was determined and the results show how well the band duplicates the actual curvature of the process.

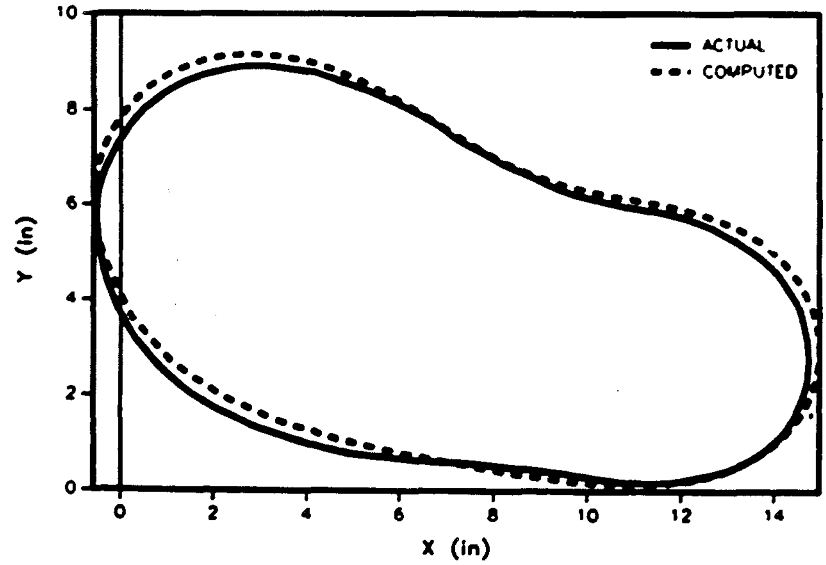
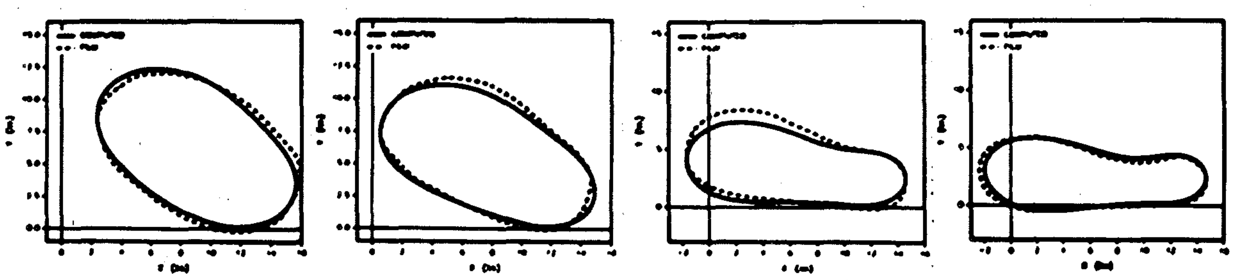


Figure 21: Comparison of actual and computed geometrics
Static test

Some of the early dynamic tests are shown in Figure 22. It is clear that the computer program fairly well duplicates the actual geometry that is determined from the photographic process. Figure 23 shows actual deformations determined from cadaveric specimens being either tested with a seat belt or with an airbag. Thus the entire deformation process can actually be recreated in time at that cross section at the position of the band.



(i) 0 ms (ii) 28 ms (iii) 56 ms (iv) 70 ms

Figure 22: Comparison of computed and film geometrics
Dynamic test
Solid lines = computed Dotted lines = film

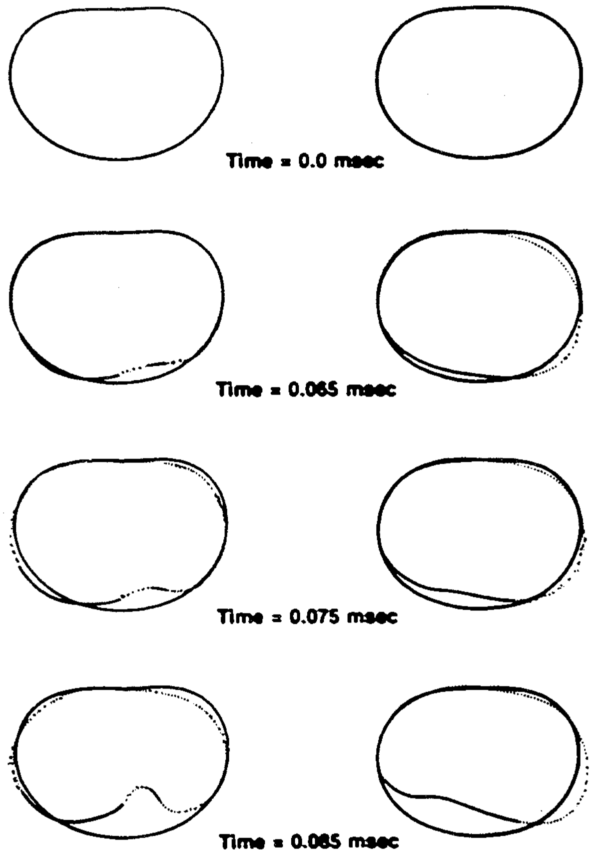


Figure 23: Chest band shapes in 3-point belt and airbag

Now biomechanics and various aspects of research are not all in neat packages. One thing that people often take for granted is the AIS and its ranking of injury severity. I would like to end this talk by presenting some comments about a study carried out some time ago, which examined the biases that exist within the AIS. The AIS is a catalogue of injuries, indexed by body area and ranked by something the authors called injury severity. The definition of injury severity has changed somewhat over the years. For 1976 and before, there were 6 levels of injury severity as follows:

6 — maximum (currently untreatable)
5 — critical (survival uncertain)
4 — serious (life threatening — survival probable)
3 — severe (not life threatening)
2 — moderate
1 — minor

In the 1980 and 1985 versions there were still 6 levels, but the adjectives used did not classify them as rigourously as previously, if the earlier system could have been called rigourous. The 6 levels were defined as:

6 — maximum — injury virtually unsurvivable
5 — critical
4 — severe
3 — serious
2 — moderate
1 — minor

A rider to this classification was that it was said to consider threat to life, permanent impairment, injury dissipation, treatment period and incidence. That is a potpourri of things and not very quantifiable. It is probable that most common notions currently see AIS as representing a threat to life measure. A variety of studies have begun to try to analyse and see if that is a true statement and to check that all injuries of the same rank in AIS regardless of body location pose the same threat, whatever that threat is.

Another common process recommended in the AIS manual which attempts to address multiple injuries is the Injury Severity Score (ISS). It is defined as the sum of the squares of the highest AIS scores for 3 different body areas. In other words, for an AIS 5 in the thorax, 4 in the head and 3 in the abdomen the ISS would be $5^2 + 4^2 + 3^2 = 50$. If there were two AIS 5 injuries in the head, the second would be ignored by this rule. This is questionable. Why should only 3 AIS scores from separate body areas be considered in the formulation of the ISS? And why is it the sum of the squares? In the original paper in which the ISS was proposed, there was no justification for using the sum of the squares, it just seemed to pop up out of the paper and there it was.

Two alternative approaches to be described here were made separately, using data from the NASS files for all occupants who had multiple injuries. The cases were categorised by either 2 or 3 highest AISs, regardless of where they occurred in the body area, and percentage deaths were recorded as shown for example in Table 3. It was clear that the risk of death decreased with the AIS categories.

Table 3 : Alternative approaches to AIS as an indicator of risk of death

Eppinger (uses 2 AIS values)			**Stalnaker** (uses 3 AIS values)		
AIS1, AIS2		Risk of death	AIS1, AIS2, AIS3		Risk of death
5,5	=	86%	5,5,5	=	86%
5,4	=	58%	5,5,4	=	75%
5,3	=	48%	5,5,3	=	50%
	•			•	
	•			•	
	•			•	
3,3	=	2.1%	3,3,3	=	11.6%
3,2	=	0.6%	3,3,2	=	4.8%
	•			•	
	•			•	

The results showed that the categories are very monotonic in the sense that all 5s were always higher than all 4s, and all 4s were always higher than all 3s. The second injuries tended to modulate the probability for fatality significantly. Thus a 5,0 or a 5,1 had about a 20% chance of dying and a 5,5 has 86%, suggesting that if the major injury was an AIS 5 and the second injury could be changed from a 5 down to a 1, the fatality rate could be changed from 86% down to about 20%. This suggests that the AIS rating is not absolute but relative and has, when considered in isolation, only a remote relationship with the overall threat-to-life.

An extension of the above studies examined whether all injuries rated AIS 5 posed the same risk, whether they were to the head, thorax or other body part. Three different levels, AIS 5, AIS 4, and AIS 3, were considered, for each of the three different body areas, the head, chest and abdomen, so there were 9 different categories altogether, From NASS 1982 through 1985 data, a matrix was constructed, using AIS levels of the primary and secondary injuries (H5 is an AIS 5 injury to the head, C5 to the chest and A5 to the abdomen, for example), to show survival rates and numbers of cases. A survival rate model based on this work is shown in Figure 24. This shows how a survival rate of 70% from a single injury is reduced, for example, to 56% by a second injury with 80% survival. This is not claimed as an absolute function that works perfectly, but it illustrates the assumptions that were used in the analysis.

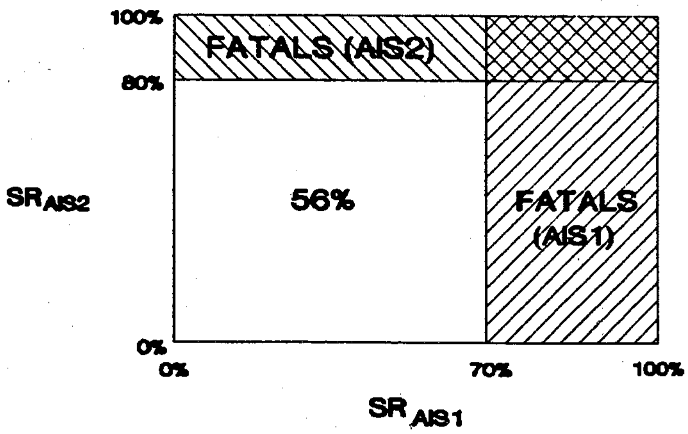


Figure 24: Survival rate model

The solution process is basically to assume that there are 9 unique survival rates for each of these injury conditions. The survival rates associated with each condition are calculated on the diagonal matrix, then a squared error calculated at each of the boxes; the errors are then weighted and summed and the process maximised to find the best survival rates. These procedures are summarised below:

SOLUTION PROCEDURES

1. Assume 9 survival rates for each BR/AIS
2. Calculate 45 $SR_{1,2} = SR_1 * SR_2$
3. Calculate 45 squared error
   $E^2 = (SR_{1,2} - SR_{ACT})^2$
4. Weight and sum
   $$WE_{SQ} = \sum_{i=1}^{45} E_1^2 * W_1$$
5. Find best SR's to minimise WE_{SQ}

SOLUTION

H5 = 0.65	C5 = 0.36	A5 = 0.69
H4 = 0.74	C4 = 0.45	A4 = 0.88
H3 = 0.84	C3 = 0.99	A3 = 0.98

As can be seen, the solution was quite interesting. It can be seen for a head 5, chest 5 and abdominal 5, to match the data in the best least squared sense, there had to be different survival rates for each one of the AIS values. In other words, for an AIS 5 injury, in the case of the head the survival rate was 65% whereas for the thorax area it was only 36% and for the abdominal area it was close to 70% Also, as the level of AIS decreases in a particular body area, the rate of risk changes. Again there is no uniformity between the body areas, nor does the rate of change from 3 to 4 to 5 come out to be the same.

So, using these analyses as an indicator, it is probably necessary to review and question a variety of assumptions used to establish injury and benefits and that there is a tendency to take certain things and keep them as gospel and say 'well AIS is associated with fatality rate and a 5 is a 5 is a 5' — but there are indications that it might not be. This is an area in which there should be further research.

The conclusions from this study were that:

- AIS really lacks quantitative definitions for injury severity and it really prevents a retrospective analysis of how well those definitions worked. In other words if it is really associated with risk to life, it should be analysed as it is used;

- AIS does rank threat to life in major body regions. However threat to life is not equal to the same AIS in different body areas; and

- the simple multiplicative survival rate model works well, but there are other influences in that process. It is not a clear analysis.

QUESTIONS/COMMENTS on this paper follow the next talk

DEFORMATION OF HUMAN THORAX UNDER BELT LOADING

Dominique Cesari

This talk will address the specific subject of deformation of the human thorax, including comparison with Hybrid III, under thoracic belt loading. There is great interest in improving protection in frontal impacts which cause the largest number of cases of accident victims, and even if the belt is probably the most efficient device at this time, there is a limitation to the protection it can give, so if the limits are to be increased it is necessary to understand how the injuries occur in belt loading.

The work presented here is still going on, so the results are only preliminary. In continuation of work already done in Canada with volunteers, INRETS has a program of biomechanical research using cadaver tests in which the thorax is loaded with a dynamically operated shoulder belt: the belt is pulled down with an impactor, dynamically, and this loads the human thorax. As much as possible of the deformation of the thorax is recorded as well as the belt load, so the belt load is measured with a belt force sensor.

The deformation is measured in different directions by different devices. First, there are 10 displacement transducers, of which 9 are vertical and one is horizontal at one side; these are attached to different parts of the body, over the ribs, and 3 are on the belt, as shown in Figures 1 and 2. All of these transducers measure vertical or horizontal deformation versus time. There are also 2 chest bands which measure the circumference deformation of the thorax. These were borrowed from NHTSA and the theory of their function was explained by Rolf Eppinger in the previous talk.

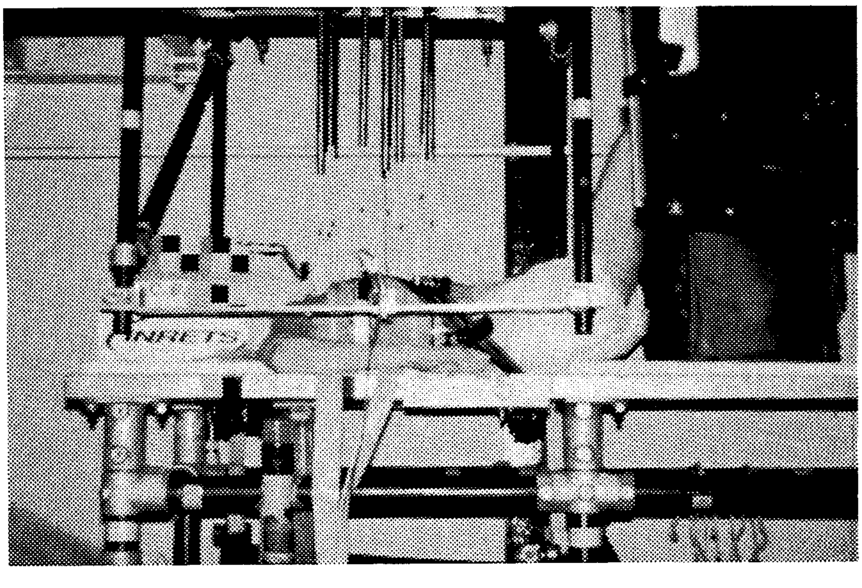


Figure 1: Lateral view of experimental set up for thoracic deformation tests

The same set up is used with the Hybrid III dummy (Fig. 3) and up to now 25 cadaver tests and 30 dummy tests have been conducted. The chest bands were used in some of the Hybrid III tests and not in others because the chest bands were obtained after the program was started. Some preliminary results of 12 cadaver tests and 20 dummy tests will be presented here.

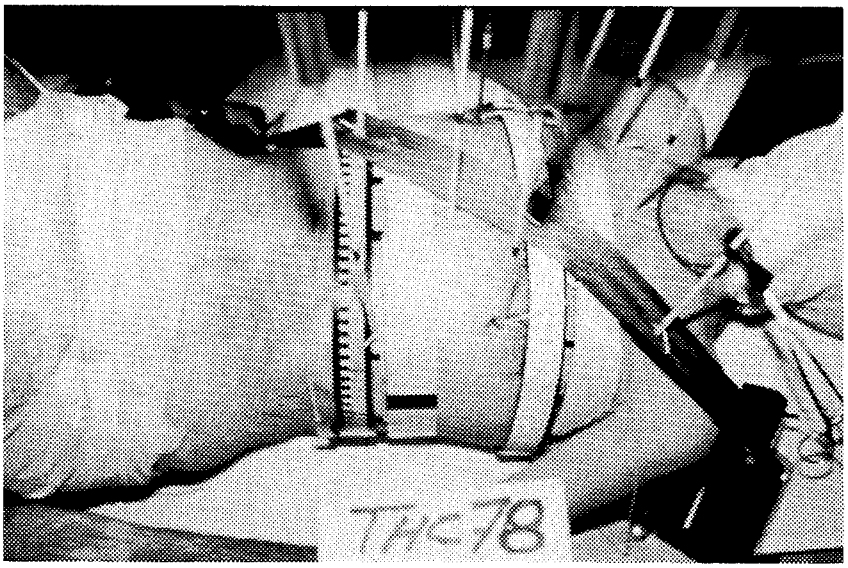


Figure 2: Vertical view of experimental set up for thoracic deformation tests

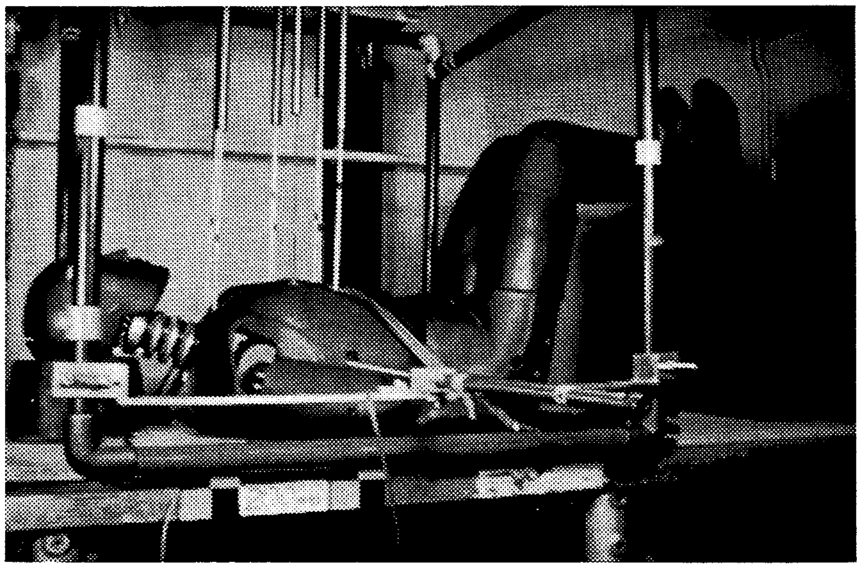
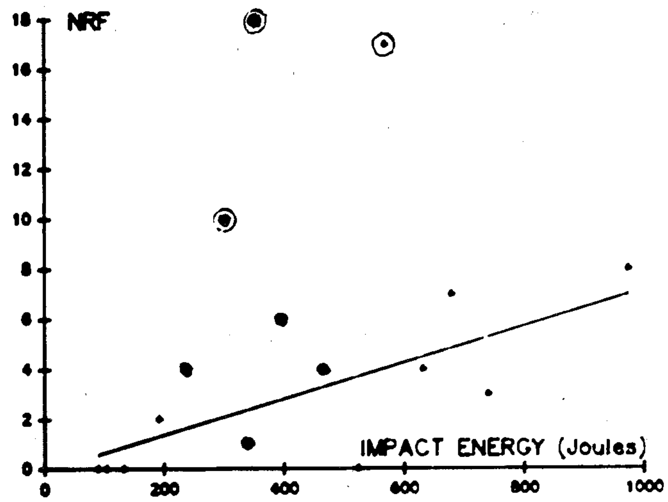


Figure 3: Hybrid III dummy in experimental position

From the data recorded, the correlation of several parameters was looked at; especially for the cadaver, the number of rib fractures which is an indicator of the severity of the impact. Figure 4 shows the correlation between number of rib fractures and impact energy. Energy, not impact speed was used because the tests were made in two conditions, one with a low mass and higher speed impactor, of mass about 20 kg, and another with higher mass, about 76 kg and lower speed. So the common indicator is the energy and not the speed. The results show that there is some correlation between the number of rib fractures and the energy of the impact in which the belt is pulled. For about 6 rib fractures, which can be considered as more or less a moderately severe value, the corresponding impact energy should be about 800-950 J.

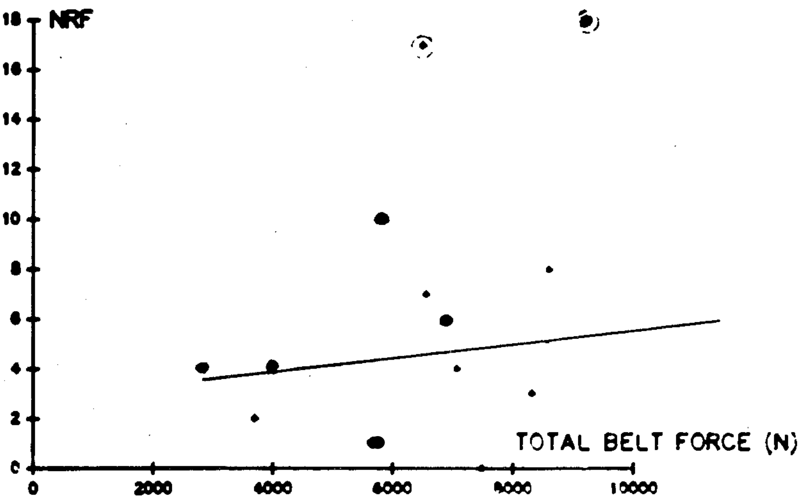


+ = Low mass impact ● = High mass impact

Y = 0.007X - 0.081 Coef. of correl. = 0.719

Figure 4: Number of Rib Fractures (NRF) in relation to Impact Energy

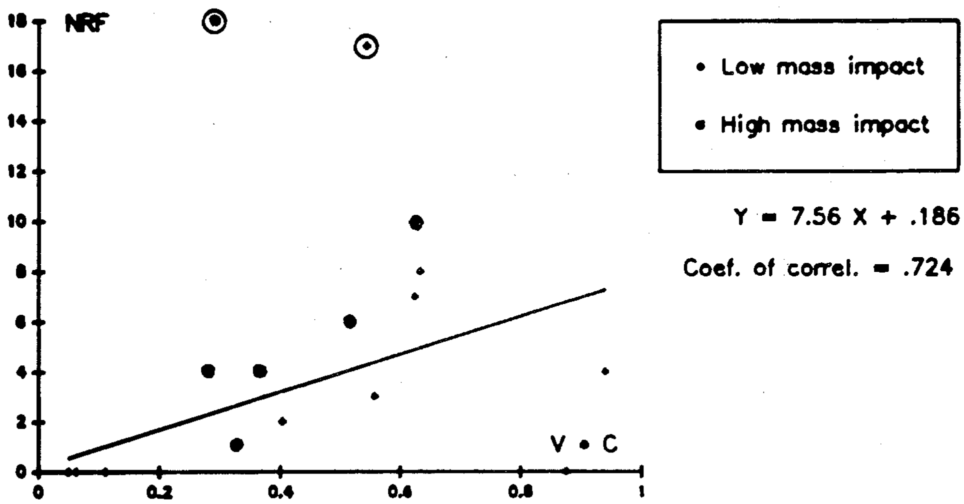
The correlation with belt force was worse than with energy. Nevertheless there is some increase in the belt force with the number of rib fractures, but the sensitivity to that parameter is not very good (Fig 5). For that test, the V*C value is computed as the deformation and the deformation versus time are known and therefore the speed of deformation and the V*C value can be calculated. As far as it has been tested, the V*C correlates quite well with the number of rib fractures (Fig. 6). In fact, when the tests made with two different masses were examined in more detail, the V*C depended on the speed of impact and not the response of the cadaver.



+ = low mass impact ● = high mass impact

Figure 5: Number of Rib Fractures (NRF) in relation to the Total Belt Force

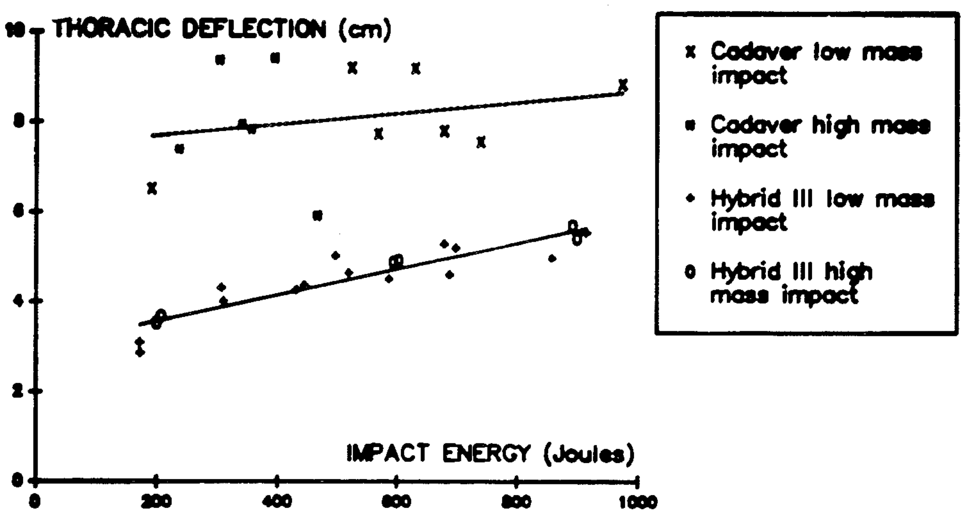
When the deformations of the cadaver and Hybrid III tests were compared, quite a large difference was found. The external deformation was measured in exactly the same way in the cadaver and dummy, from the vertical transducers, and the transducer on the belt in the middle of the sternum was the one which showed the maximum deformation. The results in Figure 7 show that there is a difference of about 4 cm between the dummy and the cadaver. The test results have been grouped into 2 categories: those with the number of rib fractures equal to or below 4, which give a deformation of about 7.5 cm and are those below the line for the cadaver in Figure 6; and those with more than 5 rib fractures, which give more than 8 cm deformation. The corresponding values for the Hybrid III external deformation are between 4 and 4.5 cm. Of course, the Hybrid III records the deformation internally, as there is an internal transducer, and if that difference is added, there is a difference of around 1.5 and 1.8 cm between the internal and external dummy. In fact, the deformation recorded on the Hybrid III is only 3.5 cm for around 8 cm on the cadaver.



+ = Low mass impact   ● = High mass impact

Y = 7.56 X + 0.186   Coef. of correl. = 0.724

Figure 6: Number of Rib Fractures (NRF) in relation to V*C



Cadaver: Y = 0.00123 X + 7.452   Coefficient of correlation = 0.249
Hybrid III: Y = 0.0029 X + 2.991   Coefficient of correlation = 0.919

Figure 7: Comparison of thoracic deflection in cadaver and Hybrid III dummy

So the chest band is used to record the deformation of the thorax and this can be viewed in cross section as in Figure 8 which shows a cross section of the top of the sternum, from the upper chest on the cadaver. The results are for 3 time periods after the beginning of the test. Deformation has not occurred at 16 ms, but it is obvious at 41 ms and at 77 ms the deformation is at a maximum. So for the different points the deformation can be computed for each cross section on which the chest band is placed — high on the sternum for the cadaver, and the same for the dummy (Fig. 9) where there is less deformation. This difference is greater in the results for the second chest band low on the thorax: Figures 10 and 11 show that for the cadaver, there is really a crush by the belt inside the thorax, whereas the deformation of the dummy is more at the side, because the belt crosses the thorax more at the right side which is where deformation occurs.

But this difference in deformation is not necessarily as high when the living human is considered. Because of the lack of muscle tone of the cadaver, the local deformation is probably increased and the belt causes more local deformation than on the living human.

To finish this talk, a short video was presented to show the comparison between the cadaver and the dummy. In the cadaver test the belt caused considerable deformation of the thorax whereas in the Hybrid III test the deformation was less and more regular.

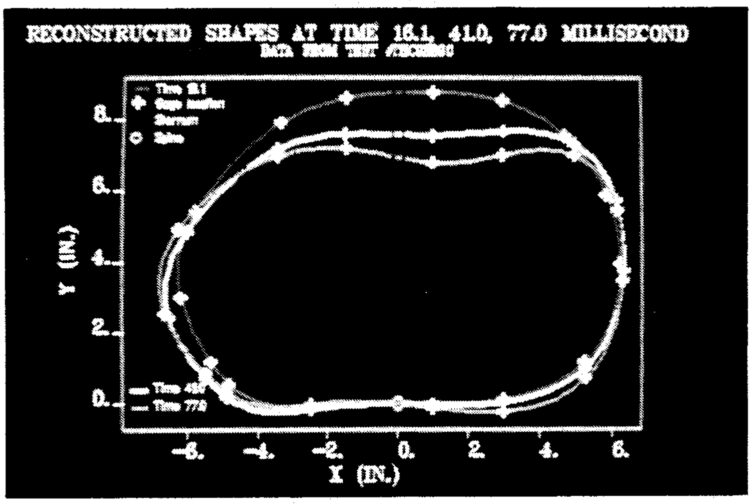


Figure 8: Deformation of upper thorax of cadaver

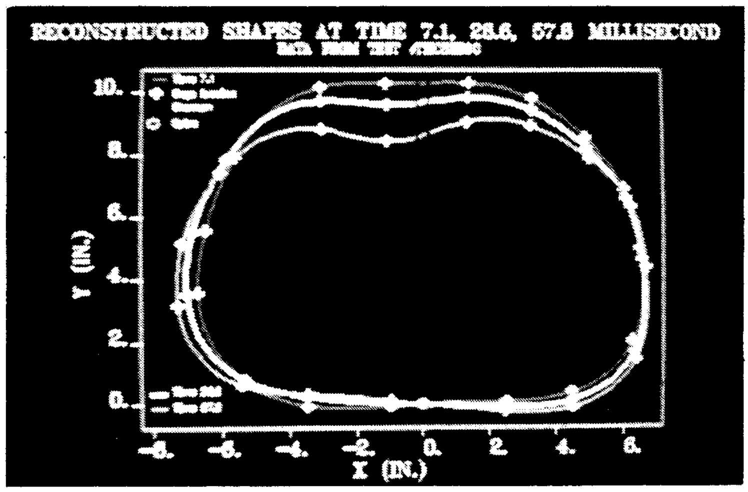


Figure 9: Deformation of upper thorax of Hybrid III

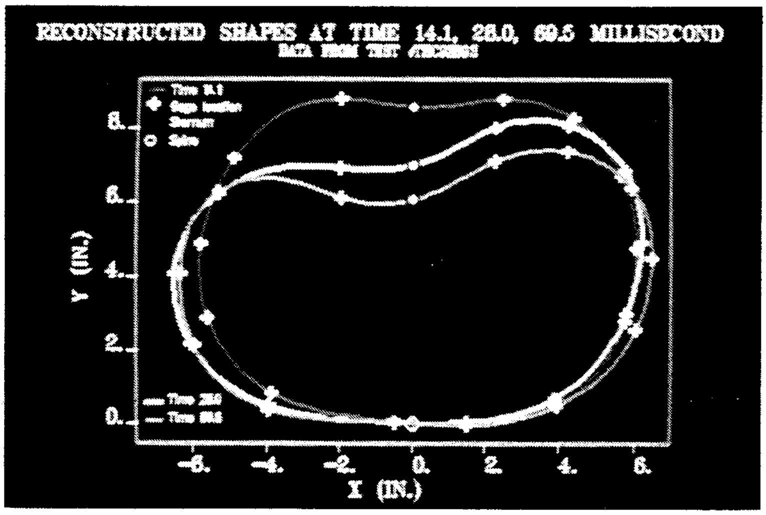


Figure 10: Deformation of lower thorax of cadaver

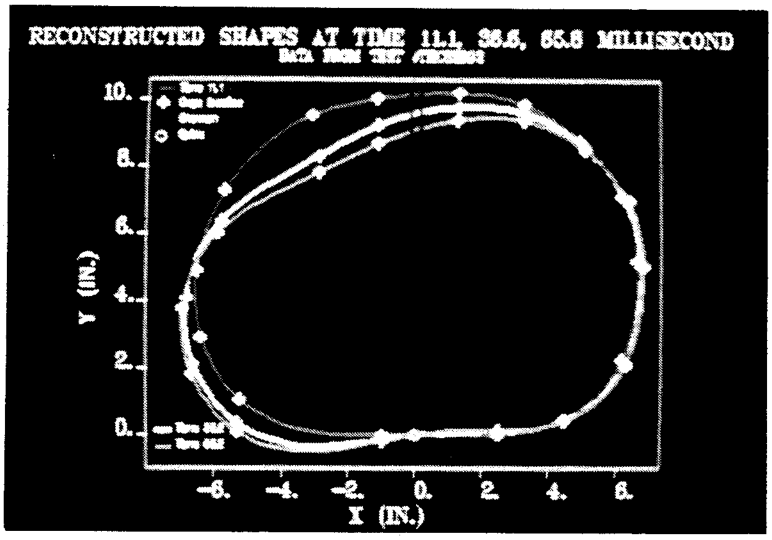


Figure 11: Deformation of lower thorax of Hybrid III

QUESTIONS/COMMENTS to Dr Eppinger and Dr Cesari

Richard Stolinski, to Rolf Eppinger: You had a slide of a batch of production cars showing their TTI performance. Has there been any opportunity to correlate that performance with what NASS injury outcomes might show? Also, with the range of cars that had good to bad performance, were there any discernible differences in their design parameters — for example, did the good cars have any padding that might be like the optimum padding that you had on one of the curves in your presentation?

Rolf Eppinger: Relative to the first question, at this point I don't think we've been able to do that because of insufficient depth of representation of each of those vehicles in the files: the vehicles are from all different accident modes by make and model and there just is not the depth of information for correlation with experimental results.

To your second point: our structural engineers did look at the vehicles and obviously there was a variety of design approaches to a particular body area and a variety of failure modes going on at that point, in terms of A and B pillars shearing off, for example. So there were different types of structural performances going on in these vehicles. And in fact some manufacturers, on receiving the information, that they had a poorly performing structure actually modified production vehicles and significantly altered the performance and brought down the TTI. So there are ways and methods to go about doing that.

Richard Stolinski: Did the cars have, for example, a range of padding materials in the doors, that might contribute?

Rolf Eppinger: My recollection is that they were all baseline cars that didn't have any padding and I believe even the subsequent modifications, although they didn't come down to NHTSA's regulation level TTI, did most of that work through structural modifications and without padding at the time.

Michael Henderson, to Rolf Eppinger: Further to that, I noticed that in that graph, some of the cars had more than one top to the histogram. I take it that some of the cars were impacted more than once and that reflected the range of results? Were those cars modified, between each of the results?

Rolf Eppinger: They were completely different cars.

Bryan Knowles, to Rolf Eppinger: In the results that you've got, did you notice any differences in wheel base of the vehicles in side impact tests?

Rolf Eppinger: I don't know if we did a detailed analysis by each of the vehicle parameters that could define a vehicle by wheel base and weight and so forth. We just basically tried to take a spectrum of what we thought were the vehicles available on our markets — compact and sub-compact and so forth — and tried to understand if there was a variety of performance and what was the best and worst performance available. Sometimes large cars performed less well, sometimes small cars. I don't recall exactly what the distribution was, but it was not generally associated with typical types of vehicle descriptors.

Tony Ryan, comment: I'd like to make a comment about your analysis of AIS, the Abbreviated Injury Scale. I started working with injury scales in 1963 and in those days there was no AIS. The way I proceeded was to look at the range of injuries and decide whether one injury was worse than another. I knew that there was a rough rule that you were only allowed to have one injury per body area. That worked really quite well. It was based on work that had been done at ACIR at Cornell. It seems to me that the AIS has now been pushed way beyond the capacity of this simple concept, that one injury is worse than another, and that there's a confusion of biomechanical concepts and clinical concepts and factors like social cost and impairment and disability and a whole lot of other things which have been put in together, with a need to catalogue injuries in a simple kind of way. So I'm not surprised if you find grave inconsistencies in the results when you try to use the AIS in a quantitative way. If it was originally intended the way I saw and still feel that these scales work, as a rough estimate of some measure of the severity of injury, then that's all you can say. The AIS can't be refined closer than that without getting all these inconsistencies. I do believe that in your fatality file there would be enough information so that you could actually calculate a threat to life for a particular injury, which we didn't have the luxury of in those days.

Bryan Fildes, to Rolf Eppinger: A question about the AIS: if I understand correctly, you did your survival analysis on AIS 85. Is that correct? Since then we've had AIS 90, which supposedly has major differences, particularly in the head/body region. I wonder whether you've thought of repeating the analysis on this latest version of AIS?

Rolf Eppinger: I really haven't considered that. I somehow have this mental picture about how the AIS keeps evolving at this point, by a group of people discussing the rating for a particular injury and when everyone agrees, that's the number that it gets. That's a consensus process and it's a very valuable process and it has developed this scale, which is a very useful scale. I don't want to imply to anybody that AIS is a useless scale and has served its purpose. But if people get the notion that this now represents fatality rate or risk, in an absolute sense, and that it's equal across body regions and so on, I think that's not a correct notion to keep perpetuating. The reason why I did that analysis was just to bring to the attention of people that those commonly held beliefs seem to have evolved by themselves. It should be at least noticed that that's not an absolute truth. I agree with what Tony Ryan said in that sense, that as we amass the accident data files that we have and the descriptions of the various injuries, we should attempt to let the data tell us what the risk associated with each of these injuries is. It becomes a sort of deterministic process — in other words, here's the data, the data suggests that these are what the risks are — rather than going ahead in time and saying 'well the risk is a 5, does it really work out that way?' But when these scales were made, the data didn't exist, so this was the best way to do it and it was a very successful process, I believe.

Michael Henderson, comment: As the antipodean member of the AIS Injury Scaling Committee in the 1970s, I would comment that the process, by correspondence, was exactly what you stated. And it hasn't changed much!

Jack McLean, comment: On the same topic, one of my favourite questions, when I'm in a sneaky mood, to people new to the field, is to hand them an AIS tome and ask 'why do you think this is called the *Abbreviated Injury Scale*?' And in fact, if you go back to the 1969 Stapp Conference, when John States presented the first AIS paper, in the same paper is an appendix with the Comprehensive Research Injury Scale. It was thought at that time that AIS would be used for casual activities and if you were really serious you'd use the Comprehensive Research Injury Scale which covered all of, I think, 4 pages. The current AIS is infinitely more complex than that. And now, last year, Felix Walz from Zurich and Klaus Langwieder from Munich have proposed an abbreviated Abbreviated Injury Scale, because they recognise the current scale is far too complex for use by relatively untrained people in, say, hospital emergency rooms. Also, with ISS there is a concept of squaring these magic numbers that go from one to 6: why shouldn't the numbers go from one to 100 — say one, 10, 50 and so on — and obviously there would be dramatically different answers for an ISS calculated in that way. I sometimes think that what we've seen here over the last 20 years has been the rapid establishment of dogma in what is still very much an evolving field. So I'd like to congratulate you on what you said about AIS. Clearly it has its uses for some purposes but there is a tendency for it to replace thinking to a greater extent than might be desirable.

Michael Henderson, comment: I'd just like to point out that those of you who use the AIS in your laboratories and work places, might usefully take time out to go to one of the injury scaling courses. I think you might get a nasty shock about how the numbers that you're using in your data analyses have in fact been generated. You can get a sense of the problems by actually talking to the people who have been writing the numbers down on the forms.

Ken Digges, comment: In the last year or so, Ted Miller of the Urban Institute has published an analysis of AIS in which he, rather than looking at threat to life, went to the accident data and put a cost on each maximum AIS level associated with head, chest, brain and the rest of the body. To some degree this allowed the differences to work out, and you got a much higher monetary cost for certain injuries than for other injuries. It didn't deal with threat to life but to some degree it made the adjustment on what people are paying for different kinds of injuries. The real challenge that still lies here, though, is the disaggregation of multiple injuries to the same person. We're still using MAIS (maximum AIS regardless of body region injured) and single AIS as a descriptor and indicator of injury severity when, indeed, multiple AISs and multiple injuries somehow need to be taken into account. I see that Rolf Eppinger did it with a series of assumptions. That particular area is one that needs to be addressed very badly.

Michael Henderson, to Dominique Cesari: Would you care to comment on the AIS?

Dominique Cesari: I can say a few words on this topic — at least that apparently the AIS is used in more cases than it should be. Especially, as we are interested in biomechanical tests, we see many publications in which they use AIS values in reference to cadaver tests. In fact, the AIS is a threat to life and a large number of injuries cannot be found on cadavers, and probably there are some mistakes made in this way!

Keith Seyer, to both speakers: On the matter of side impact, there has been a lot of discussion in recent years about the USA regulation and the European regulation. I was wondering if the gentlemen could comment on the differences between the two and whether they both drive the designer to the same solution?

Rolf Eppinger: I'm not completely familiar with all of the distinct differences between the European proposed procedure and our actual implemented procedure. But I believe there's a barrier weight difference, and differences in terms of the barrier stiffness and geometry and also in the test conditions where the Europeans run a strictly 90° test, non-crabbed, at close to the same velocity as we use — the metric equivalent of 30 mph. We have a dummy in the front seat and the back seat. We go through a probably very unique seating process which would be different if a different dummy was used, and we implement the TTI. We tried to assess the effects of the various test dummies and in developing that experimental technique we knew that if we had an extremely soft system of 3 inches of padding, it was in effect a rigid wall — in other words if it was soft enough, it acted as if it was totally transparent and there was a high impact force, while if there was an infinitely rigid pad 3 inches thick, it was again a rigid wall but 3 inches closer to the vehicle occupant. Really, that is a sort of design problem, if the parameters of the crush are considered or the intrusion velocity is considered the same. So we went through that process to see if these devices and the criteria led us to substantially different solutions. We have a small report on it if you'd like to look at the details, but our general conclusion was that they all tended to bottom out right at about the same padding characteristics or crush strength. So I'm of the opinion that they all drive us to basically a common solution. The potential difference that I see is that the other 2 devices seem to have additional measurements available that the SID doesn't. The SID is a slightly more primitive device — it has 2 acceleration readings on the side of the chest and one on the spine and that derives the entire performance process, whereas both the BIOSID and EUROSID have articulated arms, they have the choice of putting the arm up or down and they have a couple more measurement points and they have different parameters to evaluate. So some things might creep in there to make some difference, but from the overall gross performance I really don't see any difference.

Dominique Cesari: Yes, I can add that I agree in principle that they are similar to each other. I think the differences found in the test conditions are really the prime interpretation of accidents or different situations between Europe and the USA. In Europe we have considered, for example, that it is not necessary to give a speed equivalent to a stationary car by angles just because we think the mechanism producing injuries is mainly perpendicular to the body and if something is added, the situation will not be the most serious. So it's really a prime interpretation, whereas in the USA, they try to be closer to the reality of the situation. In terms of the barrier, the difference of car fleets between the countries can explain the difference in mass and width — it explains why the American barrier is wider. For the stiffness of the barrier, there are probably also different choices made. In Europe we have tried to be as close as possible to the stiffness of the reference car population, not only in overall stiffness but also in the distribution of stiffness between the side and middle, and between upper and lower rows. It seems to me that in the USA it is more the global stiffness of the population of cars than a local distribution of stiffness. With respect to the occupants, the difference is a little more difficult to understand except with the work that we have done in parallel, more or less. And they are also related to differences of interpretation or differences of point of view. For example, in the committees we have in Europe there was multiple questioning against taking acceleration based parameters because in principle acceleration does not provoke any injury. Injuries are related to deformation, force and pressure but not directly to acceleration. It's worth explaining, in fact, that there is some interaction (with acceleration), but that was the point of view and the reason why acceleration-based parameters were avoided as much as possible on the EUROSID dummy proposed for the European regulation under discussion. Then for the pelvis there is a force measurement, for the abdomen there is a crush/force measurement and for the thorax it's a displacement. In fact, the displacement can be transferred, for example to V*C or some other parameter. These are, I think, the main differences. Another is that in Europe we have included head injury parameters, but we think that it is not enough to protect from head injury in one full scale side impact test. There will probably be some proposal to check specifically the question of head protection in component or subsystem tests, then in the side barrier test.

Keith Sayer: Do you know of any comparison test that was done to see if a vehicle that was designed to meet one regulation would meet the other?

Rolf Eppinger: I believe Transport Canada is doing a variety of testing of that nature. Not necessarily cars that have been designed to meet any of these particular requirements, but a car that's been designed and they used the European barrier and the European dummy, and they used our barrier and SID and I even believe there are other permutations going on in that test matrix.

Ken Digges: Yes, I was going to refer you to the Transport Canada tests, which was an extensive matrix of tests which used different dummies and different barriers, all substituted in the same model car. There was also a series of modified cars provided by the Motor Vehicle Manufacturers' Association (MVMA) that were tested with different dummies. If I remember correctly, the biggest difference found was the barrier difference. The USA versus the European barriers might drive you to different designs. The difference was that the European barrier, being softer, tended to test the door more than the A and B pillars whereas the barrier used in the USA, being stiffer, tested the A and B pillars. Now the consequence of that is that the USA barrier will be more forgiving to large cars because it will accelerate the entire large car, whereas the European barrier will be more forgiving to small cars and more difficult for large cars because it will load the door heavily because it won't accelerate the mass — it means the door has got to accelerate the mass, not the A and B pillar. So you see that's the big difference, I think, between the standards. The dummies will drive you to the same solution.

THE PRESENT SITUATION OF PEDESTRIAN ACCIDENTS IN JAPAN

Koshiro Ono

The nature of pedestrian injuries, including those to the lower limb, will be dealt with in this talk.

The United Nations, ISO and other organisations have carried out technical studies on pedestrian accidents in recent years, in order to establish standards on passenger cars for the protection of pedestrians. In this regard, it will be of value to improve pedestrian protection measures in vehicles in an effective manner based on actual conditions of traffic accidents. Needless to say, it is also necessary for proper steps to be taken in Japan.

As the first step, therefore, Japanese national traffic accident data were analysed at JARI in order to study them from a statistical point of view. The data, however, did not include specific details such as vehicle and impact speeds, details of pedestrian injuries and the part of the vehicles that came in contact with the pedestrians.

As the second step, in-depth case studies were conducted and pedestrian accidents in 1987 and 1988 were analysed in order to determine the relationship between the region of injury and contact area per vehicle impact speed.

Lateral impact by the front portion of the bonnet of passenger type cars against pedestrians crossing roads make up the highest proportion of pedestrian accidents according to analysis of the national statistical data. A similar tendency exists in the United States and European countries according to their reports. This talk will present the relationship between the vehicle contact area and the region of injury classified by age and stature of the pedestrians, according to the in-depth case studies and analysis in Japan.

The case studies covered 113 pedestrians in both casualties and fatalities. The total number of injuries of 113 pedestrians was 425. The severity of each injury was judged according to AIS 85, known as the Abbreviated Injury Scale of 1985. Figure 1 shows component ratio region of injury classified by age group. The ordinate represents the component ratio while the abscissa represents age. Leg injuries account for the highest rate of 37% followed by head injuries, arm injuries and so forth. The tendency for the occurrence of injuries to the head and leg regions to predominate is the same amongst different age groups.

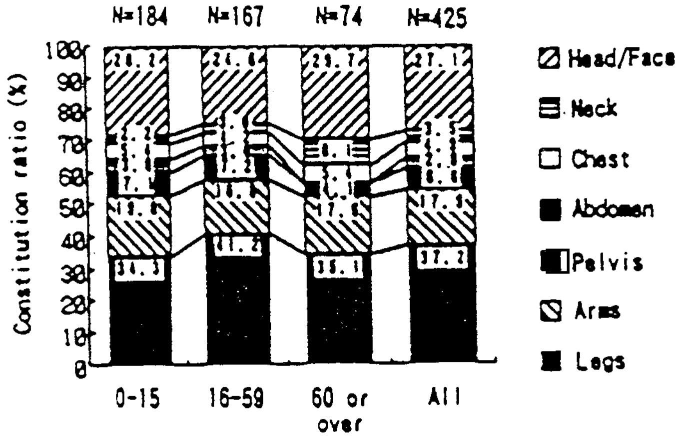


Figure 1: Constitution ratio of injury region by age group

Figure 2 illustrates the injury region component ratio classified by the severity of injury. While there are many leg injuries of AIS 1 or AIS 2 severity levels there are relatively more at AIS 3 and AIS 4 levels; head injuries account for the highest rate in AIS 5 and AIS 6.

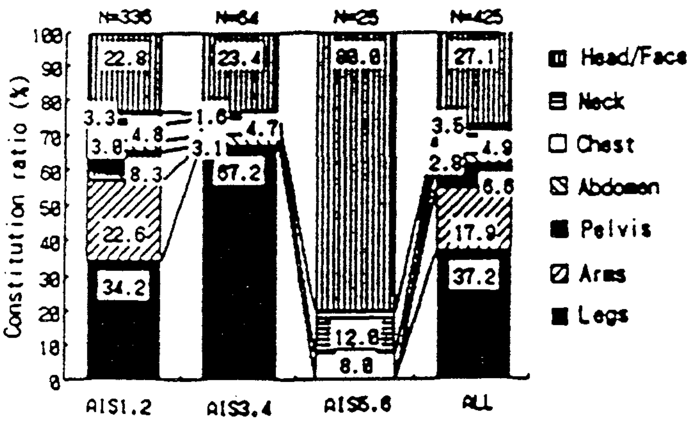


Figure 2: Constitution ratio of injury region by AIS group

The component ratio of injury severity classified by age groups is shown in Figure 3 in which the ordinate represents the component ratio and the abscissa represents the age group. Minor injuries of AIS 1 or 2 account for the highest rate for the age group of 0 to 15 years old, while the severe injuries of AIS 5 or 6 account for the highest rate in the age group of 60 years or older.

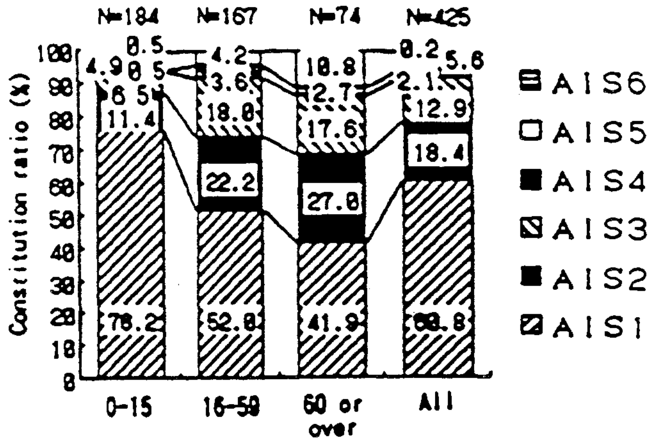


Figure 3: Constitution ratio of AIS by age group

In Figure 4, impact speed by AIS level is compared for the pedestrians of 16 to 59 years old and those of 60 years old or older. Children are excluded here since they are shorter in stature and the area of contact may differ even if the injury region is the same. The solid line represents pedestrians of 60 or more years and the dotted line represents the age group of 16 to 59 years old. The encircled numbers 1,2,3, are the AIS ratings as indicated. It is clear that, if severity of injury is the same, pedestrians who are 60 years or older tend to have injuries at lower impact speeds compared with those who are 16 to 59 years of age.

The data for component ratio of cause of injuries given in Table I show that injuries caused by impacts against vehicles are more than twice the injuries caused by impact against road surfaces: impacts against vehicles account for 63% of total impacts whereas those against road surfaces are 31%. Of the injuries caused by vehicles, those caused by bumpers account for the highest rate, followed by the bonnet top surfaces and bonnet leading edges.

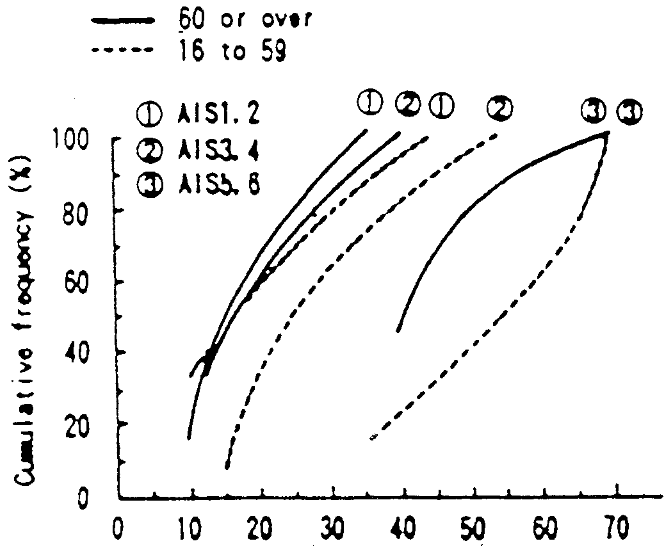


Figure 4: Relationship between vehicle impact speed and AIS by age group

Table 1: Number of pedestrian injuries related to vehicle contact location and body region

Contact location \ Body region		Head	Face	Neck	Chest	Abdomen	Pelvic	Arms	Legs: Over-all	Legs: Femur	Legs: Knee	Legs: Lower leg	Legs: Foot	Legs: Sub-total	Total
Part of the vehicle	Front bumper					2	1		1	18	29	30	1	79	82
	Top surface of bonnet and wing	38	5	1	9			22							75
	Leading edge of bonnet and wing			1	2	7	20	6		18				18	54
	Windscreen glass	9	6	1				1					1	1	18
	Windscreen frame and A pillars	8	3					2							13
	Other	1	3				1	6	1	2	2	3	6	14	24
	Sub-total	56	17	3	11	9	22	37	2	38	31	33	8	112	267
Indirect contact injury				12			2	3		2			4	6	23
Road surface contact		22	20		10	3	4	34	3	3	20	1	10	37	130
Unknown								2				1	2	3	5
Total		78	37	15	21	12	28	76	5	43	51	35	24	158	425

The relationship between the region of injury and contact area of the vehicle is classified in Table 2 into 2 groups: children of 0 to 15 years old and adults of 16 or more years. It can be seen that the vehicle contact areas are different for children and adults, even if the region of injury is the same; also locations of impact are different, even if the contacted area is the same. When the relationship between impact location of contacted area and the injury region was considered, the relationship against stature was studied first, because stature was likely to affect the relationship. Stature was classified as follows: the average height of 730 mm for bonnet leading edges of vehicles covered by the in-depth case study was used as the reference height; persons with pelvis height lower than the reference height were considered as children and the rest were adult (Fig. 5). The pelvis height is assumed as the iliospinal height. Maximum stature of children is assumed to be 131cm and the minimum height of an adult is 150 cm.

Table 2: Number of pedestrian injuries related to vehicle contact region and body injury region, according to age group

15 years or less

Contact location	Body Region	Head	Face	Neck	Chest	Abdo men	Pelvic	Arms	Legs						Total
									Overall	Femur	Knee	Lower leg	Foot	Sub-Total	
Part of the vehicle	Front bumper					2	1		1	12	13	5	1	32	35
	Top surface of bonnet and wing	20	4	1	4			8							37
	Leading edge of bonnet and wing			1	2	4	9	5		3				3	24
	Windscreen glass														
	Windscreen frame and a pillars														
	Others	1	2				1	3	1	1	2	1	1	6	13
Total		21	6	2	6	6	11	16	2	16	15	6	2	41	109

16 years or more

Contact location	Body Region	Head	Face	Neck	Chest	Abdo men	Pelvic	Arms	Legs						Total
									Overall	Femur	Knee	Lower leg	Foot	Sub-Total	
Part of the vehicle	Front bumper									6	16	25		47	47
	Top surface of bonnet and wing	18	1		5			14							38
	Leading edge of bonnet and wing					3	11	1		15				15	30
	Windscreen glass	9	6	1				1					1	1	18
	Windscreen frame and a pillars	8	3					2							13
	Others		1					3		1		2	5	8	12
Total		35	11	1	5	3	11	21		22	16	27	6	71	158

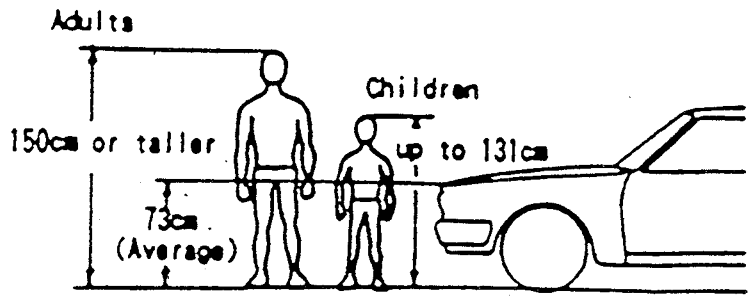


Figure 5: Classification of stature

With regard to head injuries with high incidence rate, for example those caused by impact on the bonnet top surface, the highest rate was found to be for children while those on the area between the bonnet top surface and windscreen showed the highest rate for adults. Impact locations for children were in the two thirds area of the bonnet from the leading edge (Fig. 6). Children accounted for the great majority of head impact against top surfaces of the bonnet.

For chest, abdomen and pelvis injuries, the severity tends to be greater for persons 16 years or older compared with children up to 15 years old. The graphs in Figure 7 show the relationship between speed and AIS of chest, abdomen and pelvis injuries for children and adults. As the impact speed increased, the severity of chest injury tended to become higher for adults. This was likewise true as stature increased and was presumably because the secondary impact speed of the chest against the bonnet became higher.

In the case of the abdomen, the relationship between impact speed and the AIS of injuries showed that abdominal injuries were not found among adults at low impact speeds. This was because the abdomen did not come into direct contact with the bonnet leading edge owing to the relatively high stature of adults. Injuries of AIS 3 or greater severity were found in adults at impact speeds higher than 45 km/h. This is presumably because the legs are apt to be caught by the front bumper on the primary impact and dragged in the vehicle running direction when the impact speed becomes higher. Upon impact, the amount of parallel travel of the pelvis increases, then the pelvis drops downward. This may cause the secondary impact against the bonnet leading edge.

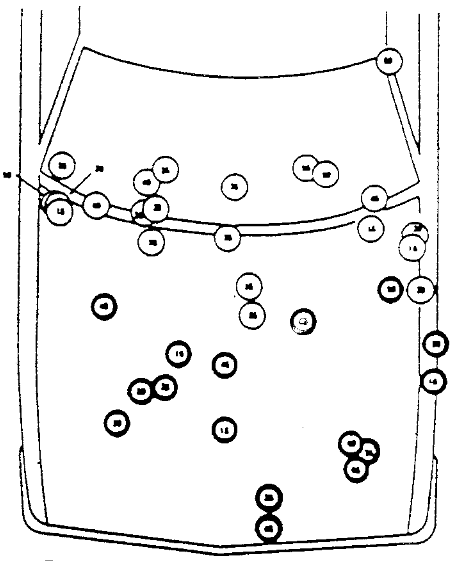


Figure 6: Distribution of head/face contact positions by stature of pedestrians on bonnet and other portion

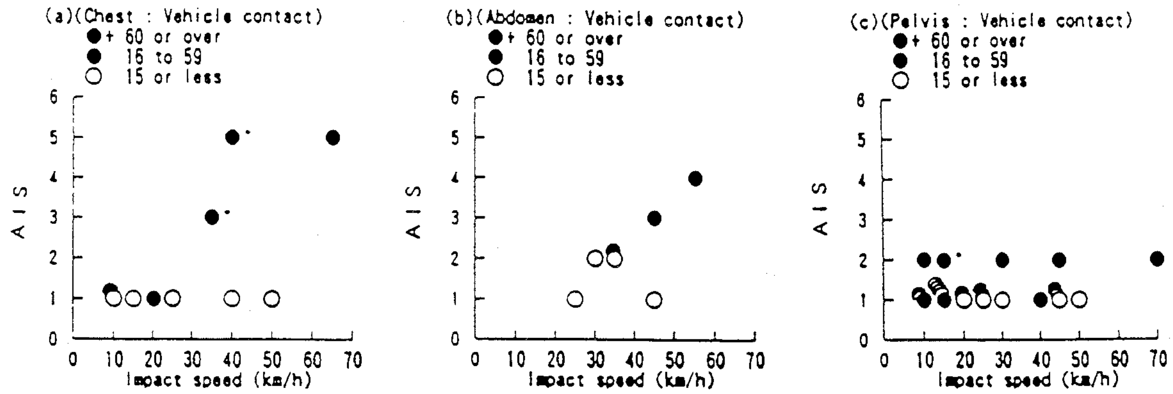


Figure 7: Relationship between vehicle impact speed and injury severity of body region (chest, abdomen and pelvis), imposed by vehicle contact for age group

The relationship between the impact speed and AIS of pelvis injuries classified by age showed that all AIS 2 injuries were those of adults and they were bone fractures in the pelvis region. This was presumably because the adults, due to their higher stature, were hit around the lower region of the pelvis near the hip joint by the bonnet leading edge, which caused fractures of the relatively weak pubis. Thus stature appears to affect the incidence rate, region and severity of injuries in the chest, abdomen and pelvis.

Various factors are important in the case of leg injuries. The relationship between the bumper-lead length and angle is seen in Figure 8. Fractures were investigated for the femur and lower leg bones and injury of ligaments was investigated for the knee region. The femur fractures of children tend to occur when the bumper lead exceeds about 80 mm or so (Fig. 8a). For adults, the front bumper tends to become the area of contact when the bumper lead-angle is 65° or less (Fig. 8b), whereas if the bumper lead is long, the bonnet leading edge tends to become the area of contact when the bumper-lead angle is larger than 65° — that is, if the bumper-lead length becomes shorter.

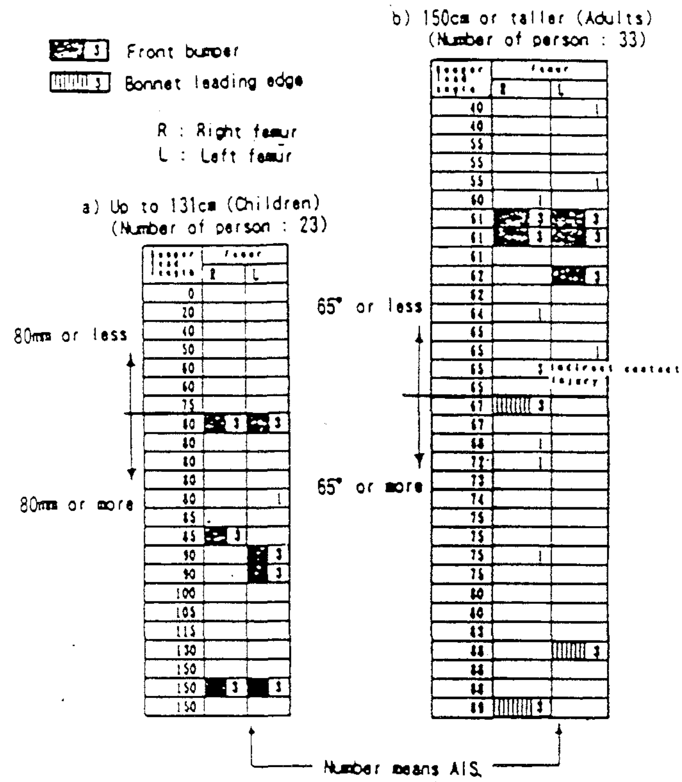


Figure 8: Relationship between fracture of femur and bumper-lead length or bumper-lead angle in pedestrians of heights up to 131 cm and 150 cm or taller

When the bumper lead is long, the impact load tends to concentrate on the impact between the bumper and femur in children, as can be seen in the upper left diagram of Figure 9, causing the femur fracture. In the case of adults, if the bumper-lead length becomes shorter, some bone fracture is likely to occur near the hip joint as the impact load concentrates on the femur and bonnet leading edge as shown in the lower right diagram of Figure 9. In short, the bumper lead appears to affect femur fracture.

Injuries of knee ligaments were not found in children. The bumper-lead length and the angle did not show any relationship for adults either. The bumper height appeared to affect knee ligament injuries since such injuries are found only in adults. As more leg bone fractures are found even when the impact speed is as low as 15 km/h, the posture of the pedestrian, such as whether the pedestrian weight is concentrated on one or both legs at the moment of impact, appears to affect the incidence of leg injuries.

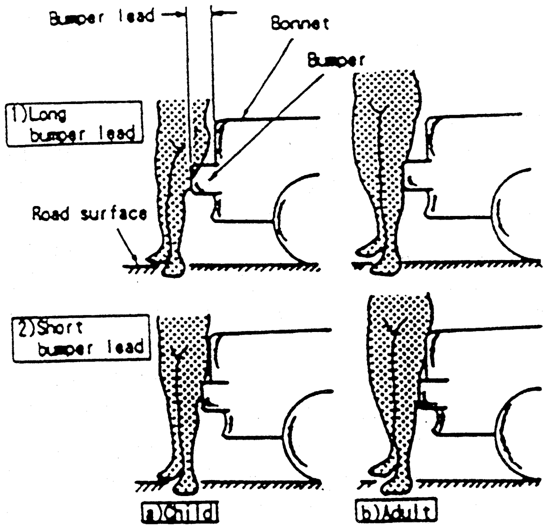


Figure 9: Comparison of impact pattern for children and adults (long bumper lead and short bumper lead)

The basic findings of the in-depth study described so far have been considered in relation to the vehicle test conditions for the protection of pedestrians which the ISO and the other organisations are currently working on.

To summarise this discussion:

- children account for two thirds of impacts against the top surface of the bonnet;

- the injuries to chest, abdomen and pelvis of adults tend to be more severe than those of children, even if the impact speed is the same. For the impact test against bonnet leading edges currently studied by ISO and other organisations, it will be appropriate to set such test conditions that will be effective in the bonnet leading edge impact test assuming the impact is against the femur for adults, especially for those adults who tend to have more severe injuries than children;

- femur fractures of children are found when the bumper lead length exceeds about 80 mm; and

- ligament injuries of knees of adults are found from the low impact speed of 15 km/h or so. In test conditions set by the ISO and other organisations for bumper impact tests, impacts against knees of adults and those against femurs of children may be assumed.

Considering the above facts, it will be necessary to place emphasis on knees of adults when setting such conditions. As the results described in this talk show, it will be vital to take a detailed account of differences in stature between adults and children for the protection of pedestrians against vehicles, including the issue of compatibility or harmonisation with production requirements of vehicles.

QUESTIONS/COMMENTS for this paper follow the next talk

BIOMECHANICS OF PEDESTRIAN LOWER LIMB INJURIES

Dominique Cesari

This talk will mainly describe the work done in Europe for the improvement of pedestrian protection. In presenting the problem of lower limb injuries of pedestrians, this does not mean that these are the only leg injury problems: for example, protection against lower limb injury is an important subject also for car occupants. However all the different accident conditions could not be covered and pedestrian accidents have been selected because they provide the largest number of lower limb injuries and also because it is in the pedestrian accident situation that several research and development studies are being carried out to improve the situation.

Without spending too much time on the accidents, it can be recorded that even if there is a decrease in pedestrian fatalities, they still constitute a large group of people as is shown by the data in Figure 1 for the period up to 1986. The percentage of pedestrian fatalities in traffic accidents is now almost stable. Considered separately, they are the largest group — their number is less than that of car occupants, but if car occupants are divided between accident conditions, pedestrians become a relatively large group, more than motorcyclists for example. In some countries like the United Kingdom, they constitute close to half the fatalities of traffic accident victims.

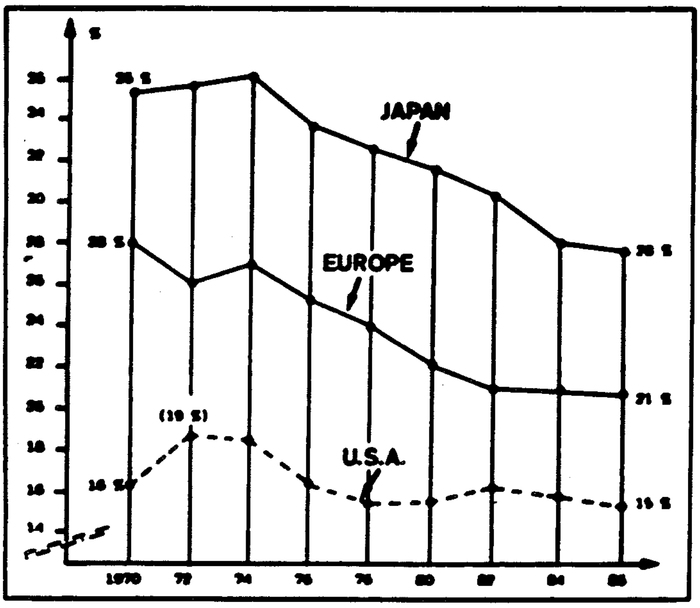


Figure 1: Comparison of pedestrian fatalities in Japan, Europe and the USA between 1970 and 1986

It was demonstrated clearly at Birmingham University some years ago that the vehicle is the first cause of severe injuries to pedestrians. Some injuries are due to contact with the ground but they do not increase, or they barely increase, with impact speed, mainly because they are related to the rate of fall which is almost the same whatever is the impact speed, whereas the severity of injuries due to the car contact are really related to the speed of the vehicle (Fig. 2).

There is also the question of which speed should be considered in terms of protection and this is probably something on which there is not full agreement. Figure 3 shows two distributions established from accident analysis, but it must be remembered that it is very difficult in accident analysis to determine the speed of the vehicle at the time of impact with the pedestrian. Nevertheless, from these results it seems that the average for all severities is below 30 km/h, around 8 m/s, and the average for fatal cases is around 14 m/s, which is not a very high speed, in fact.

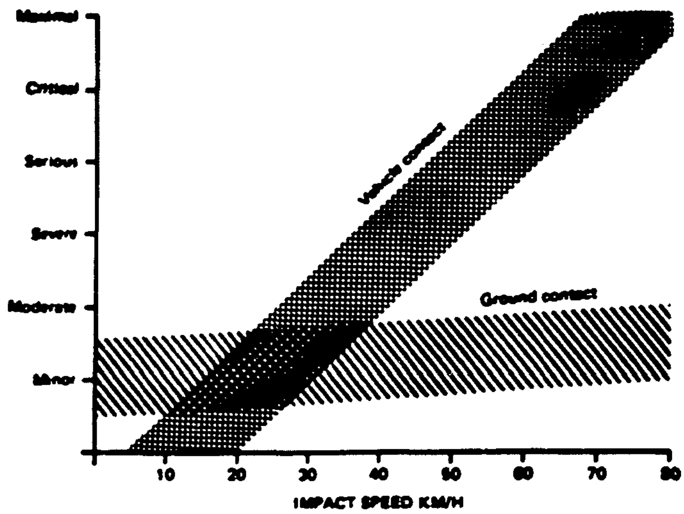


Figure 2: Predominant injury severity by impact speed for vehicle contact and ground contact injuries

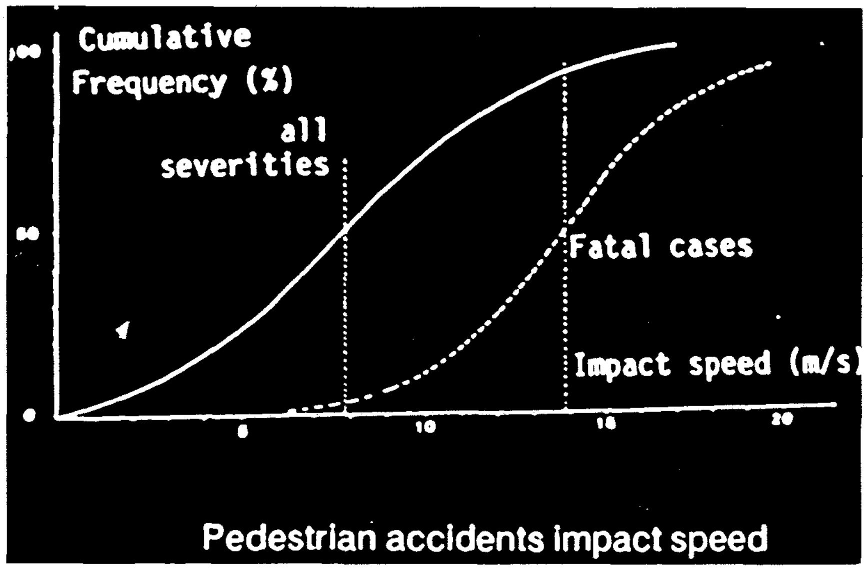


Figure 3: Impact speed in pedestrian accidents

Another thing which has been found in accident analysis, especially at a statistical level, is that there is really a correlation between severity of injury and the shape of the car. There have been several attempts to determine the type of shapes and the effect of the type on severity of injuries and there is, for example, some definition of three types: the box type, the V type and the more common rounded type. It appears that there is an effect on leg injuries because the front car surface involved with the legs is not at the same level for all shapes. Also in terms of the trajectory of the victim, the V shape, which is more common in new cars, tends to increase the speed of the head motion because there is a rotation from a lower point on the body and then the point impacted by the head is higher up on the car. This does not mean necessarily that these accidents are more severe because there is also a large difference in the stiffness of the car component that may be impacted by the head. For example, windscreen impact is in principle not severe whereas some steel parts are much more stiff.

With regard to the problem of leg injuries, it appears that whatever severity level is considered, from less severe to more severe, leg injuries are predominant with head injuries. There is some balance between the 2, but Figure 4 shows that for more severe injuries, more than 90% of the victims sustain leg injuries and for the other severities, between 60 to 80%. These data are based on accident analysis done in Hanover University by Dr. Otte.

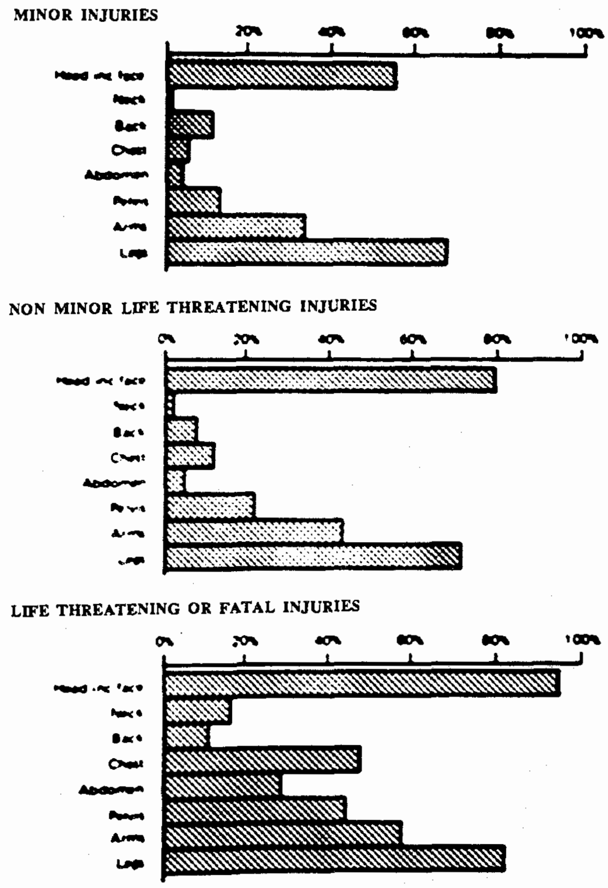


Figure 4: Injury distribution in pedestrian accidents

As speed is increased, the severity of the leg injuries increase as seen in Figure 5 which shows the severity of leg injuries according to age group. Almost 70% of the elderly sustain severe leg injuries compared with only about 30% of children.

The next question is what are the leg injuries and that is a little more complicated. Biomechanical research at INRETS has probably focussed up to now more on ligament injuries. They are not the most common injuries: the data in Table 1 shows that they occur mainly in the speed range of 30 - 50 kph, the main range of interest for protection, but they occur mainly in adults and not in children. It should be noted that these ligament injuries are not the only knee injuries: some knee injuries are just bone fracture, as of the tibial plateau or around the knee, so there is a higher frequency of knee injuries than just ligament injuries. Nevertheless, because of their long term consequences, even if they are not very frequent ligament injuries must be considered very carefully.

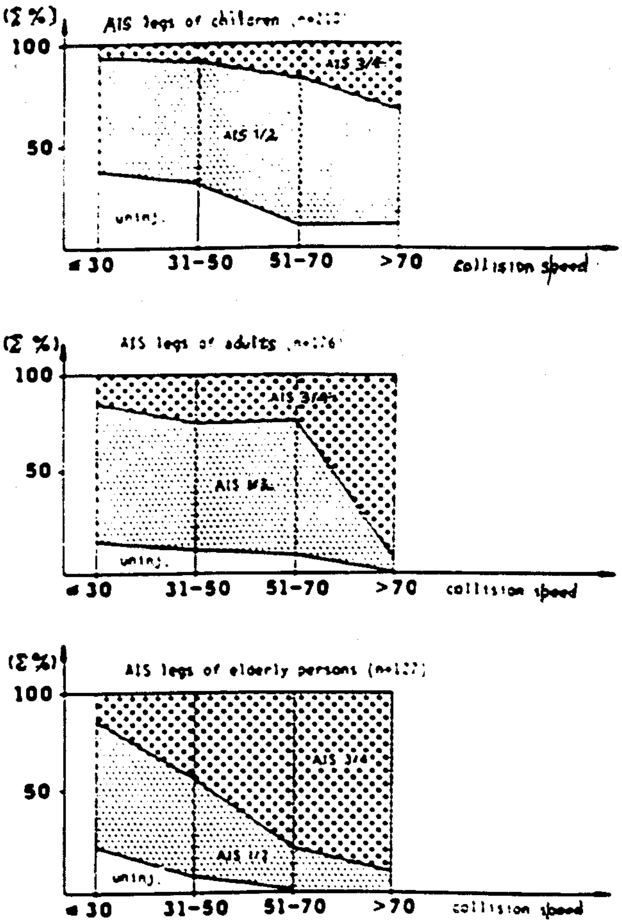


Figure 5: Severity of leg injuries according to age group

Table 1 : Frequency of knee ligament injuries

Collision speed km/h	**Children** %	**Adults** %
≤ 30	0	1.2
31 - 50	0	7.9
51 - 70	2.6	5.0
> 70	0	0

Thus accident analysis gave some information on such matters as frequency distribution of pedestrian accidents. Then some years ago INRETS decided to go further and try to understand this question of pedestrian and leg injuries. A program has been started with different cars, normal cars, impacting cadavers and dummies. This program has proceeded step by step. The first step was a full scale cadaver test in which the deformation of the leg at the time of the impact was observed from high speed films. What was interesting was that when this deformation of the leg occurred around the front of the car, the upper part of the body, which was free and not suspended at that time, did not move - it moved later. So the leg injuries can be considered quite separately from those to other body parts.

These tests with full scale instrumented cars and cadavers have shown various results because of different injury mechanisms. The first one is that the leg on the impacted side, which is impacted first because the pedestrian was hit from the side, always sustains or corresponds to a higher bumper force than the leg which is the second one impacted (Fig. 6).

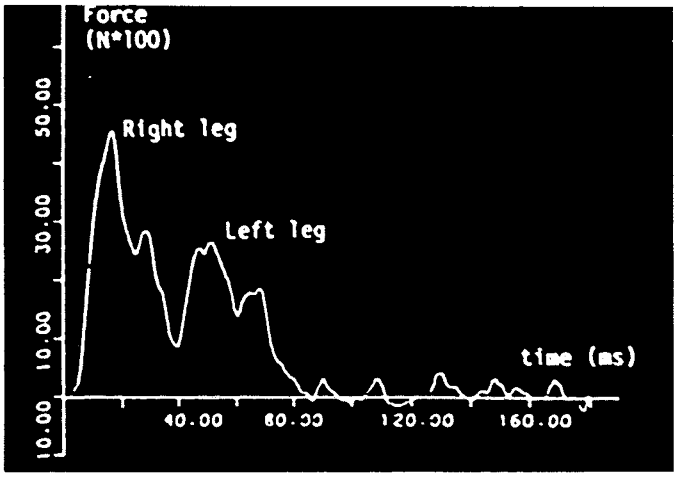


Figure 6: Bumper force in a pedestrian test

Secondly, it has been established that 3 main injury mechanisms interact in terms of injury production. In order of severity rather than of occurrence in time, these are:

- the bending deformation around the knee joint. This means that in certain conditions, the leg bends in such a way that the force applied to the ligaments inside the knee passes tolerance levels and this creates injuries inside or around the knee;

- the shearing force in the knee, which means that in a different process, the upper and lower legs move almost in parallel so that the ligaments of the knee are not all loaded identically. In the first case it is mainly the collateral ligament on the opposite side while in the second it is more the cruciate ligaments which are loaded; and

- a third type of injury which normally occurs far away from the joint, namely the long bone fractures which are mainly due to the contact force of the car component at the place where the fracture occurs.

These mechanisms depend on the surface of the impact as well as the stiffness of the impacting component, so that the first two may occur at a distance from the impact point and the third one at the impact point or close to the impact point.

After these results were established from the full scale tests it was decided to go further. What is called a car platform has been developed. This is a moving platform with adjustable bumper: both the height and stiffness of the bumper can be changed. A series of tests has been conducted with cadavers and with one dummy developed jointly with Chalmers University at Göteborg in Sweden (Fig. 7).

One clear outcome from these tests was that the knee angle, which is the knee deformation angle between the thigh and the lower leg, increased with bumper height. There was a certain optimised bumper level which almost did not produce any bending moment inside the knee and the same kind of relationship was also established on the dummy. But the question was whether, if the bumper was lowered, the risk of knee injury decreased, but the risk of injury to the long bones increased. When bumper force, which was directly correlated with the risk of injuries to the long bones, was considered it was proved also that the bumper force had increased with bumper height (Fig 8). Then lowering the bumper led to a lower impact on the leg, which to a certain amount decreased the risk of injuries to long bones.

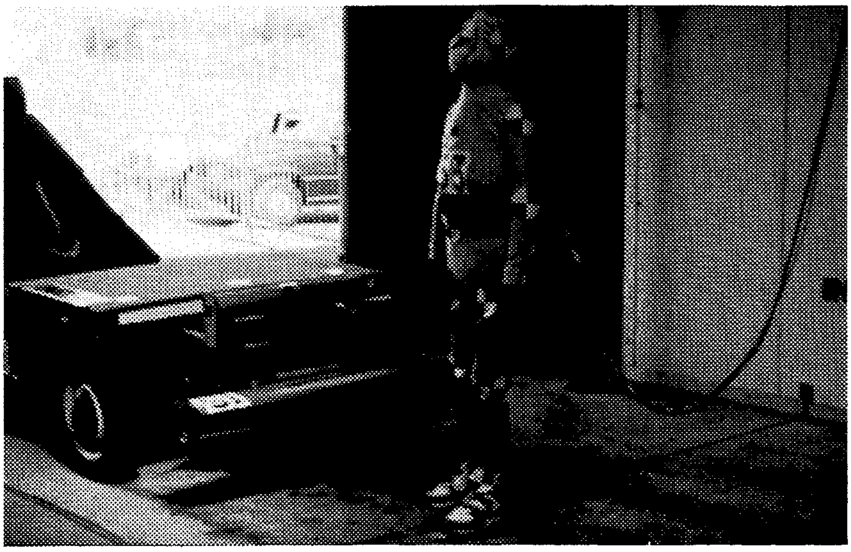


Figure 7: Dummy impact test with adjustable bumper

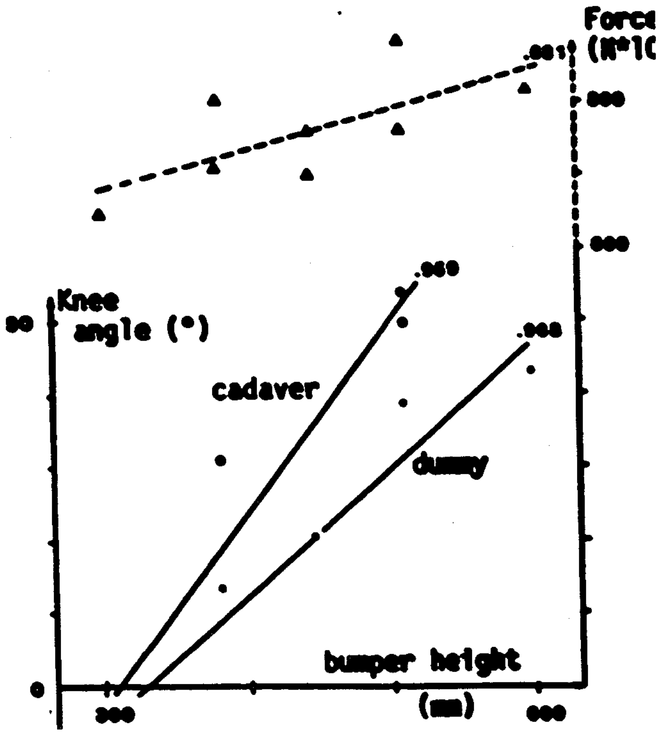


Figure 8: Knee angle and impact force variations in pedestrian tests

What has been established also from these tests is that fractures or injuries to ligaments do not occur if the deformation angle between the thigh and lower leg is less than 15 to 20° This is shown in Figure 9 in which a result in the positive direction corresponds to the shearing process, not the bending, in the knee. These values may, in fact, be too high because the analysis was made in 2-D and it seems from later tests that the motion of the leg is not purely a 2-D motion if the cadaver is hit from the side; if there is some rotation of the leg there can be a loss of some natural motion which does not produce any injury.

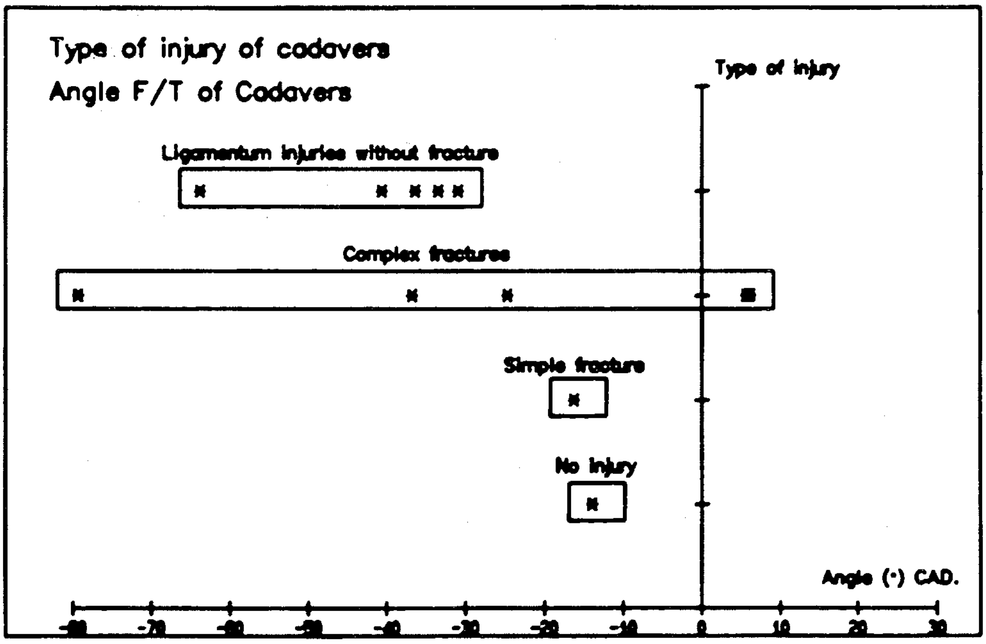


Figure 9: Injury pattern in the legs of cadavers

Experimental improvements were made in work on the tolerance and mechanisms of injury to the leg. As the mechanisms in shearing and bending had been found, they could be reproduced in just one leg rather than in the full cadaver in a dynamic laboratory test. Figure 10 shows the impactor with 2 points on it and the leg attached to a frame. By loading close to the knee, a mainly shearing process is produced. Twenty tests have been done with that procedure and similar work is going on with bending, in which the impact is only at ankle level so that the leg bends sideways at the knee. This assists understanding of what happens in the dummy during the specific process reproduced in the laboratory.

For bending, results are not yet available. For shearing, the force/time histories for impacting force and sustaining force are shown in Figure 11. There is what is called an injury window in which injury occurs and for shearing this is very early if the timing is considered, about 5-7 ms from the beginning of the impact. So it is not necessarily the peak value of the force that should be considered, but the value at the time at which the injury occurs.

A short video of various laboratory tests was then shown. Both car and platform impacts with cadavers were included to show the motion of the leg at and immediately after impacts at different heights and in situations in which either shearing or bending predominated.

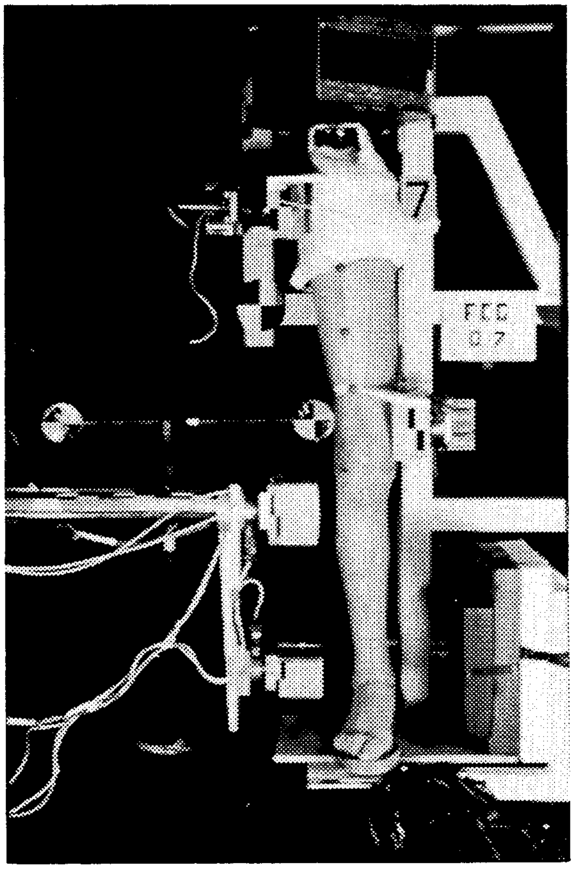


Figure 10: Test set up for cadaver leg and 2-point impactor

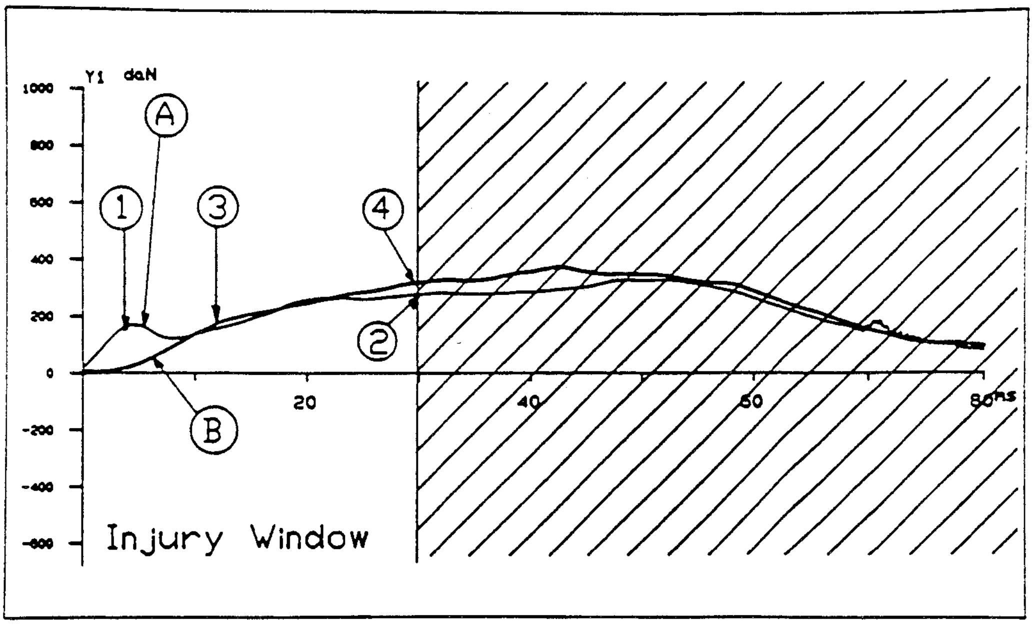


Figure 11: Typical characteristics of knee impact force (A) and knee reaction force (B)

As a better understanding of the problem of pedestrian leg injuries had been achieved, a joint European program sponsored by the European Community then developed a test procedure at INRETS to evaluate the severity of the impact of the car to a pedestrian leg. Figure 12 shows the test feature, a mechanical leg, which is propelled against the front of the car. This leg, especially the knee joint (Fig. 13), is quite sophisticated. It has a good deformation path or deformation rod with the same force deformation characteristics as the human knee hit from the side. So the characteristics of deformation of that are very well controlled.

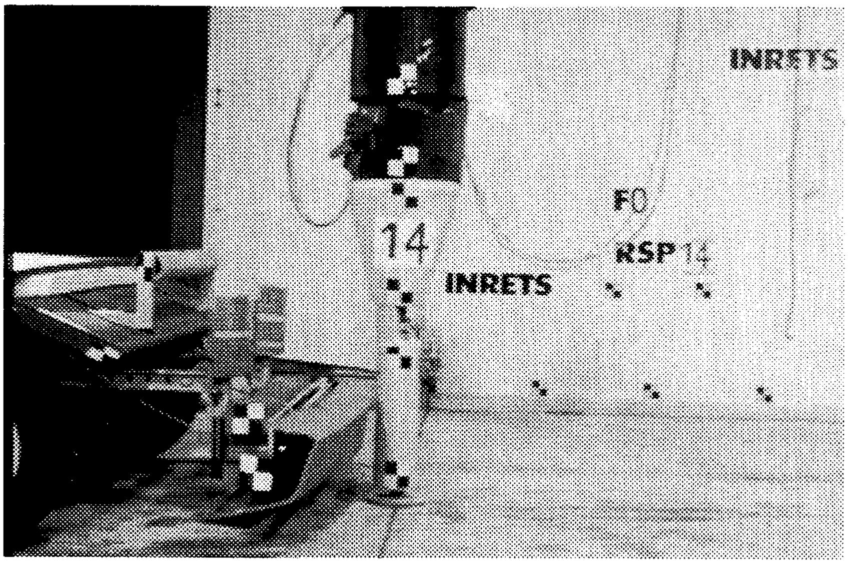


Figure 12: Mechanical leg developed at INRETS in a joint European program

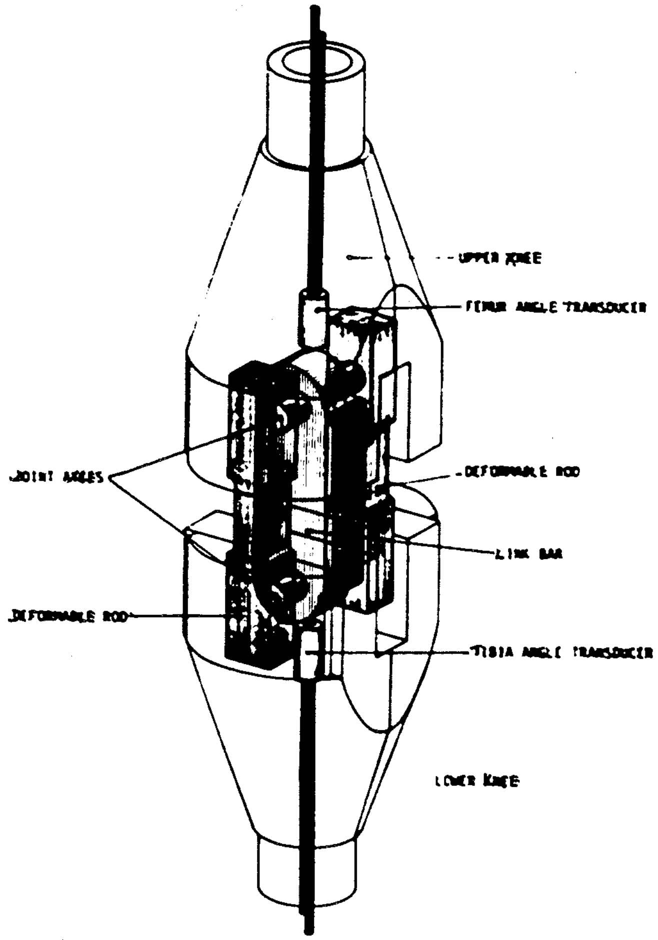


Figure 13: Principle of the new pedestrian knee model

At the same time there is what is called a link bar attachment between the top and bottom, which measures the angles between itself and the lower leg and the upper knee respectively (Fig. 14). From those two angles it is possible to determine what is due to shearing and what to bending. Also, there are the two transducers for angle and a place for accelerometers to relate to the long bone injuries. During an impact there is deformation of the deformable rods and the angles between the rods and the upper and lower legs are measured. The two processes can occur and maybe mix during an impact which is mainly bending or shearing and Figure 14 shows such a combination. If only shearing occurred, there would be parallel axes on the top and the bottom. From measuring the two angles a computer program has been developed, which is able to make the separation between bending angle and shearing displacement.

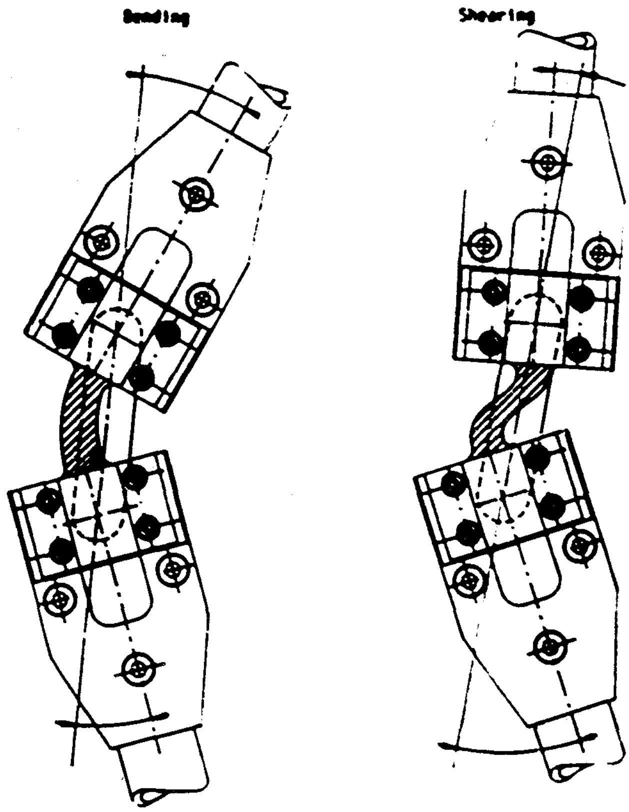


Figure 14: Deformation process of the mechanical knee

One problem was whether the weight of the leg alone was sufficient or if a mass representing the upper body was needed. Initially, it was clear from the high speed film that the upper body did not move during the time in which the leg was loaded to its maximum deformation, or a least to the injury level. So to be sure that a mistake was not being made, a check was made using MADYMO — in 2-D at that time — to model the leg with the joint characteristics and with the same leg, the kinematics were examined fully as well as some other parameters like bumper force or angle to the knee or acceleration. Figure 15 is the model of MADYMO with the leg alone and Figure 16 is the model with the leg and the mass which was added at the top of the leg. The same model was used with a complete dummy (Fig. 17) but with a knee joint having the same characteristics as the mechanical leg. This model was used with different bumper heights and different speeds of impact. Not all of the results are shown here, but as can be seen in Figure 18, if the impact is at knee level or below knee level, there are very few differences whether or not there is a mass for the upper body. This Figure shows the bending angle of the knee, in radians, for the leg alone, the leg and the mass, and the dummy; the values which are reached with the dummy and the leg alone are very close. So if the test is done at knee level, or even below knee level, closer values are obtained. At knee level or below, this leg can be used without a mass corresponding to the upper body, but this is not the case if the impact occurs above the knee.

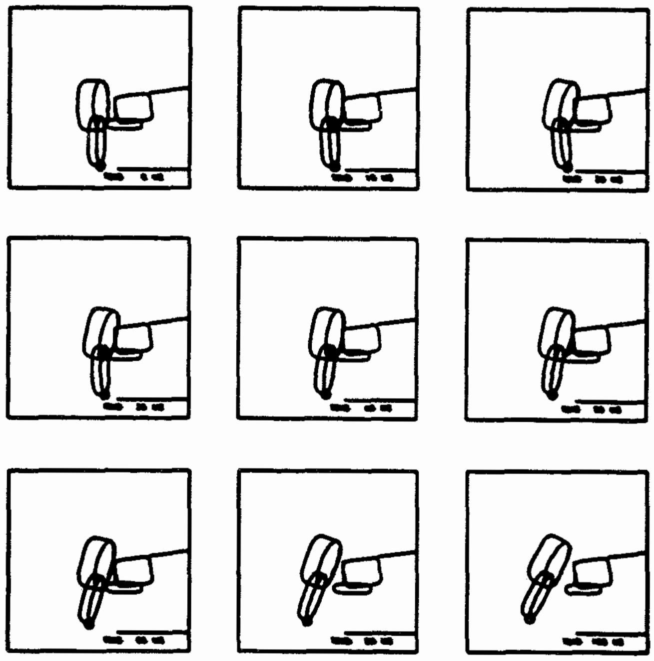


Figure 15: Movement of the leg impactor
Times, from left to right: 1st row - 0, 10, 20 ms;
2nd row - 30, 40, 50 ms;
3rd row - 60, 80, 100 ms.

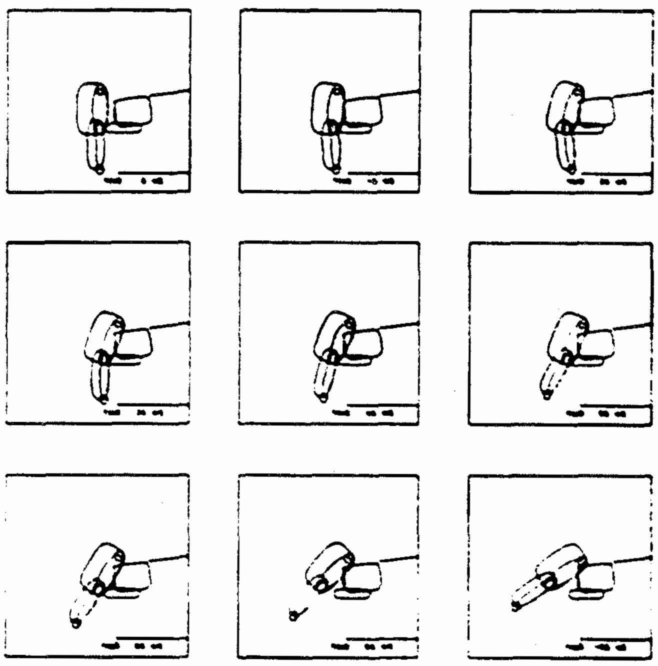


Figure 16: Movement of the leg+mass impactor
Times, from left to right: 1st row - 0, 10, 20 ms;
2nd row - 30, 40, 50 ms;
3rd row - 60, 80, 100 ms.

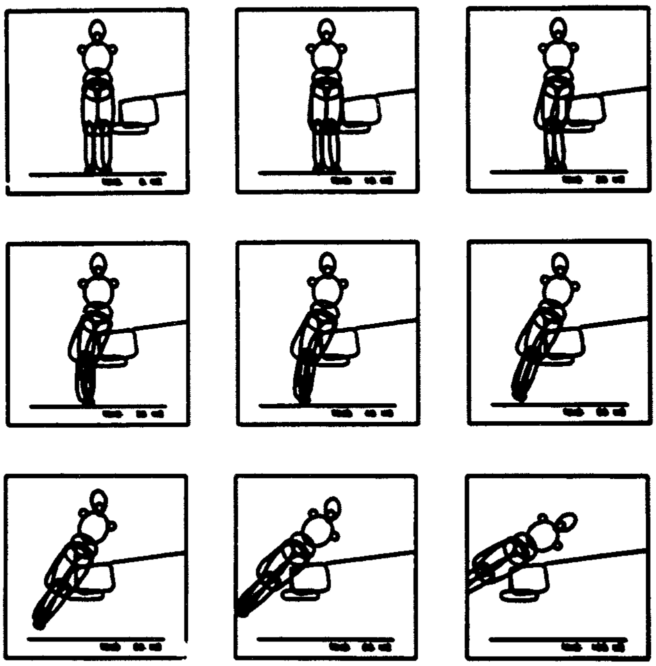


Figure 17: Movement of the dummy
Times, from left to right: 1st row - 0, 10, 20 ms;
2nd row - 30, 40, 50 ms;
3rd row - 60, 80, 100 ms.

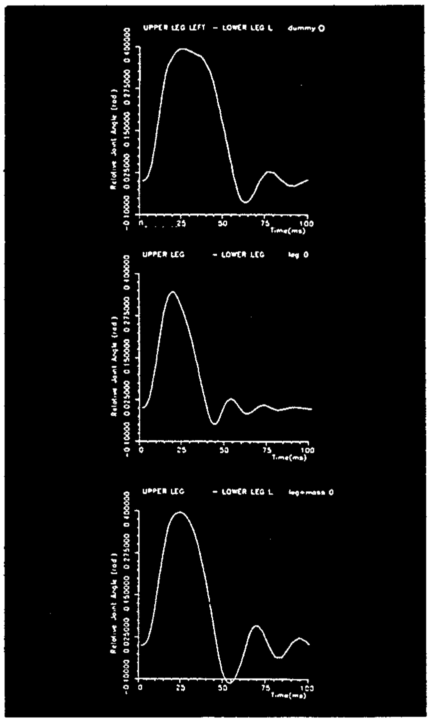


Figure 18: Knee joint angle by time for dummy (top), leg alone (centre) and leg+mass (bottom)

This work is part of a subsystems test procedure proposed for the EEC, which includes the bumper test, as has been discussed, and also a bonnet leading edge and a bonnet top test. The bumper test involves impacting the front of the car with the instrumented leg. The bonnet leading edge is mainly for child protection, for the thorax especially, and it is an impact on the bonnet leading edge with a guided impactor. The bonnet top test is a drop test for head impact on different parts of the bonnet with adult and child models. All of these tests are based on car to pedestrian impact velocity of 40 km/h.

To conclude this presentation, a video of impacts against model legs was shown. The dynamic deformation of the leg was measured for different bumper heights and different speeds and it was seen that the deformation was less when the bumper height was relatively low.

QUESTIONS/COMMENTS for Koshiro Ono and Dominique Cesari

Chris Coxon: I wonder if you have any confidence in the speed of the pedestrian impact, because it seems to me to be a very troublesome thing to estimate, if the car is slowing down quickly, how you would know what the impact speed might be. Both of you had charts that showed vehicle speed at the impact point. Would you comment on that?

Dominique Cesari: From the accident, it is not very easy. Nevertheless there are different studies done by different people which show a range of variation and you can see if you are inside that range. From the cadaver tests we have done, we know at least for the leg injuries, in which range of speed these injuries occur and they occur in the range from 20 km/h and above. This gives the range in which we have to work to improve the situation. The objective for this work that I have just presented is to go up to 40 km/h for the severity of injuries which may occur at that time at around 30 km/h, so as to have a gain of 10 km/h for the same severity. But I agree, it's very difficult from accident analysis and there is no full agreement between different studies where they are considering the same type of car, the same type of population, there is some variation.

Rolf Eppinger: I was wondering if both of you would comment on the ratio of tibial fractures to knee type of injuries. The reason I ask is that, in the majority of the tests that people conduct for pedestrians, the weight always seems to be on the impacted leg and the leg is in a stiff or straight position. But a pedestrian, I imagine, is in a walking cycle and there would be times when the leg is higher and you would have, say, the bumper striking the mid shaft of the tibia. So I was wondering how you could come to some test condition, or could you gain some information from the accident investigation to understand whether the leg is off the ground or on it?

Dominique Cesari: I agree with you that probably the long bone fractures, especially of the tibia, are in terms of frequency more common than are knee area injuries. But in terms of long term consequences which cannot be found just from AIS values, as was discussed before, knee injuries are probably more severe. That is one thing. The second is that it seems from the tests we have done that knee injuries occur more in a certain range of speed. Higher speed may produce more bone fractures because the impact is a very quick variation of speed, but on the other hand a soft bumper, which we think is desirable, may protect from bone fractures but may increase knee injuries because it would involve a longer loading. So I don't have any idea on the frequency except that bone fractures are more frequent than knee injuries. The other thing is that we find more ligament injuries in tests we are doing with cadavers than on the road in traffic accidents, probably because we have the foot on the ground whereas with people who are walking with one foot not in contact with the ground, the loading process may be in a different point. Also, ground friction does not apply to that and this makes some change. But by having the feet on the ground, we think we have the worst conditions and that is the reason why we are using that position. Nevertheless, in the method we are proposing, we have a measure of the acceleration to the long bone and we should avoid whatever the risk of long bone injury is.

Rolf Eppinger: Are you doing anything to measure the potential for long bone fracture, like a bending moment at mid-shaft or something?

Dominique Cesari: Not at present. This work can still be developed and what we are planning is to add at the top of the tibia the transducer which we have developed for the Hybrid III dummy, which is a multiple axis transducer, and to see if we can correlate that with some different input in terms of stiffness or bumper height. But this has not been done. It is just in the mind up to now.

Rolf Eppinger: I might make a few comments about NHTSA's ill-fated notice of proposed rule-making for pedestrians where we had actually proposed to put a soft bumper on a car. This was based on cadaver testing and we noticed at the time that we were testing with a particular model which had a fairly square front edge and a high bonnet leading edge. As we softened the bumper and limited the force that the bumper would produce in the knee area to 700 lbs, we significantly reduced the ligamentous damage in the knee area. And so we actually went out with a notice at that time. The way rule-making goes, it usually takes several years for things to proceed on through, and meantime we changed headlight regulations and a variety of other things which allowed much more aerodynamically clean cars to be in the fleet, and so it was later suggested that we should test with a more modern shape. Lo and behold, we did that and made both the soft bumper and the hard bumper with a much more sloped hood and we found that the upper edge, when it picks up the upper femur, has a significant influence on whether the lateral knee rotation and ligament damage occur. And by softening the bumper we could not then produce the effects in the knee area we had found with the higher shaped bumper. So it seems that if you deny yourself the ability to control the hood leading edges and what the characteristics are, I think the pedestrian problem becomes extremely difficult.

Ken Digges: One of the concerns is that as you lower the bumper height the head velocity goes up. Have you looked at that, and is that any kind of a problem that you are concerned about and do you address that at all in your regulation?

Dominique Cesari: Yes, it's clear but I did not mention that because I did not comment on the regulation for head impact. But if the lower leg impact should be done at a constant speed, 40 km/h, the head impact speed varies according to the shape of the car, taking into account not only the bumper but also the bonnet leading edge. Using a computer model there is some adjustment because, as you say, we get a higher speed with a lower bumper and a more aerodynamic shape.

Chris Hall: We've been talking about pedestrian accidents and car accidents but we seem to have forgotten the most important form of transport, which is the motorcycle. I was wondering if any of your work on lower limb injuries is relevant to transfer across to motorcycle accidents where lower limb injuries are quite severe and very very common. Could you comment on that, please.

Dominique Cesari: We are not doing this but I had the opportunity to look in depth at the work done specially by the Transport Research Laboratory on leg protectors. This showed that there is some possibility to protect part of the leg injuries of motorcyclists, at least those which occur in collision with cars: this does not necessarily hold for single vehicle crashes with the motorcycle, in a long flying motion, impacting any of a variety of objects. But there is also a much more important risk than for the pedestrian, that the trajectory of the upper body will be changed completely, because restraint at the knee level produces a different rotation of the upper body. This is mainly the reason why this work has not been through legislative process. The main discussion was on whether protection from injuries which are quite severe, because leg injuries of motorcyclists are more severe than for pedestrians, generally, may increase the risk of more severe or fatal injuries to other parts. That is the reason why so far the matter is still in discussion. So I would say that generally speaking the question of protecting motorcyclists is a very very difficult one. Nevertheless they are a high risk population and there is an ISO working group to be set up after this European summer to discuss the question of not only leg protection but also the airbag problem for protection of the upper body.

Koshiro Ono: In Japan we have performed accident investigations on motorcycles but interest in the frequencies of the injuries to body regions is focussed on the head which has the highest frequency in fatal cases. In the case of the impact direction of the motorcycles, there are several kinds of impacts which depend on the type of the motorcycles. In Japan we have five categories: up to 50 cc; 50-125 cc; 125-250 cc; 250-400 cc; and >400 cc. These kinds of differences produce a variety of accident situations and it is a little more difficult to focus on the type of direction. So now in Japan in the case of protection of motorcyclists, we focus on head impacts by the wearing of helmets and such like.

NECK INJURY IN REAR END COLLISIONS

Ingrid Planath

Not long ago the Volvo Accident Research Group, together with a hospital in Göteborg, the East Hospital, carried out a study on neck injuries in rear end impacts. The Volvo accident data shows that, irrespective of whether there are injuries or not, rear end crashes account for about 7% of all crashes. However if only serious or fatal crashes are considered, 1% are rear impacts. So from this it can be concluded that most rear end impacts result in rather low AIS scores for the injuries.

During this seminar there has already been some discussion about what information is not given by AIS scores. In rear-end crashes the typical injury is neck pain which gets an AIS score of 1. This neck pain is what is commonly called the whiplash type because of the movement that the head, neck and chest describe in a rear end impact — it looks like a whip lash. Some of the persons who sustain this type of injury would not agree with its AIS rating as a minor type of injury, because the pain can be significant and not only in the neck but it can radiate also to the head, the shoulders and even to the arms or legs, and the discomfort can last for months and even years, in some cases, after the accident. So the disabling effect on the person sustaining this injury can be significant and this certainly justifies a study on this subject. Also any insurance company would support the importance of this type of injury.

The aim of the study that was undertaken together with the East Hospital was to investigate in-depth the whiplash type of injury in rear end collisions. There are a number of research teams that have studied whiplash over the years but the injury mechanisms and also the injury causing parameters are not yet fully established. One reason for this may be the nature of the injury, because examination in normal diagnosis is by methods like X-rays in which the whiplash injury will not appear and there usually are no outer signs of whiplash injury: there are no scars, no other signs. So the physician will have to rely on what the individual says. In this study the plan was to look for any correlations between accident parameters, also occupant and vehicle parameters, and the occurrence and severity of whiplash injury. Severity was measured by the duration of the discomfort caused by the injury.

The basis of the study was a number of rear end collisions that were reported to Volvo's Accident Group, all of which occurred in the Göteborg area. Only occupants who had no previous experience of severe neck or shoulder discomfort were selected and a rather limited sample of 26 rear end collisions in which 33 front seat occupants were travelling, resulted. All of the occupants were belted and all the cars that were studied were equipped with head restraints. The head restraints were of various kinds and the vehicles were different types of Volvo cars. For each of the occupants, extensive interviews and a medical examination were performed and these two items were undertaken as soon as possible after the accident. Also, the car was examined, car deformation was measured and attempts were made to position the occupants in the cars. The last item was a unique feature of the study.

During the interview, information like age, sex, height and weight was requested. The subjects were also asked if they had been aware of the rear end crash approaching, if they thought they had been turned during the crash and if in their daily occupation or daily work had hard physical work included or had previously had such work experience.

The medical examination was carried out by physicians and they would look at symptoms and signs that are associated with neck injuries. They would look at the range of motion of the cervical spine and also at the discomfort that could be related to specific movements of the neck. This information was documented very thoroughly and, of course, if there was injury, appropriate treatment would be given.

The long term consequences of the neck injuries were assessed by further questionnaires which were given to the occupants one, 3 and 12 months after the accident took place. If there were still neck symptoms after one year, a second medical examination was made.

The neck injuries were classified according to a set of criteria that have been developed by a Finnish researcher by the name of Varis and in this classification, factors like pain in the shoulder or head, tenderness in muscles or numbness in arms and other similar symptoms are used. Figure 1 summarises the results from the medical examinations. Immediately after the accident there were 4 people who did not have any neck discomfort at all and the other 29 sustained various types of neck pain, with neck or shoulder discomfort of different degrees. A new questionnaire one week later showed that the number of persons still having neck pain had been reduced by about one third. Another drop was shown by the one month follow-up, but this promising trend did not continue. Regardless of whether it was one, 3 or 12 months after the accident, there were still about the same number of people with neck discomfort.

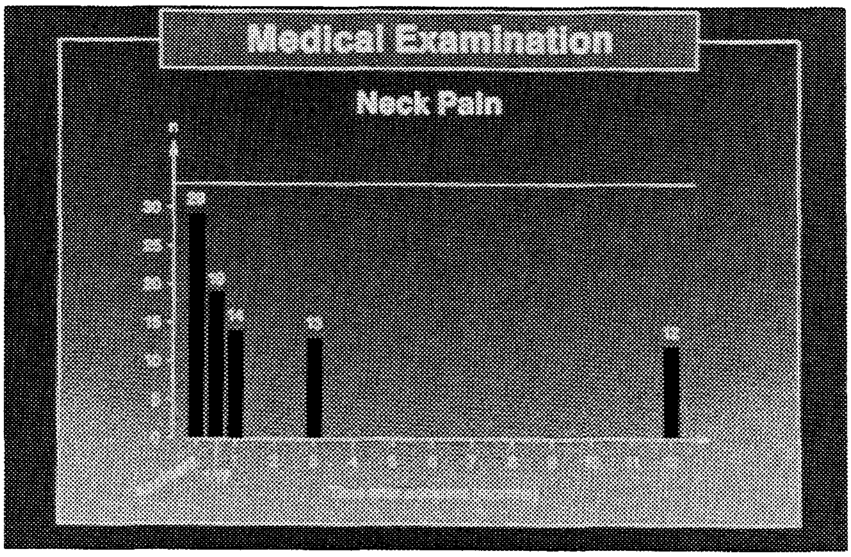


Figure 1: Cases with neck pain at intervals (months) after their accident

Regarding the vehicle parameters, the maximum deformation of the car was measured and by comparing this with corresponding values from laboratory barrier tests, the energy equivalent speed could be determined and this was used as an estimate of how fast the impacting vehicle had run into the rear end of the investigated car. Most of the impacting vehicles were cars; in one case there was a truck impacting and in 2 cases, buses. The investigation also included a classification of the crash pulse: the terminology of a *stiff accident pulse* was used if at least one of the rear side members had been hit or permanently deformed and a *soft crash pulse* if this was not the case and the rear side members remained unchanged (Fig. 2).

One of the reasons for the limited sample size that was chosen for the study was the rather time consuming activity of occupant positioning. It was intended to position the occupants in the car that they had actually crashed in, but in some cases this was not possible so a similar car had to be obtained. The occupant was asked to seat himself or herself in the ride position that they considered as their normal ride position, and the horizontal and vertical distances between the head and the head restraint were measured. It was hoped thus to obtain the state as it had been before the impact. The seat back angle normally used was also looked at and observation of the car also showed if there had been any seat back deflection. The residual seat back deflection was defined as the difference between what the persons described as their normal deflection and the deflection that was measured after the crash (Fig. 3).

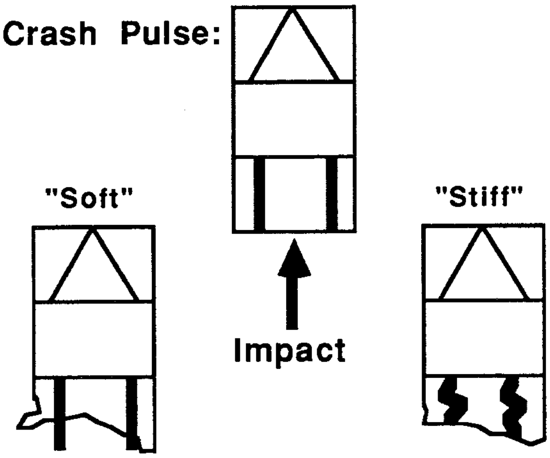


Figure 2: Crash pulses

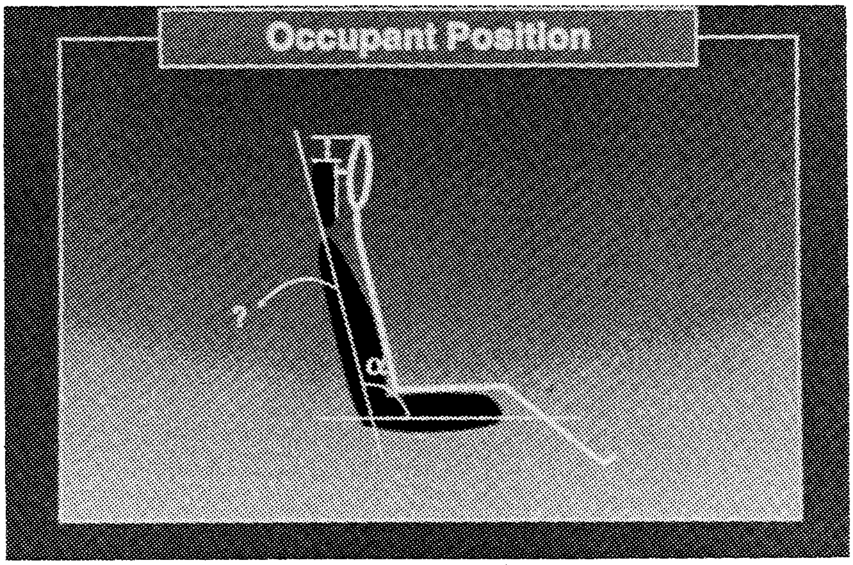


Figure 3: Seat back deflection

So after making all these measurements, one might ask what conclusions could be drawn. Logistic regression was applied to evaluate the correlations between the parameters obtained and the occurrence of longer neck pain duration and for this regression the horizontal distances between the head and the head restraint were divided into 2 groups: those longer than 10 cm and those closer to the head restraint. It was found that prolonged neck discomfort which lasted at least one year, correlated with the longer distances to the head restraint. Neither impacting speed, which had been determined from the degree of rear end deformation of the car, nor the crash pulse by themselves correlated with prolonged neck discomfort. However, in cases with a stiff crash pulse, the square of the maximum deformation would somewhat relate to the occurrence of prolonged neck discomfort. This was not statistically significant but was found as a trend anyway. An earlier study had showed that the height of the head restraint had some effect, or rather that the shorter persons in the study did not sustain neck pain to the same extent as taller people did. But this correlation with the vertical distance from the head and the head restraint could not be verified with the limited sample in this study.

Just one more word about this limited sample is that although the sample was too small for general conclusions about neck injury occurrence to be drawn from it, some interesting trends occurred that would be worth further research.

Another aim of the study was to find out more information about the injury mechanisms behind whiplash injuries, but this could not be clarified in detail in this study. But as the relationship between the increased horizontal distance between the head and the head restraint, and whiplash occurrence was found, it was thought that this indicated that it would be useful to reduce the backward movement of the head in relation to the chest. As illustrated in Figure 4, this backward movement of the head is probably a combination of two mechanisms — shear and hyperextension — and it is reasonable to believe that the shorter the distance between the head and the head restraint, the less is the likelihood that the various tissues that are involved would be strained.

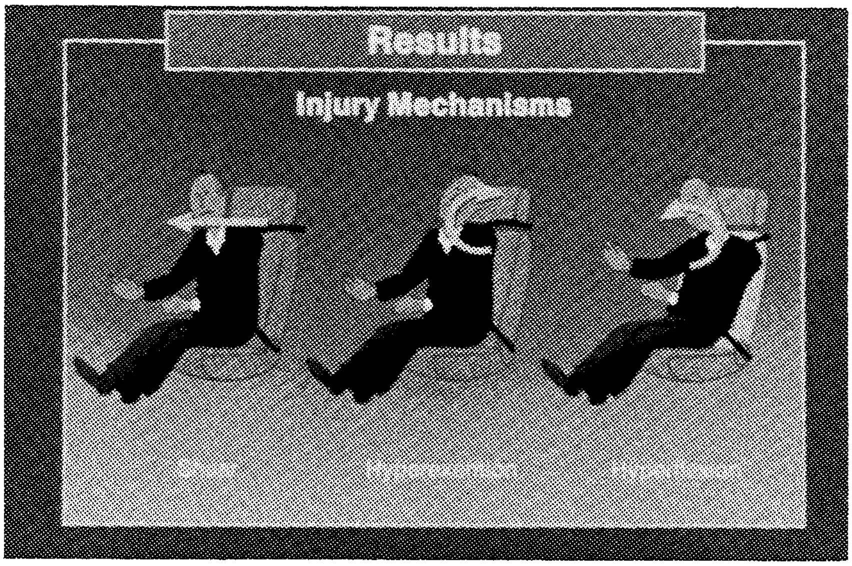


Figure 4: Injury mechanisms in whiplash

Hyperflexion is by definition a part of the whiplash motion, but an isolated hyperflexion would cause symptoms that would be difficult to distinguish from those found in a combined hyperflexion and hyperextension movement. So it seems reasonable to reduce both of these motions. Even if the head restraint is designed to be in an optimal position to prevent hyperextension, hyperflexion might still occur. It seemed that if the seat back and head restraint were very elastic, hyperflexion would be likely to occur. This is described by the diagrams in Figure 5: the elasticity of the seat back is shown by the springs and in a rear end collision the head and spine will move forward, the springs will be compressed and this will keep the occupant away from the seat after the compression and at a certain moment, the chest of the belted occupant will be restrained by the seat belt but the spine and head can still move forward, so a hyperflexion occurs. Of course this should not be taken as a criticism against the seat belt, because the overall effects of that device were justified long ago, but some modification may be necessary.

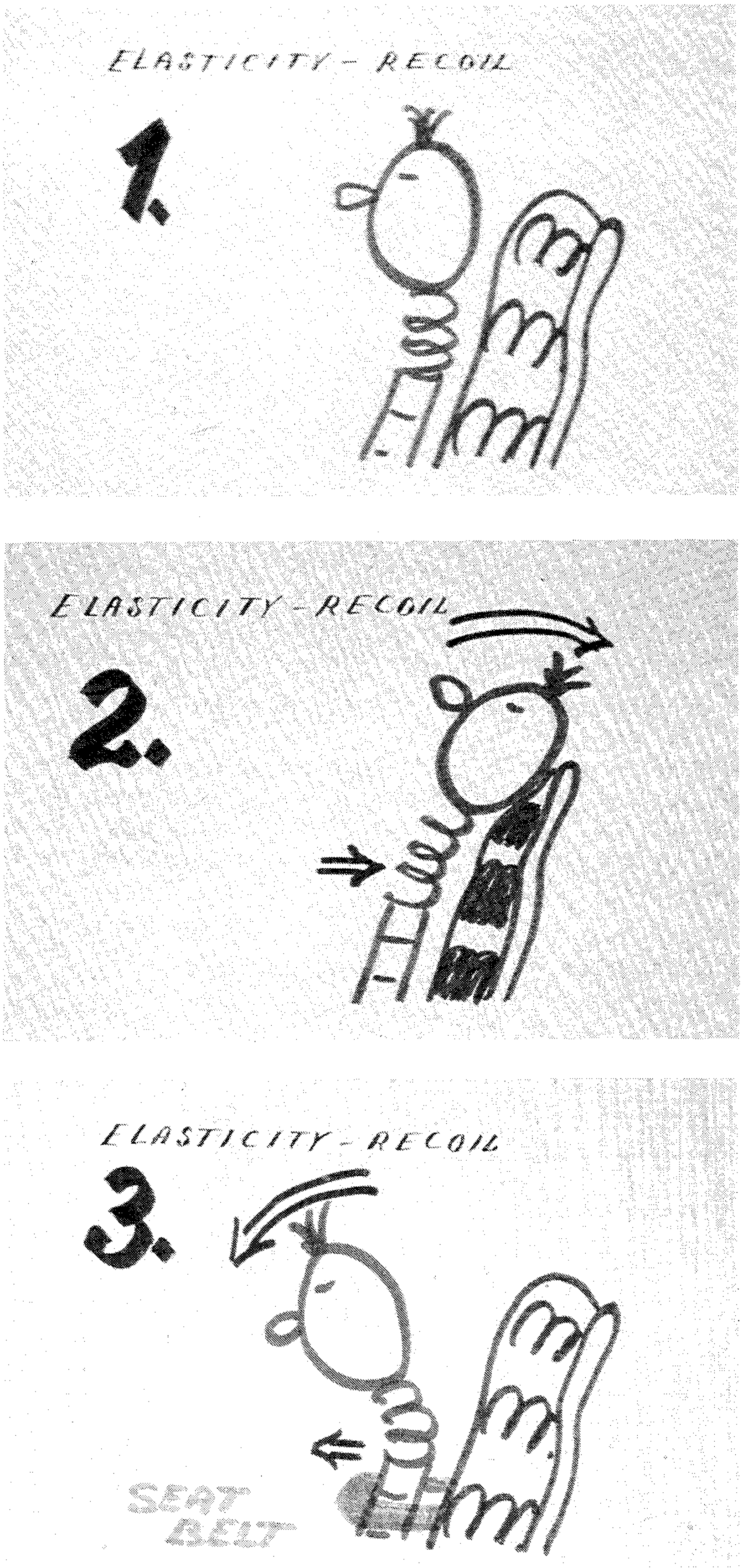


Figure 5: Diagrammatic representation of elasticity recoil

So although based on a small sample, the findings of this study indicate that to decrease the probability of whiplash injuries the head restraint should be designed so that it is close to the occupant in the case of an impact (Fig. 6). Also, the side members should be designed with care so that they do not deform too much in the case of a rear end impact and the seat back, together with the head restraint, should not be too elastic.

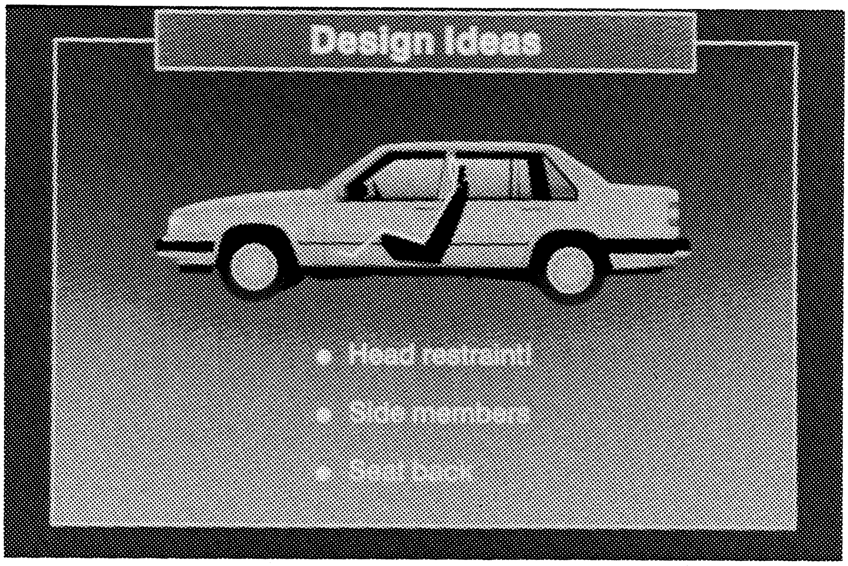


Figure 6: Ideal positioning of head restraint

So these are some conclusions, but they probably need some time to be applied. One thing that everyone could apply readily in daily life follows the finding on the horizontal distance between the head and the head restraint. Anyone who is driving in a car should take advantage of the head restraint which is in most modern cars and just relax against it so that they will not add another item to the accident statistics on neck injury.

QUESTIONS/COMMENTS for this paper follow the next talk

NECK INJURY

Rolf Eppinger

With regard to projects that are currently underway, NHTSA has no work in terms of regulation for neck injury, but is considering merely something possibly to control neck tension during a forward impact. There is likely to be a continuing and increasing need to understand neck injury problems because, as higher performance seat belt systems are developed, there may be a lot more flexion going on in vehicle occupants. As restraint designers attempt to minimise contact of the head with the steering system by the use of pretensioners and things of that nature that tend to pull the body back, the amount of load and how fast the load is being applied to the neck and head will probably increase. So it could be conjectured that that would be a potential problem.

Certain types of restraint systems are evolving now which could present new hazards to the head and neck that have not been anticipated. One possibility is the airbag inflation hazard to unusually situated occupants. At present, the probability of exposure, how many people would probably be in that type of situation, is unknown. But if, in that kind of position, the unfolding and rapidly deploying device could possibly capture the chin and cause the neck to experience an extension and tension load, depending on what the crash constraints and parameters of the system are, you would definitely want to examine and control neck loads. NHTSA is currently in the process of formulating its own programs to research the various types of neck loading conditions, but no testing efforts have been initiated at this point.

There is a little information from a study carried out for NHTSA by Dr. Jerry Barancik from Brookhaven Laboratories on Long Island, which has not yet been formally published. He looked at people who complained about chronic neck pain, very similar to the types of car occupants just described in the Volvo study. These individuals had progressed through the medical process and they had been told that although they had a neck pain, nothing could be found and there really shouldn't be a problem. Dr Barancik performed MRIs on these individuals, typically 4 weeks to 6 months subsequent to the accident in which the subjects were involved and found that in a very high percentage of these cases, identifiable lesions were present. These patients were not exposed to an MRI immediately after the accident because their injuries seemed to be of fairly low severity when they were in the emergency room and they basically were just dismissed from the hospital. So it is not known whether the lesions were there at the onset, or took some time to develop and manifest themselves. But although these things are seen and seem to be very low level initially and are not considered problems, people who do suffer from the chronic problems may have lesions which can be found and potentially alleviated by appropriate medical procedures.

I regret being unable to present more material about spinal injuries, but would like to make you aware of a report that NHTSA has published. It is called 'A Review of Biomechanical Impact Response and Injury in the Automotive Environment' and was a Phase I report that came out of the advanced dummy program. It comes under a DOT number and is available at NHTSA. What it does is review most of the current literature on each body region in turn, so it has lots of references in it and I'd like to recommend that to anybody who is interested.

QUESTIONS/COMMENTS for Ingrid Planath and Rolf Eppinger:

Tony Ryan, comment: I just happen to have some preliminary results of a study of neck strain that we are carrying out here in Adelaide, that I'd like to show you. I was very interested in the description of the Volvo study because our findings are paralleling that study.

The subjects are referred from doctors and physiotherapists; they have an assessment with a physiotherapist which includes the range of motion of the neck, and MRI scans are done of the necks of those who have had their collisions within about a week of the time of the examination. This assessment will be repeated 6 months after the original injury, so that changes in physical condition and MRI scans can be determined. The vehicle is examined — both vehicles if possible — and the site is examined, so that the magnitude of the impact can be determined.

The object of this exercise is to try to correlate the persistence and severity of the symptoms with the magnitude of the impact. There have been a lot of studies of psycho-social-economico-compensation factors associated with persistence of symptoms, but I'm not sure that there has been a lot done on trying to determine the magnitude of the impact that the victims have suffered.

Preliminary results have been obtained for about half the cases so far. Part of the assessment from the physiotherapist is his estimate of the severity of the symptoms related to a ΔV for the impact. A ΔV could be obtained only for a little over half of the cases so far and the results show clearly that most of the cases fall within the middle range of severity and in the range of 10-20 km/h ΔV. We also asked the subjects at the time of the examination to report their symptoms on a scale from one to 10 in which one is no pain and 10 is absolute agony, the worst they've every had. This was related to ΔV and there does seem to be quite a nice correlation in most cases: there are only 2 out of 15 that don't quite fit this pattern .

This is just something that I thought was particularly interesting in view of the discussion. The other thing is that half of these cases were rear impacts. The remainder were frontal and side impacts — so it's not just rear impacts that are associated with these symptoms.

John Lane: I heard the other day that someone in Newcastle in New South Wales has also been using MRI and found definable lesions in the intra-articular joints. The paper, I believe, is coming out in the British Journal of Radiology soon. I have another comment. There are other data in the literature about the role of the horizontal offset between the head restraint and the head, and the liability to get whiplash, and I've often wondered why the offset is so large. It's not mandated by the design rules, to my recollection, and I wonder why the manufacturers don't put the surface of the head restraint a bit closer to the head.

Ingrid Planath: I guess I should answer that. We would like to put the restraint very close to the head, but by doing ergonomic studies or asking people what they actually think of that, we have found that most people don't want to have something touching their head. So, it's desirable to get closer, but there are a number of people, elderly drivers especially, that still will lean forward.

Laurie Sparke: From personal experience in evaluating head restraints that are very close to the head, a guaranteed way to get a neck pain is to drive a car where the restraint is too close to the head and you actually lean forward to provide clearance.

Bryan Knowles, to Rolf Eppinger: In work we've done in the past, we've used neck injury criteria or guidelines that were generated by General Motors some years ago, which were a time weighted analysis of the various neck loads. One question is, are those criteria still applicable or useful, or not? The second question is, that without a position-dummy-testing with a Hybrid III dummy, the lack of a representative neck skin can cause some quite alarming head rotations and, in fact, we've grafted on some Hybrid II neck skins to do that; is that appropriate or are any changes planned to the Hybrid III dummy to incorporate that officially?

Rolf Eppinger: Regarding the neck injury criteria — we're in the process of trying to evaluate them. At present they seem to be the most appropriate criteria available. When we initiated our Hybrid III rule making, we had proposed to put the neck moments and the shear forces into the regulation. But a difficulty was that the available neck criteria defined moment/injury characteristics in the presence of either hyperflexion or hyperextension. We found that there were certain types of test conditions in which we could create large neck moments in the absence of hyperflexion or hyperextension. So from a rule-making stand point, to be consistent with the published research, we had to determine whether there was something we could measure on the dummy to see first if it was in either hyperflexion or hyperextension and then apply the moment process. We thought that if the neck was in either one of those conditions — i.e. hyperflexion or hyperextension — the neck load cell should show large shear values, and if shear exceeded a certain level, then the moment criterion would become effective. We were unable to demonstrate that process. We were also reluctant to go back and require some sort of photographic coverage to establish the position of the dummy neck. So we're trying to re-assess how we could apply the moment criterion, either by some kind of angular measurement in the neck or some other parameter. But I think that if, today, I were a restraint designer who had to try to create a system that's safe, I would definitely monitor neck shear and moment loads and use the current criteria as recommended practice. There is always a difference between regulation and design. So, I would say that if you're designing a system, use the available criteria as bogey values — that would be a very good thing.

At this point, there's no formal plan to incorporate a neck skin into the Hybrid III system that we use for our testing situations, but I think the industry is free to adapt those things to these more unusual test circumstances, with the development of restraint systems.

Bryan Knowles, to Ingrid Planath: Seat back strength has had some media attention here in recent months. In the rear end studies that you did at Volvo, did you draw any conclusions with regard to appropriateness of seat back strength or deformation that would either help or hinder injuries to people in cars?

Ingrid Planath: We only looked at these neck injuries and we could not find any correlation between the seat back and injuries such as that the persons were more likely to sustain a neck injury if the seat back deflected, or that collapse of a seat back was unfavourable. In fact, I think it is Peugeot Renault who have presented a research paper where they proposed a controlled seat back collapse in order to reduce the neck injury.

Michael Graham, to Ingrid Planath: Was there any correlation between age, sex or occupation and neck injury?

Ingrid Planath: No, the only correlations that we could find were those that I presented, so we could not attribute any higher probability of getting a neck pain for older people or females.

Ken Digges, to Ingrid Planath: I was wondering if you did any reconstruction to determine what kind of moments and shears you were getting in the population you looked at?

Ingrid Planath: We did not reconstruct these particular cases but we have done sled tests to see if we could find out design criteria for neck loads in whiplash; however the dummy is not very good in rear end impact — it is far too stiff, we believe, over the shoulder and back areas. So when we have something we will present it, but I think that is far away.

Ken Digges, to Ingrid Planath: Could you comment at all on the level of the Mertz criteria in comparison with what you've been seeing in whiplash?

Ingrid Planath: What we have looked at is the extension moment and the 57 Nm that are proposed by Mertz I would suggest are too high.

Ken Digges, to Ingrid Planath: One more question: in your examination of these patients, could you see any evidence that would relate the pain to some kind of lesion?

Ingrid Planath: Not that I know of. I was not involved personally in the medical examination but I don't remember that we ever discussed that there was such a sign.

Ken Digges, to Rolf Eppinger: In the Barancik research, is there any way to tell whether or not the lesions are related to hyperflexion or hyperextension or both, or is there any insight that we can get out of that to determine what the nature of the loading is?

Rolf Eppinger: At this point I don't think there was. We knew generally what the crash condition was but there was no way of coming back because it was such a long retrospective study and there was no information about what the vehicle was and so forth.

Tony Ryan: In talking to the people that we are recruiting into our study, we find that their recollections of the impacts are that they were thrown forwards and then backwards, in rear impacts. The rear component seems to get lost from recall. They all comment on being thrown forwards and then backwards.

Brian Fildes, to Ingrid Planath: I understand that in Sweden, 3 or 4 years ago, there was discussion, if not implementation, of a prospective long term study of whiplash patients that was supposed to be undertaken in the north. I wonder if you are aware whether that study ever got under way and if it did, whether there are any results of it available yet?

Ingrid Planath: I don't know if it has been underway. If it has, there haven't been any results published as far as I know.

Brian Fildes: I think it was aimed more at looking at what the mechanisms of whiplash might be amongst a large sample of patients.

Jack McLean: I have two comments. A couple of years ago in the Unit, Tony Ryan and a surgical registrar, Chris Cain, managed to investigate about half a dozen cases where people presented in hospital immediately after an accident complaining of sore necks. About half were in side impacts, the rest in rear impacts. MRI scans were done at the time and again a week and sometimes a couple of weeks later. Where the MRI scan identified disc lesions in the cervical spine, the lesions increased in size a few weeks later.

Also, about 20 years ago in North Carolina, I worked on a study for the Motor Vehicle Manufacturers' Association on the effectiveness of head restraints, identifying cases through the State Highway Patrol who reported back from accidents in which passenger cars were hit from the rear, and collected information on the position of head restraints, deformation of the seat back and so on. The overwhelming factor that we identified in about a thousand cases was the difference between men and women — this is self reported, admittedly — in susceptibility to neck injury. There were also differences between drivers and passengers, which didn't account for the male/female difference, and we had enough cases to be able to control for the height of the individual, and that didn't account for the male/female difference. But there was a difference in susceptibility of about 50% greater for females than for males. The only head restraint benefit we could detect in that study was with the high seat back and that accounted, from memory, for a reduction of about 10% in the severity of neck injury.

Peter Caldwell, to Ingrid Planath: Referring back to the case of the failure of seat backs and the media interest in that, my understanding was that people have actually been killed because the body was allowed to then go into the back seat and either damage the person in the rear seat or get injured themselves. So I'm surprised that you say you're looking at collapsible seat backs. Maybe we're talking about different things?

Ingrid Planath: I just need to correct this — I do not propose that. It is a proposition from Peugeot Renault in a research paper where they said just that this might be favourable. We have not found reason to suggest such a design feature in the cars.

Chris Hall, to Ingrid Planath: It appears that one of the conditions in which seat belts also do not provide protection against neck injuries and spinal injuries is in rollover cases. The work by General Motors has in fact shown that a properly adjusted seat belt will enable an occupant to move up to 100 mm in the upside down direction, thereby enabling the occupant to strike the roof. Mackay has indicated that, for deformations of in excess of 6 inches, the seat belt in fact does not afford any protection to the belted occupant. Perhaps that's more an indication of the severity of the roll rather than the crushing mechanism causing the injury. I was wondering if Volvo had done any work on the rollover situation in relation to effectiveness of the seat belt?

Ingrid Planath: We have recognised this problem also because of the comparatively high occurrence of lumbar spine injuries in the rollover and we attribute this to the fact that in a rollover case one might not be that well restrained by the seat belt during the whole crash. We are looking into it and we want to define some kind of test method where we can evaluate this properly because the full scale rollover is not very repeatable. We certainly recognise the importance of this issue.

Chris Hall, to Rolf Eppinger: Can you comment on any work being done on the rollover case?

Rolf Eppinger: I'm trying to remember the specifics of it. We've been doing a variety of accident investigation and analysing the accident records for rollover with and without the presence of seat belts. At the moment I don't have the details but my recollection is that, from the study, the seat belt becomes fairly effective in rollover.

Chris Hall: Moffatt and others at General Motors made some comparisons between a roll-caged vehicle and the normal vehicle, and also compared the belted and unbelted occupants. Their conclusion was that the roll cage, in fact, offered no benefit.

Rolf Eppinger: They're entitled to their opinion!

MECHANISMS OF BRAIN INJURY IN PEDESTRIANS AND CAR OCCUPANTS

Tony Ryan

This talk will explain some of the mechanisms of brain injury and then describe the head injury research that has arisen from studies of pedestrians and car occupants carried out at the Unit.

Head injury includes soft tissue injury of the scalp, which will not be considered further, fractures of the skull, and injuries of the brain.

Brain injury can be focal or diffuse. *Focal injuries* include *contusions*, which is another name for bruises, and *lacerations*, which are cuts in the brain substance. There are also *haematomas*, or collections of blood, which are extradural if they lie outside the dura, a membrane which surrounds the brain substance, or subdural if they lie between the dura and the brain itself. Intracerebral haematomas occur within the brain substance. *Diffuse injuries* are spread throughout the brain. They may be *functional*, as in *concussion*, or *structural*, as in *diffuse axonal injury* or *diffuse vascular injury*.

The mechanisms of brain injury are summarised in Figure 1. If the duration of the impact is less than 200 ms it is considered a dynamic impact. Static impacts tend to be more like crushings, and will not be considered here. A direct impact to the head causes head motion and contact injuries at the point of impact. Head motions can also occur indirectly from impact to the body. Contact injuries include local skull bending due to the impact, where fracture of the skull occurs in tension so that if the skull bends inwards, the fractures appear on the inside and also on the outside of the skull and then propagate according to the local conditions of the skull at the time. This inbending will also cause changes in volume which can cause injury .

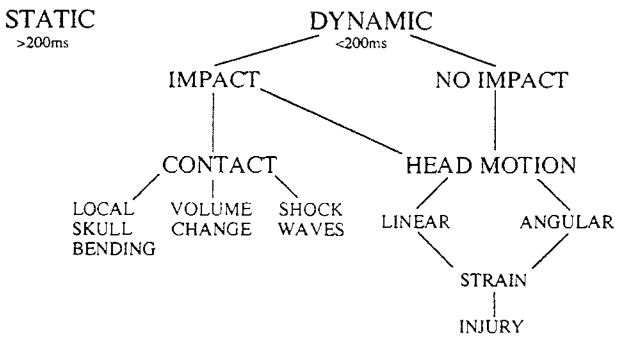


Figure 1: Mechanisms of brain injury

Movement of the head following an impact has both linear and angular components which collectively cause strains in the brain substance and hence injuries. Experiments by Thibault on isolated squid axons, which at about 0.5 mm diameter can be seen with the naked eye, making them easy to study, have shown that if the axons are stretched 5%, there is a reversible change in function; if they're stretched 10%, the change is irreversible; if they're stretched about 15-20%, the axon tears. It's thought that what happens in the brain is analogous to the squid axon, so that impacts to the head cause strains in the brain such, that, clinically, with a 5% strain one would get concussion and with higher degrees of strain there would be increasing levels of coma and then death.

In order to study how these effects interact in the mechanism of brain injury, Dr. Rémy Willinger from INRETS in France spent some time in the Unit last year. He drew together his work on mathematical modelling of the mechanical impedance of the brain and skull, with the Unit's observations of brain injury distribution. He suggested that if the head strikes a structure which has a natural frequency well above that of the skull, the brain and the skull become decoupled, move independently, and then brain injury tends to occur where the brain meets the more irregular bony parts of the skull — that is, in the frontal and temporal regions. When the natural frequency of the structure is around that of the brain and the skull — that is, around 100 Hz — then the brain and skull tend to move together and shear strains occur in the deeper parts of the brain, so that injuries tend to be observed in the central parts of the brain.

In 1943 a physicist called Holbourn deduced that as the contents of the brain were largely incompressible, translation would not cause brain injury but rotation would be very important. He postulated that the brain would move with a motion as shown by the arrows in Figure 2, with the largest movements at the surface and the smallest movements down towards the centre. He also made a gelatine model of a cross section of the brain and under polarised light showed that there were high strain areas below the falx and up in the parasagittal areas (Fig. 3).

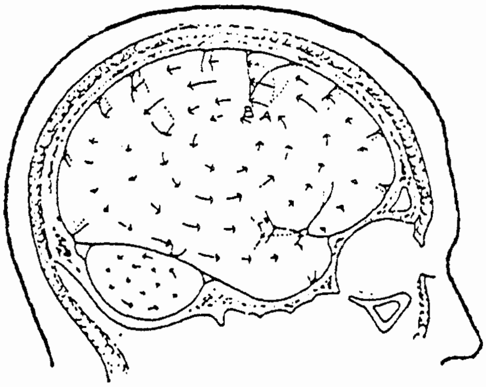


Figure 2: Brain contents movement postulated by Holbourn

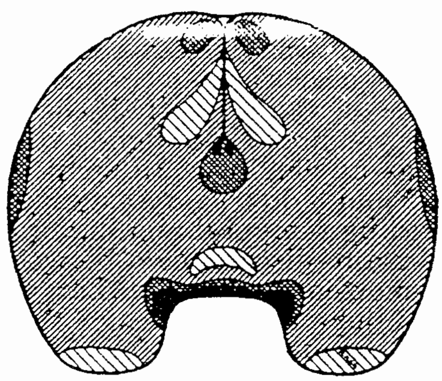


Figure 3: High strain areas in model of brain

In 1990, Thibault at the University of Pennsylvania, again using a model (Fig. 4), showed that when the skull rotated in a clockwise direction, the brain got left behind and then tended to strike the falx. Then when the skull stopped, the brain kept moving and hit the falx again. Thus the areas of increased strain were around the inferior edge of the falx and in the parasagittal region. In the present studies, observations of injuries occurring in brains in lateral impacts show that there are haemorrhages in the corpus callosum just below the edge of the falx and also up in the parasagittal area corresponding with the predicted areas of high strain. The injuries on the inferior aspect of the brain may well come from the collision between the brain and the skull itself.

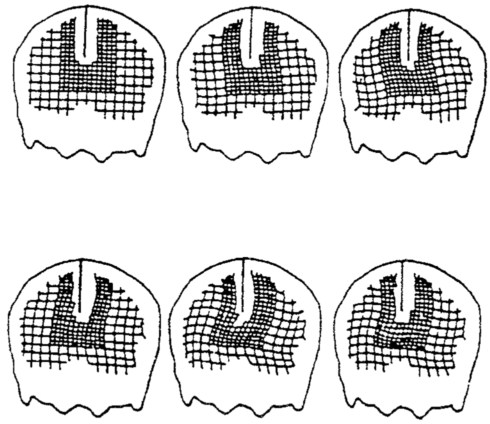


Figure 4: Thibault's human skull model deformation patterns

The head injury study at the NHMRC Road Accident Research Unit began in 1984. The aim was to look at the relationship between the location and the magnitude of the impact to the head and the distribution of injury to the brain. The study started with fatal pedestrian collisions: from the autopsy, the point of impact of the head was determined; at the site of the crash, the impact velocity was estimated; from inspection of the vehicle, the structure(s) hit by the head were identified; and then by pathological examination, the distribution of injury throughout the brain was determined. Impact velocity was estimated from the length of skid marks on the road. The position of dents on the bonnet or other structures on the car or of shattering of the windscreen showed where the head had hit. The position of the impact on the head was recorded on a diagram of a head using a clock face and a series of levels from the top downwards as shown in Figure 5. The brains were fixed for about 2 weeks in formalin and then sectioned and photographed by the neuropathologists.

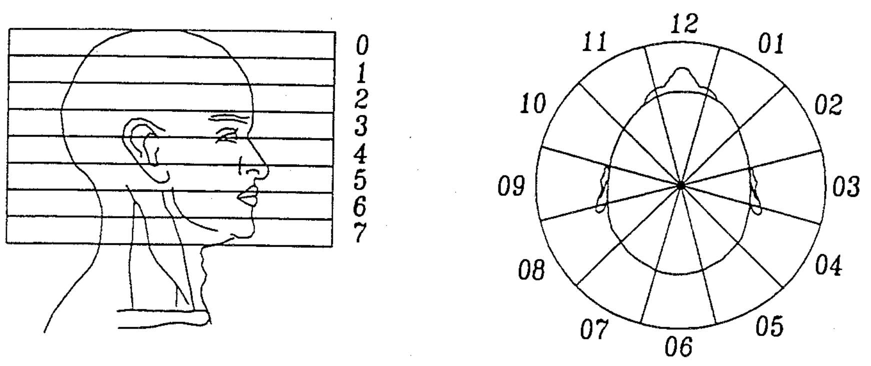


Figure 5: Head diagram used for positioning impacts

The method of recording damage to the brain is illustrated in Figure 6. The neuropathological information from each of the 11 cross sections of the brain is transferred to diagrams which then are put together to give a map of the distribution of injury to the brain. This summary map is called the 'banana diagram': Figure 6 shows the Mark 4 model — it just keeps evolving.

The peak linear acceleration of the impact is estimated by assuming that the head is a rigid body. It is also assumed that there is no sliding of the hip over the front of the bonnet, no tangential velocity at the impact and that the impact is at right angles to the skull. A combined stiffness of skull and vehicle structure is estimated and then peak linear acceleration is estimated using the equation given in Figure 7. This was derived by assuming that the kinetic energy of the head is all converted into strain energy in deforming the vehicle structure; this is reasonable if only the loading phase of the impact, when there is no rebound involved, is considered. As the mass of the head is known, the force can be determined and by estimation of the offset of the impact from the centre of mass of the head, an angular acceleration can be calculated. Comparison of estimates made with this method and measurements made in cadaver/vehicle impacts show that estimates of linear acceleration are within 30% of the measured value. Unfortunately there are no similar measurements available for comparison with our estimates of angular acceleration.

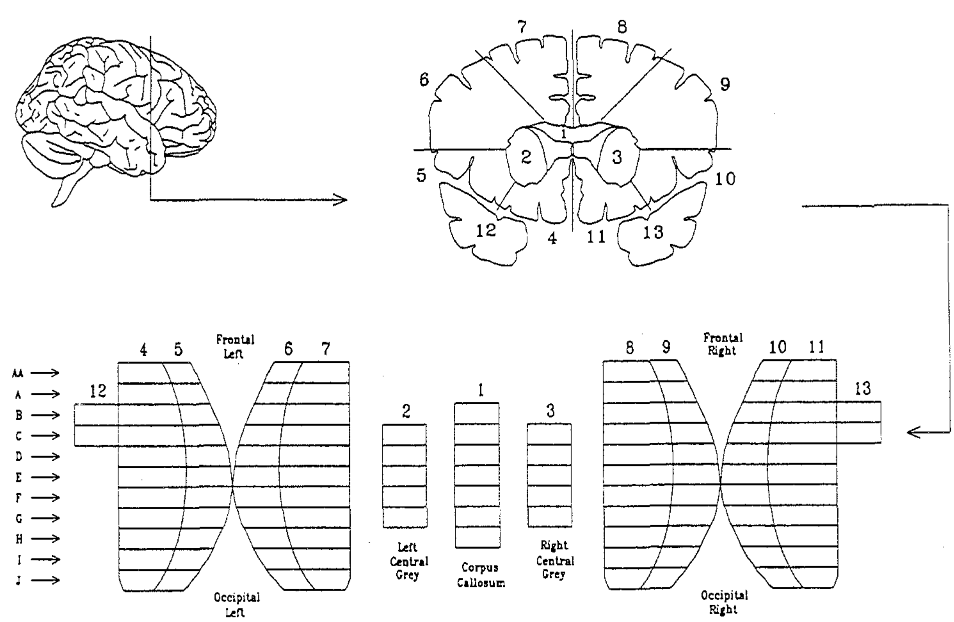


Figure 6: Diagrammatic method of recording brain damage

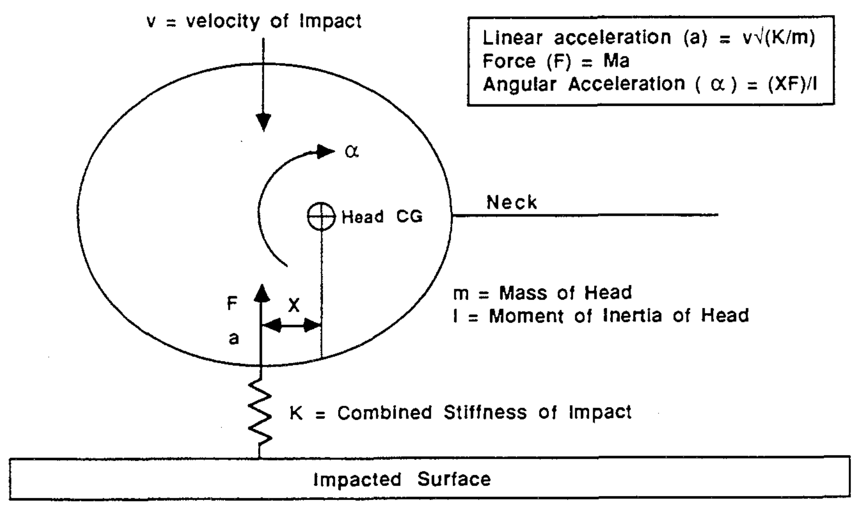


Figure 7: Head impact model for calculating accelerations to the head

Vehicle panels fell into 3 classes of stiffness as shown by the data below. These are distributed around the vehicle as shown in Figure 8. The A pillar was classed as the hardest part, the medium area included the edges of the bonnet and mudguard, while the middle of the bonnet, the windscreen and the roof were classed as soft.

Classes of stiffness of vehicle surfaces

Hard	=	600,000 N/m
Medium	=	350,000 N/m
Soft	=	140,000 N/m

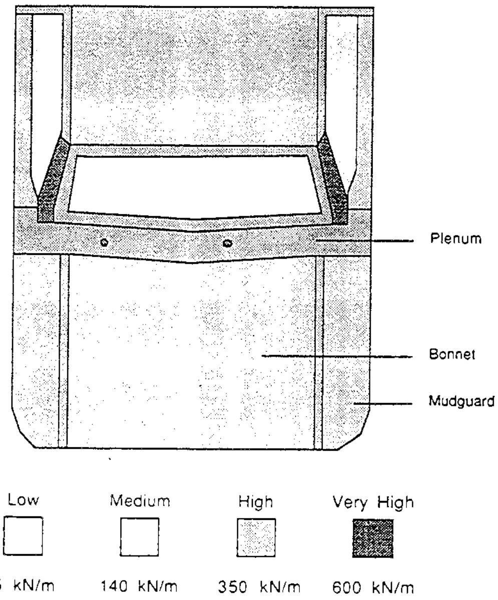


Figure 8: Distribution of stiffness of head impact areas on vehicle

The impact distribution (Fig. 9) was rather like that described by Koshiro Ono in that the adults tended to be higher up the car than the 2 children, but there were more impacts on the harder parts of the car, the A pillar and windscreen frame, than in the Japanese study.

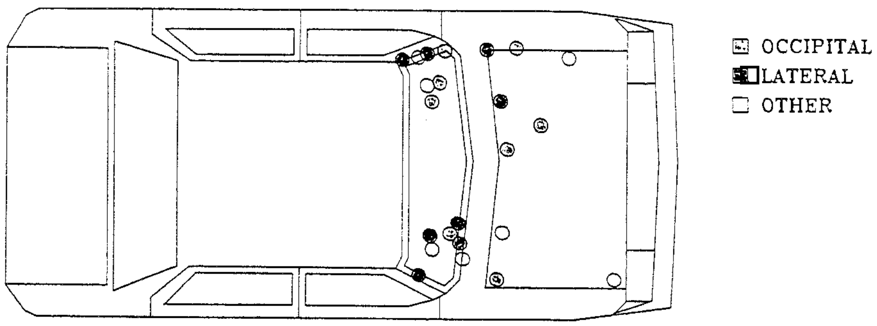


Figure 9: Head impact location on car and location of impact on head

Table 1 shows the results for 31 cases: 13 occipital and 18 lateral impacts. For the lateral impacts, the mean velocity was lower than for the occipital impacts, there were rather fewer impacts on medium or hard surfaces, and the linear acceleration was somewhat lower, although there was an overlap of the standard deviations for each variable. But the mean number of brain sectors injured, which is a measure of the degree of brain injury, is very similar for both types of impact. This suggests strongly that the brain is more sensitive to lateral impacts than to occipital impacts.

The position of the impact also determines the distribution of injury. The frequency distribution of injuries to the cortex and the white matter for lateral impacts is shown in Table 2. To determine these frequencies, the lateral impacts were all transposed so that they were represented as coming from the right side. It can be seen that the frequency of injury in the white matter in the right side (i.e. the side of impact) is significantly higher than that on the left side, whereas the cortical injuries are similar, right and left.

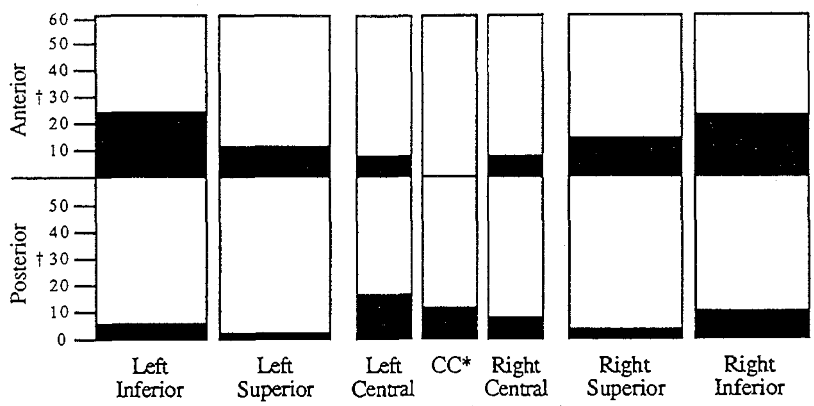
Table 1: Overall results for 31 cases

	OCCIPITAL	LATERAL
Number of cases	13	18
Sex (% male)	76.9	77.8
Age (% 50+ years)	53.8	55.6
Head impact velocity (km/h, mean ± sd)	64.2 ± 12.1	53.3 ±13.1
Stiffness (% medium or hard)	76.9	55.5
Peak linear acceleration (m/s^2, mean ± sd)	4984 ± 1988	3702 ± 1583
No. brain sectors with injury (mean ± sd)	31.3 ± 31.6	32.7 ± 27.8
Skull (% with fractures)	53.9	55.6
Cervical spine (% with fractures)	30.8	33.3
Heart and great vessels (% with injuries)	30.8	33.3

Table 2: Right lateral impacts
% of brain sectors with injury, for 18 pedestrians

TYPE OF INJURY	BRAIN REGIONS								
	Left					Right			
	Inferior	Superior	Para-sagittal	Centre	Corpus callosum	Centre	Para-sagittal	Superior	Inferior
Cortical	26.5%	15.6	12.2	13.2		18.9	16.7	11.2	26.2
White matter	11.8	10.3	8.3	7.5	41.9	8.5	18.3	14.2	13.1
Number of sectors	396	180	180	106	124	106	180	180	396

As well as the location of impact, the magnitude of the impact has an influence on the distribution of injury in the brain. The results in Figures 10 and 11 show that, for occipital impacts under 5,000 m/s^2, the injury was found up in the frontal half of the brain, but for impacts over 5,000 m/s^2, the injury is spread throughout the brain. This difference was not found so clearly with lateral impacts.



* Corpus callosum † percentage of sectors with injury

Figure 10: Brain injury distribution — occipital impacts < 5000 m/s/s

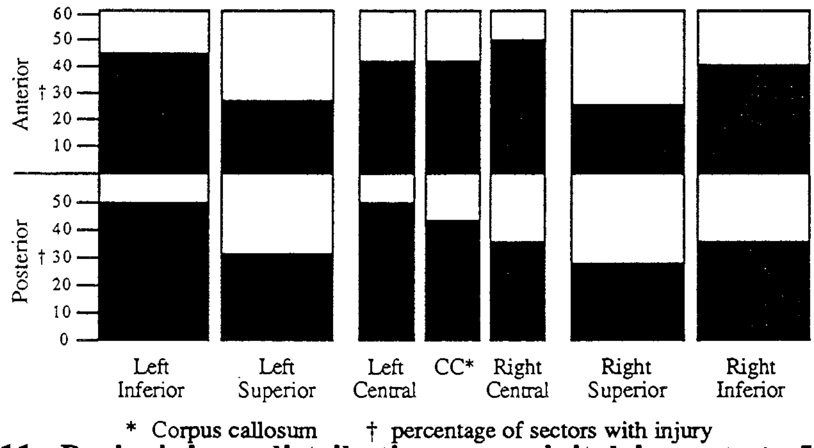


Figure 11: Brain injury distribution — occipital impacts ≥ 5000 m/s/s

Also, when peak acceleration was plotted against the number of injured brain sectors, a linear, positive association was seen for occipital impacts, but when the same thing was done for lateral impacts there was no association at all (Figs. 12, 13).

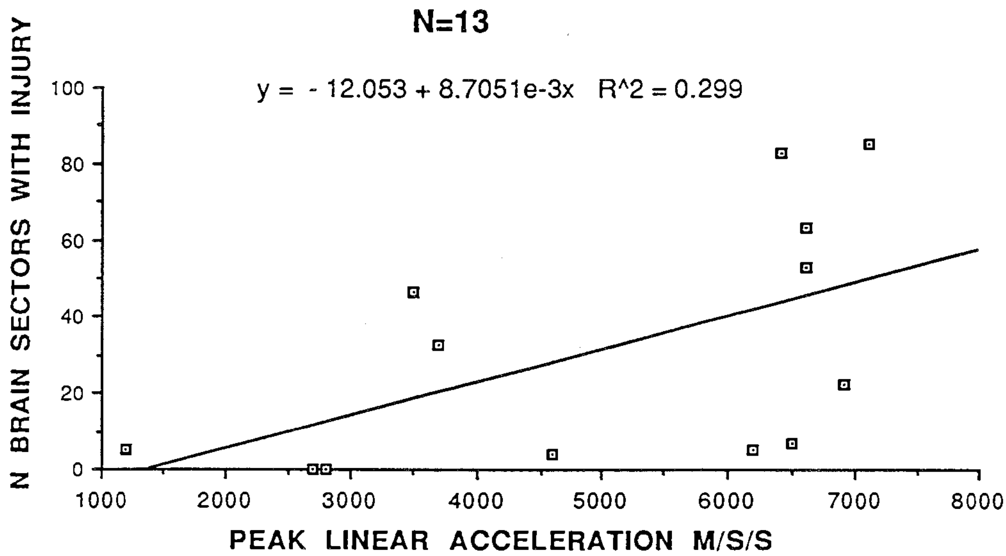


Figure 12: Pedestrian occipital head impacts

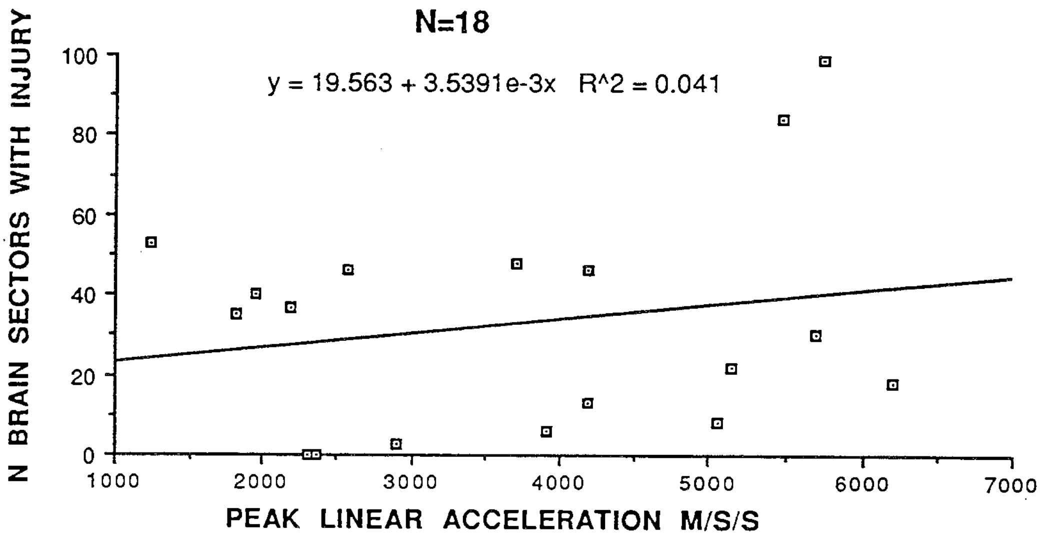


Figure 13: Pedestrian lateral head impacts

The same basic techniques were used in studying car occupants, in cases where there was severe or fatal injury within about a 100 km radius of Adelaide. Twenty two crashes were included in the study and amongst the occupants there were 19 fatal cases and 7 non-fatal cases. For the non-fatal cases, the neuropathology cannot be obtained directly, so CT (Computerised Tomograph) and MRI (Magnetic Resonance Imaging) scans are used instead. These scans, particularly the MRI, can be used as a good substitute for neuropathology in lesions of about 5 mm diameter or greater.

In 17 of the 26 cases it was possible to determine what object the head had hit, and in most cases the objects were those parts of the car around the occupants at head height (Table 3); there were a few intruding objects as well. Of these 17 cases, 9 were frontal impacts and 6 were lateral impacts. Table 4 shows that 5 of the lateral impacts were fatal as were 4 of the frontal impacts. Also there were more skull fractures in the lateral cases than in the frontal cases. The range of accelerations was very similar for all of the groups. So again there is some evidence that the brain is more sensitive to lateral impacts than, in this case, to frontal impacts.

Table 3: Objects struck in 17 head impacts (car occupants)

Object	N
Roof side rail	5
Door and/or glass	3
B-pillar	3
Header	3
Steering wheel hub	1
Steel and concrete pole	1
Bonnet of striking car	1

Table 4: Impact location, accelerations and brain injury for fatal and non-fatal cases

Impact Location	NON FATAL Case No	NON FATAL Acceleration Linear m/s^2	NON FATAL Acceleration Angular $rads/s^2$	NON FATAL COHL*	FATAL Case No	FATAL Acceleration Linear m/s^2	FATAL Acceleration Angular $rads/s^2$	FATAL COHL
FRONTAL	08.1	1000	15000	6 +	17.2	1800	14000	0 +
	17.1	1800	14000	0	11.1	2900	22000	0
	19.2	2300	18000	0	22.1	3900	19000	10 +
	23.2	4300	30000	4	06.1	4400	33000	0
	09.1	5800	43000	5				
LATERAL	03.1	2800	21000	10	12.1	1900	19000	0 +
					19.1	2000	6000	0 +
					07.1	3000	23000	39
					15.1	4800	24000	97 +
					05.2	6600	50000	67 +
OCCIPITAL					18.1	2900	22000	12 +
					05.1	3300	16000	31

* Cumulative occurrence of haemorrhagic lesions
\+ Skull fracture

The distribution of injuries within the brain is illustrated diagrammatically in Figures 14 and 15 which show that there were fewer injuries in cases of frontal impacts compared with the lateral impacts. In the latter, injuries were more severe and more widespread — and again with roughly the same linear accelerations as the frontal group.

When the numbers of sectors with brain injury were plotted against linear acceleration for frontal impacts, there was no association. However, for lateral impacts there was a positive association (Fig. 16). This is different from the result for lateral impacts in the case of pedestrians.

EXAMPLES OF STUDIES ON HUMAN HEAD IMPACT TOLERANCE and CURRENT SITUATION OF EXISTING HEAD IMPACT TOLERANCE THRESHOLD

Koshiro Ono

The incidence of head injuries in automobile accidents has become high and, in many cases, the injuries tend to be very severe or fatal. The clarification of the mechanism of head injuries on impact is, therefore, a crucial task in trying to find appropriate measures for modification of the objects that human beings are likely to collide against, to reduce the severity of human injuries.

It is said that the occurrence of head injuries depends on various impact conditions, but the mechanism has not been sufficiently clarified. It is particularly important to find answers for the questions:

- what is the nature of head injuries, and what kinds of mechanisms and observations are important;
- what are important parameters for researchers to measure in the experimental environment. That is, what kinds of impact conditions would cause head injuries; and
- what are the types of head injuries that occur, which types are important causes of death and disability and conversely, which types are more regularly associated with a favourable outcome.

In this regard, this talk will present the findings obtained from the Japan Automobile Research Institute (JARI) study on the human head impact tolerance, using sub-human primates and human cadaver skulls. Also introduced will be some examples related to the questions of how to interpret mechanisms of head injuries currently being proposed and to what extent the head injury criteria can be applied, based on relevant studies of the JARI findings.

The parameters that were set for the head impact conditions were aimed primarily at clarification of several issues. These were: how injuries would occur, with what degree of severity, on which regions of the skull or brain, and with what type of pathophysiological occurrences; also what type of forces act on the skull or brain, and how these factors could be related to such forces.

The experiments were classified into head impact tests using subhuman primates, and free-drop impact tests using human cadaver skulls. Impacts were applied to 3 contact areas on the head in each test - that is, frontal, occipital and temporal regions.

Primate Experiments

Table 1 shows the grouping of monkeys used in the sub-human primate experiments according to the combination of impact conditions, and the number of monkeys used in each test.
For sagittal impacts:

- Group A = experiments under the translational impact condition;
- Group B = experiments under the rotational impact condition;
- Group C = experiments under the combined condition of translational and rotational impact.

For lateral impacts:

- Group D = similar to the condition of the Group C experiments under the combined condition of translational and rotational impact.

Table 1: Impact conditions and subjects in the classified experiments

Grouping	Impact Condition: Impact Velocity (m/s)	Impact Condition: Component of Head Motion	Impact Condition: Contact Area	Location of Impact	Subject: Species	Subject: Age (average)	Subject: Body Weight (Kg) (average)	Sex: M	Sex: F
Translational Impact (Group A)	27 <	Translation	Mask (Board)	Frontal	Macaca fuscata	5.7	8.2	17	3
					Papio cynocephalus	20	25	1	0
				Occipital	Macaca fuscata	7.8	7.7	2	2
					Papio cynocephalus	9	35.8	2	.0
Rotational Impact (Group B)	12 ~ 28	Translation + Rotation	Mask (Board)	Occipital	Macaca fuscata	10.1	7.9	5	4
					Macaca fasicularis	6	8.1	8	0
					Macaca mulatta	6	9	1	0
Direct Impact (Group C)	5 ~ 15	Translation + Rotation	Padded flat surface (narrow)	Frontal	Macaca fuscata	12	8.8	1	1
					Macaca mulatta	6.7	7.1	3	4
				Occipital	Macaca fuscata	6.8	6.9	0	5
					Macaca mulatta	5.8	6.1	1	3
Direct Impact (Group D)	9 ~ 36	Translation + Rotation	Padded flat surface (narrow)	Laterale	Macaca fuscata	3 ~ 9	10.0	6	1
					Macaca mulatta		9.0	11	1
					Macaca fasicularis		6.4	4	0

Figure 1 shows the impact apparatus used in Group A. It is a newly developed double-sled system, with a slider mounted on a HYGE sled. When the sled is launched, the head mask and the slider travel along the slider guide and collide against the lead block. On collision, the pin impinges on the lead block and the impact is transmitted to the head of the subject through the slider and mask (Fig. 2).

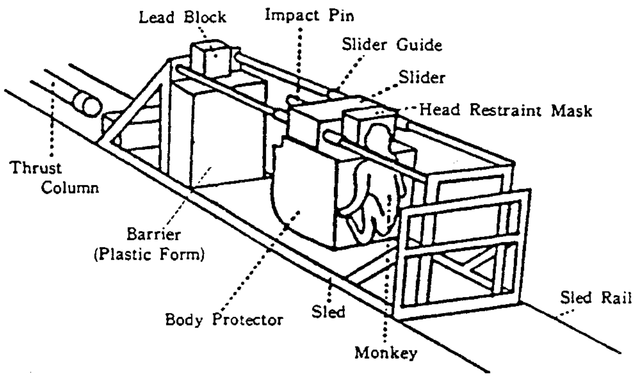
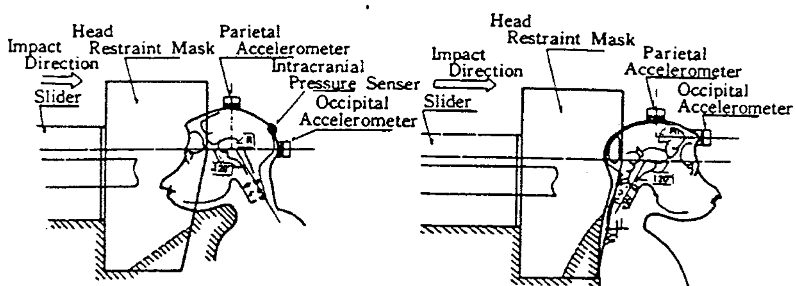


Figure 1: Overall view of experimental set-up of newly developed impact system on HYGE sled



(a) Frontal impact (b) Occipital impact

Figure 2: Translational head impact test configuration and set-up for masking of subject's head

Figure 3 shows the outline of the rotational impact system used in Group B. This system consists of a chair device to which a monkey is strapped, and a compressed air type impactor-ejector which delivers a blow to the monkey's head. The monkey is seated on the chair, and its head is fixed by means of the head impact mask made of plaster of Paris (Fig. 4). When the impact is applied, the head mask rotates around the neck. The rotation is then transmitted to the body, which in turn pivots around the hips.

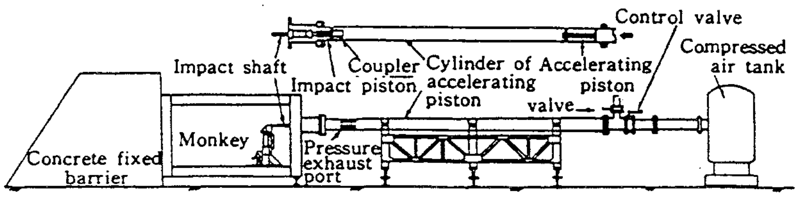


Figure 3: Illustration of the impactor system for rotational and direct head impact

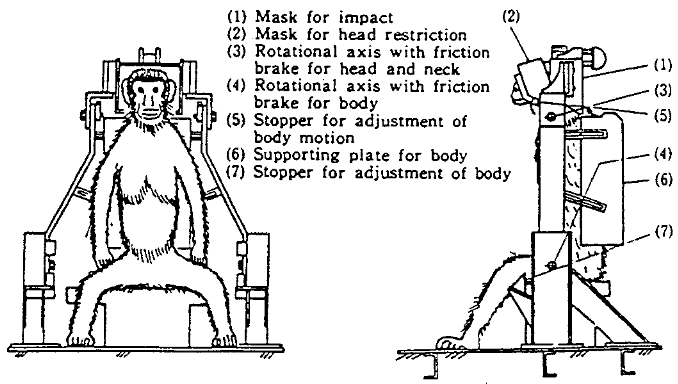


Figure 4: Rotational head impact test configuration and set-up for masking of subject's head

The setting of the subject for Group C is shown in Figure 5. The impactor-ejector is the same as in Group B, except that the impactor head is covered with a rubber block designed to accommodate the entire head of the monkey. The impactor causes a direct impact, either on the front or back of the monkey's head.

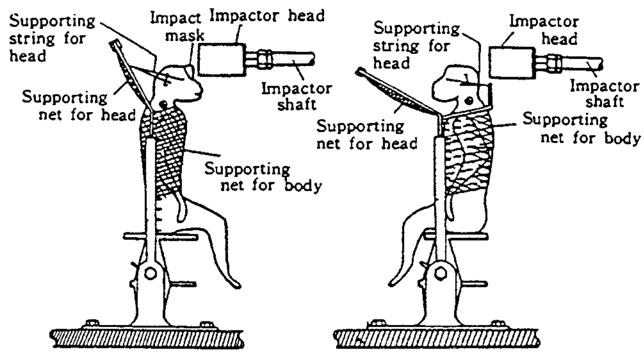


Figure 5: Set-up restraint system

Figure 6 shows the setting of the subject on lateral impact for Group D. The restraining device and impactor-ejector are the same as in Group C. In every test, each monkey is seated in a lateral sitting position on the chair, and an impact is delivered onto the head at the upper region of the lateral zygomatic arch.

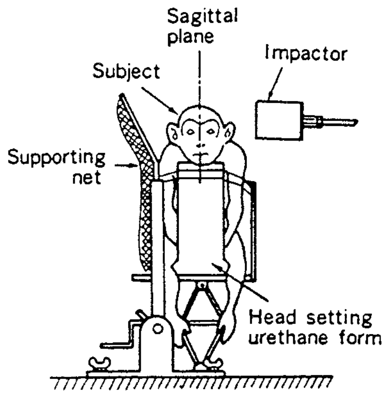


Figure 6: Subject restraint device

Physical measurements and biological observations made in the tests are set out in Tables 2 and 3. Each subject was kept under anaesthesia for 2 to 6 hours or so, from the preparation period to the application of impact. During this period, measurements of the body are taken, and various sensors are placed in position. Subjects are X-rayed and set on the chair according to each impact method. The neurological status is then examined 5 to 20 minutes after each impact. If brain concussion or some other injury is not detected, the subject receives another impact after a rest of 60 to 120 minutes.

Table 2: Physical measurements

Grouping \ Items	Velocity	Acceleration	Intracranial pressure	Motion	Dimensions recorded by X-ray
Group A	Sled	Head (X, Z axes) Slider (X axis) Sled (X axis)	Epidural space	Head (4000 f/s) Head and body (2000 f/s)	Location and direction of the attached accelerometers Skull diameter Thickness of skull bone
Group B	Impactor Head Mask	Head (X, Y, Z axes ; 9 ch) Head Mask	–	Head (4000 f/s) Head and body (2000 f/s)	Location and direction of the attached accelerometers Skull diameter Thickness of skull bone
Group C	Impactor	Head (X, Y, Z, axes ; 9 ch)	–	Head (4000 f/s) Head and body (2000 f/s)	Location and direction of the attached accelerometers Skull diameter Thickness of skull bone
Group D	Impactor	Head (X, Y, Z axes ; 9 ch)	–	Head (4000 f/s) Head and body (2000 f/s)	Location and direction of the attached accelerometers Skull diameter Thickness of skull bone

Table 3: Biological observations

1) Physiological survey
   Vital sings (respiration, pulse rate, blood pressure, ECG)
   EEG
2) Biochemical survey
   Blood gas analysis (PO_2, PCO_2, PH)
3) Neurological survey
   Eye movement, blinking, pupillary response
   Light reflex, corneal reflex, behavioral response to pain
   Oculocephalic response
4) Pathological survey
   Autophy examination
   Optical microscopic or electron microscopic examination

The severity of concussion is judged according to the following signs detected immediately after each impact:

- loss of the corneal reflex persisting for at least 20 seconds;
- apnoea persisting for at least 20 seconds; and
- moderate or severe abnormality in the blood pressure (Fig. 7)

The severity of brain concussion is classified into 4 grades, 0 to III according to the presence of none, one, 2 or all 3 of these signs respectively.

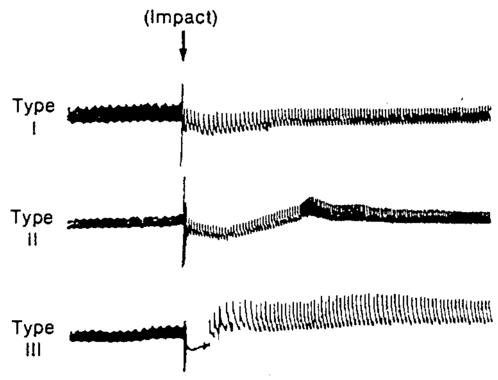


Figure 7: Pattern of blood pressure after concussion

The 4 groups in these experiments comprised 86 subjects and results were obtained for 60 frontal, 59 occipital and 75 temporal impacts.

In the Group A experiments, concussion of Grade II or Grade III severity occurred in all subjects on the first impact. Among the survivors, concussion lasted for between 20 sec and 7 min. An example of the time history of the monkey's head acceleration, and the intracranial pressure on application of frontal impact is shown in Figure 8. Although the intracranial pressure of -1 kg/cm^2 developed as shown, only slight subarachnoid haemorrhage was seen on visual examination.

In Group B, 12 of the 18 subjects survived and 5 of the 6 deaths were instantaneous. Four of the fatal cases developed comminuted depressed fractures. Of the subjects in this group, 94% developed Grade III concussion with one exceptional case of Grade II concussion; 67% of the cases also developed brain contusions. Pathological findings were significant in this group. Brain laceration and contusion were found in all 6 fracture cases. Of the 12 non-fracture cases, 7 developed brain contusion in the parasagittal region, the tip of the frontal or temporal lobe or the brain stem.

Group C had 18 subjects, and half of them were subjected to a total of 39 frontal impacts. Of the 18 subjects, 4 had skull fractures and of these 3 died. According to the neurophysiological findings, concussions of all grades occurred, that is from Grade 0 to III, due to frontal or occipital impact. No brain contusions were found in the parasagittal region, but contusions were found in the basal surface, ellipse of the frontal lobe, the occipital lobe and the tip of the temporal lobe. Brain stem injuries accompanied by a deep-seated haemorrhage were seen in one fracture case and 4 non-fracture cases. Of the total of 8 fatal cases, 4 had cervical injuries without fractures.

For the Group D experiments, 23 monkeys were used, and subjected to a total of 75 impacts. Of the 23 monkeys, there were 18 non-fracture cases and only one fatal case. Pathological changes were observed in 13 cases of the 18 non-fracture subjects, as shown in Table 4.

Characteristic features compared with sagittal impacts were:

- the higher incidence rate of intermediary contusions, such as parasagittal subcortical haemorrhage and corpus callosum haemorrhage;
- the tendency in which the higher the concussion severity, the higher the incidence rate of intermediary contusions; and
- corpus callosum haemorrhage was found with mild concussion such as Grade 1, though the incidence rate was low.

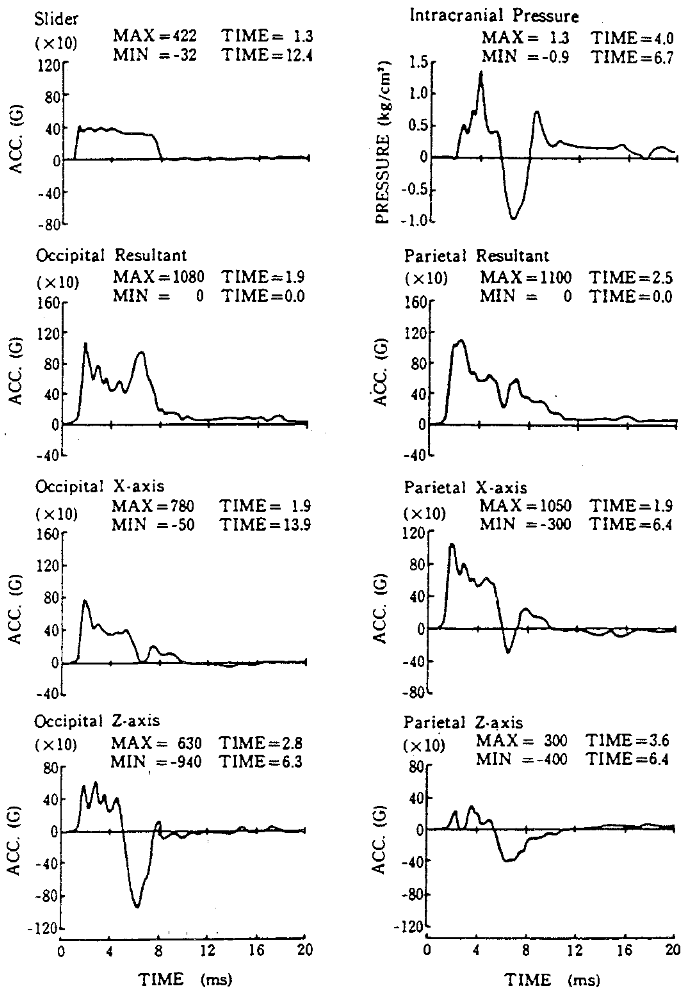


Figure 8: Head acceleration and intracranial pressure-time history in frontal impact

Table 4: Pathological changes in monkeys without skull fracture in lateral impact

monkey No. / Patho. brain injury	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18
Severity of concussion	0	I	III	II	III	II	I	II	III	I	II	III	I	I	II	0	I	0
Severity of pathological injury	0	II	III	II	III	II	III	0	II	0	III	II	0	II	II	0	II	II
Subarachnoid hemorrhage			I.L.	C.L.	B.L.		B.L.		I.L.		B.L.	B.L.			B.L.		C.L.	
Cortical hemorrhage		I.L.	I.L.		B.L.	I.L.			I.L.						I.L.		C.L.	C.L.
Subcortical hemorrhage			B.L.	I.L.	I.L.	C.L.			I.L.		I.L.				I.L.			C.L.
Callosal hemorrhage			I.L.		B.L.	B.L.			B.L.		B.L.			B.L.	B.L.			B.L.
Hemorrhage of thalamus or basal ganglia											I.L							
Brain stem hemorrhage			B.L.		B.L.		I.L.				B.L.							
Subependymal hemorrhage			B.L.		I.L.													
Subdural hemorrhage			B.L.		B.L.						C.L.	C.L.		C.L.				

I.L. : ipsilateral to the impact side, C.L. : contralateral to the impact side, B.L. : bilateral

Human Skull Experiments

In the experiments concerning the occurrence threshold of the human skull fracture under the frontal, occipital, and lateral impact conditions respectively, a total of 25 dried human cadaver skulls were used. To approximate the actual conditions of human heads, the weight of each skull was increased by filling the intracranium with a gelatine solution as a first step. Then, the outer surface was plastered with clay, and the skull was covered with a rubber skin of the Hybrid II dummy head. As a result, the weight increased to 5 kg. The skulls were suspended along a line passing through the impact point and the centre of gravity, in the experimental apparatus which is shown in Figure 9. Skull accelerations were measured by three mounted accelerometers (X,Y,Z) between the basal vomer and the foramen magnum. Fractures were classified by degrees into: rudimentary fractures; linear or radiating fractures; and multi-linear comminuted fractures. In these experiments, 42 frontal impacts, 36 occipital impacts and 58 temporal impacts were delivered to the skull.

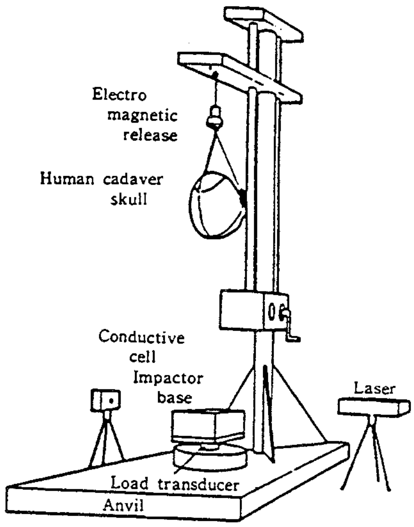


Figure 9: Overall view of free drop test on human cadaver skull

In cases where skull fractures occurred, the basic impact wave of skull accelerations showed spikewise, higher-harmonic, damped oscillations (Fig. 10). This phenomenon was seen in all fracture cases, including the basal fractures. Therefore fractures were detected by visual examination and changes in pattern in the time history of the skull acceleration. This finding confirmed a distinct boundary between fracture and non-fracture groups, as seen in Figure 11 and determined the frontal fracture threshold.

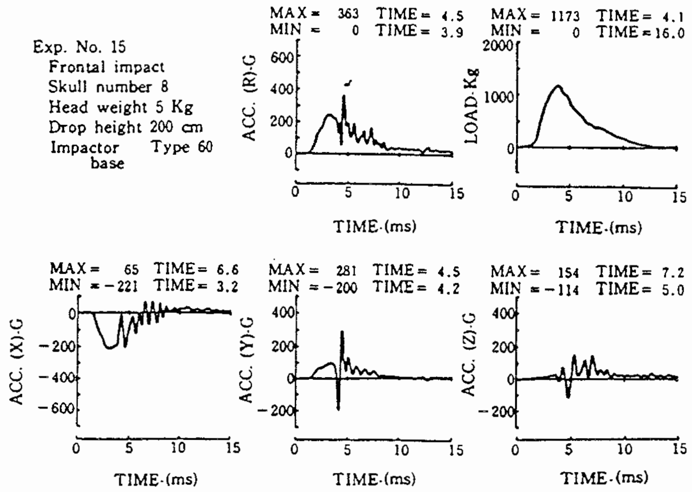


Figure 10: Skull acceleration and load from frontal impact on human cadaver skull

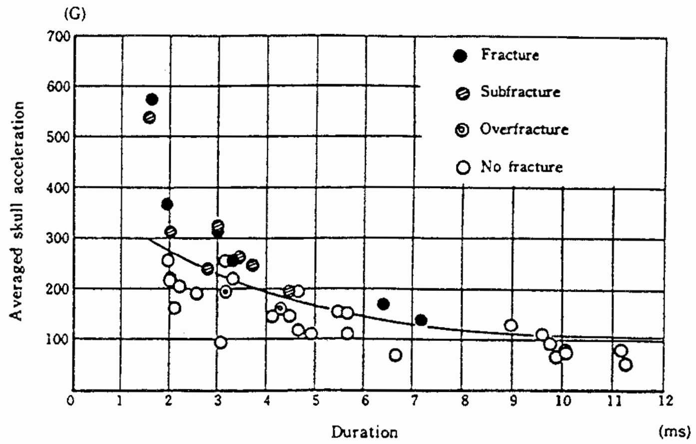


Figure 11: Correlation between averaged acceleration-duration and skull fracture in frontal impact of human cadaver skulls

According to a similar classification of fractures, the occipital fracture threshold (Fig. 12) is similar to that of the frontal impact. However, as Figure 13 shows, the temporal fracture threshold shows a value of 120 g or so around 3 ms, and a value of 90 g or so around 6 ms, which are approximately 50% of the fracture threshold on frontal and occipital impacts, respectively.

The relationships between head impact acceleration and head injury, the incidence of brain contusions and concussion in particular, were examined. The pathological injuries found in the series of monkey experiments, were classified according to clinical experiences, as was shown in Table 4. It is clear that Grade I injuries were on the brain surfaces, Grade II injuries were in the brain substance and Grade III injuries were in the brain stem.

The tendency of pathological injury incidence of Group D is that the distribution of pathological injuries can be identified clearly as shown in Figure 14, if the Grade II cortical contusion and/or subcortical haemorrhage are used as the reference. The estimation of the distribution for Group C based on the distribution in Group D is nearly approximate to that of Group D. Consequently, it can be said that the incidence of pathological injuries based on those of the Grade II is nearly the same for sagittal and lateral impacts under impact conditions of Groups C and D.

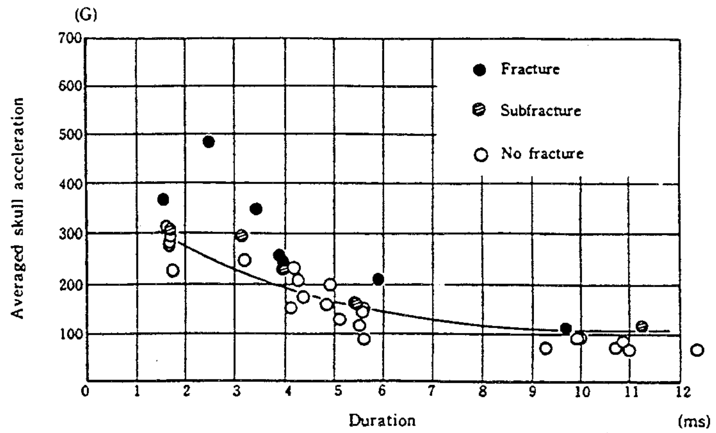


Figure 12: Correlation between averaged acceleration-duration and skull fracture in occipital impact of human cadaver skulls

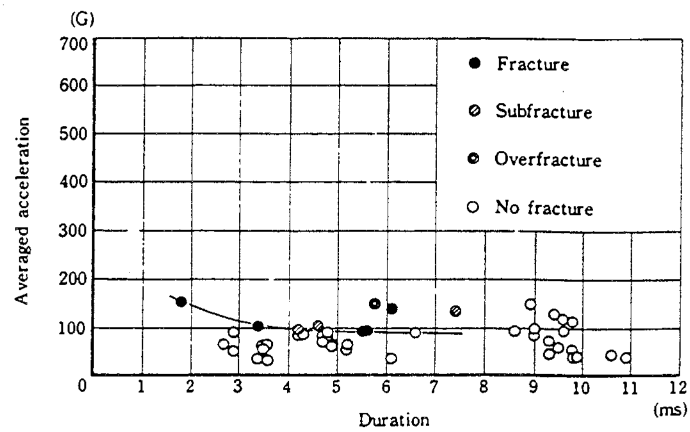


Figure 13: Correlation between averaged acceleration-duration and skull fracture in lateral impact of human cadaver skulls

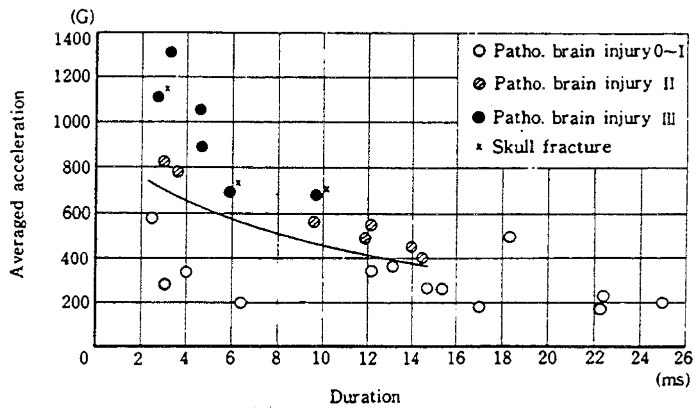


Figure 14: Correlation between averaged head acceleration-duration and grades of pathological brain injury in lateral impact (Group D); on the pathological brain injury severity 0, data of all impacts other than final impact of same subject were contained

Next, on the relationship between the head acceleration and concussion, Figure 15 shows the comparison of the head acceleration-duration and the degree of concussion among Groups A, B and C. In Groups A and B, the degree of concussion was in the range of Grades II to III or more severe in every case. Pathological findings did not necessarily show correlation with the acceleration-duration. In this regard, neurophysiological findings are also required in addition to pathological findings, as vital indices in finding head impact tolerances under various impact conditions.

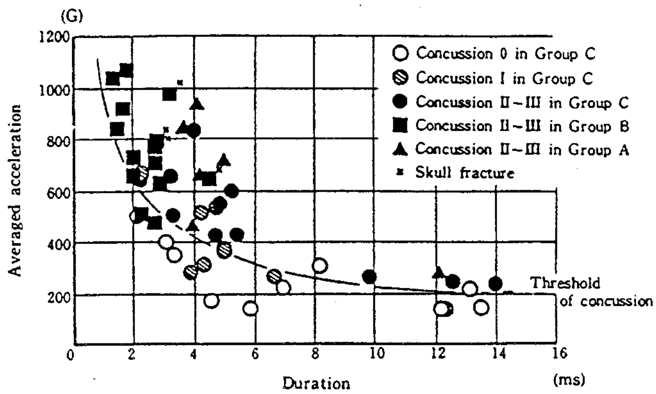


Figure 15: Comparison between averaged acceleration-duration of the monkey head and grades of concussion in occipital impact among 3 groups (A,B,C)

Since the concussion threshold against lateral impact was found from this study, the threshold could be compared with that of sagittal impact. Figure 16 shows the concussion threshold curve of lateral impact. The entire curve is located twice as high above the curve of sagittal impact. It may therefore be concluded that concussion tends to occur more often by sagittal impact than by lateral impact.

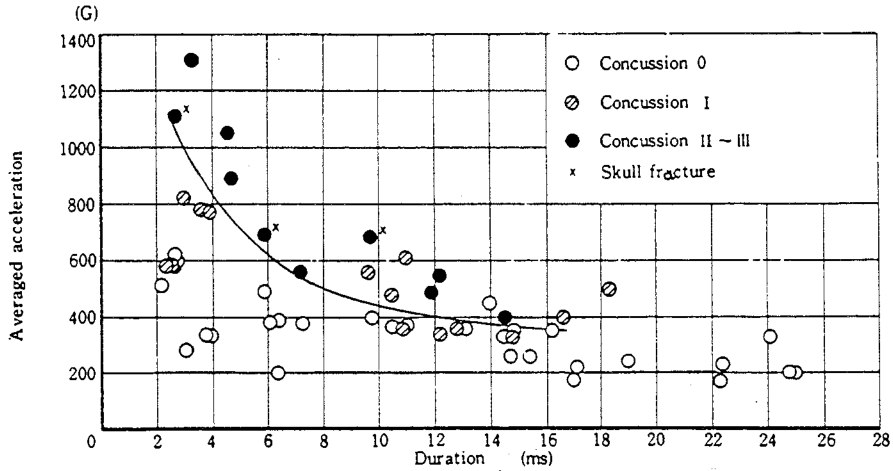


Figure 16: Correlation between averaged head acceleration-duration and grades of concussion in lateral impact (Group D)

Figure 17 shows the human head impact tolerance curve deduced by applying the dimensional analysis method proposed by Stalnaker to the concussion threshold determined by the series of sagittal impact experiments on monkeys, which was combined with the fracture threshold curve obtained by the series of experiments using human cadaver skulls. Since this curve was already reported as the JARI Human Head Tolerance Curve at the 24th Stapp Conference, the details of the process of estimation using dimensional analysis will not be included here. It has been confirmed that the criterion employed in the Wayne State Tolerance Curve is nearly the same as the concussion threshold level of the JARI Human Head Tolerance Curve.

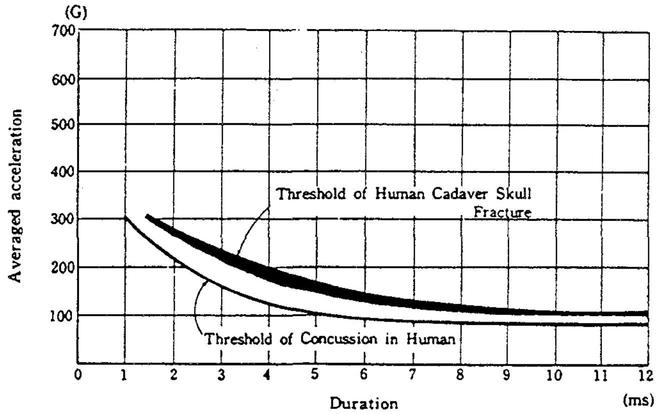


Figure 17: JARI human head tolerance curve for sagittal impact

Background

The foregoing was an example of a specific study done for the investigation of human head tolerance thresholds to impacts. The background to the study being carried out anew is reviewed below.

At a time prior to the initiation of this study, indices and thresholds on head impact tolerance shown in Figure 18 were proposed.

SI	HIC	JTI	RBM	EDI	MSC	IHI
Severity Index (GADD)	Head Injury Criterion (Versace & Nhtsa)	J-Tolerance Index (Slattenschek)	Revised Brain Model (FAN)	Effective Displacement Index (BRINN)	Mean Strain Criterion (STALNAKER)	Injury Hazard Index (FURUSHO)
$SI=\int_0^T (a(t))^{2.5}dt$ Time in seconds Acc. in g-units	$\bar{a}_{12}=\frac{\int_{t_1}^{t_2} a(t)\,dt}{(t_2-t_1)}$ $\left\{\max_{t_1,t_2}\left[\bar{a}_{12}^{2.5}(t_2-t_1)\right]\right\}$ $0<t_1<t_2<T$	$a(t)$, K, m, C, $X(t)$ $W_n=\sqrt{K/m}$ (rad/sec) $\beta=C/C_c$ $W_n=635$ $\beta=1.0$	$a(t)$, K, m, C, $X(t)$ $W_n=175$ $\beta=0.4$	K, m, C, $X(t)$ $W_n=482$ $\beta=0.707$	$a(t)$, K, m_1, m_2, C, $X(t)$ $m_1=0.6$ (lbs) $m_2=10.0$ (lbs) $C=2.0$ (lb sec in) $K=50000$ (lb in)	$IHI=\frac{\left(\frac{1}{t_2-t_1}\int_{t_1}^{t_2}\alpha^n(t)\,dt\right)^{\frac{1}{n}}}{f(\tau)}\times 1000$ $\alpha(t)$= Acc. (g) t_1, t_2 =Duration Time $f(\tau)$=Tolerance Curve n:Weighted Power
$SI_{toL}=1000$	$HIC_{toL}=1000$	$J=\frac{X_{max}}{0.0925}\ \frac{in}{in}$ $J_{toL}=1.0$	$T<20$ms: $\dot{X}_{toL}=135.3\frac{in}{sec}$ / $T<20$ms: $X_{toL}=1.25$in	$X_{toL}=$ A-P / RES; Human 0.15in / 0.18in; Dummy 0.17in / 0.2in	$\varepsilon=X_{max}/L$ 人間: $L=5.75$in (A-P) $\varepsilon_{toL}=0.0061$ in/in	GSI 1200 HIC 1350

Figure 18: Summary of head injury criteria

All of these criteria were based on hypotheses for mechanisms of head injury occurrence. Some are still under study, but the Tolerance Curve proposed by workers at Wayne State University (Fig. 19) showed considerable outcomes obtained through steady efforts made from biomedical and engineering viewpoints. Nevertheless, some unclear points remained despite various attempts made to identify the rationales of the curve. For example, questions remained as to:

- the locations where the head accelerometers should be installed;
- the rationale for the calculation of 'effective acceleration';
- the type of head injury constituting the basis of the criteria; and
- the type of impact conditions considered for the curve, etc.

These questions appeared to have been discussed for a considerable length of time during the STAPP Conference, but no definite answers were found.

Because of this background, it was concluded that a further study should be carried out in search of the answers to those questions, in order to allow or assist in determining realistic and practical head impact tolerance criteria. At JARI the head injury forms were simplified by assuming various impact conditions, and they were classified by patterns as shown in Figure 20. According to this classification, the series of experimental studies described earlier was planned and implemented.

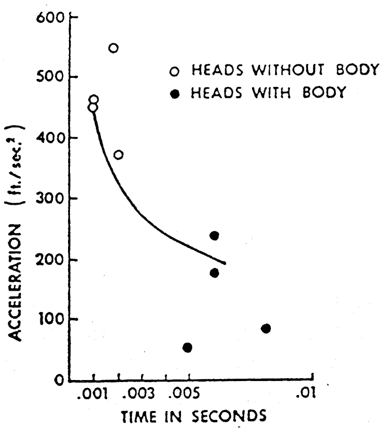


Figure 19: The original Wayne State curve

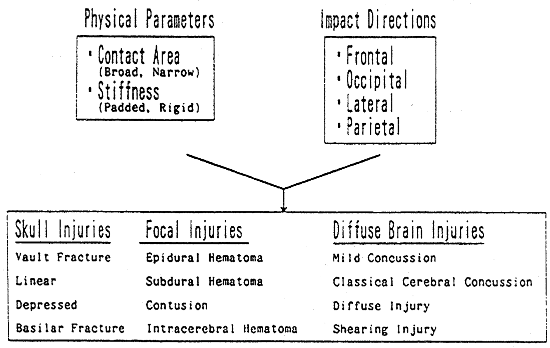


Figure 20: JARI classification of head injury forms

When considering appropriate indices for human head impact tolerances based on the findings described so far, it appears that no other indices can replace the curve developed by the Wayne State University and the JARI Human Head Tolerance Curve at present. It may also be said that no other criteria are available at present that can replace the Head Impact Criterion (HIC) currently used as the criterion for various types of impacts.

Needless to say, more detailed studies have recently been carried out on the mechanisms that generate head injuries. For example, studies on the impact-injury levels in boxing games with relatively low impact levels, or the finite element modelling of skull-brain injuries and the numerical simulations done under the initiative of NHTSA. Therefore, more advanced practical head impact tolerance and injury criteria capable of covering various types of impact patterns - that is, those capable of evaluating various impact tests - may be proposed and used in practice in the near future.

In conclusion, some comments should be made on the research activities using primates which have been reported here. This research was subjected to severe criticisms from the Society for the Prevention of Cruelty to Animals, and the implementation of such studies is becoming extremely difficult, if not impossible, in Japan as elsewhere. The fact remains, however, that basic studies on protection against human head injuries are becoming more vital than ever before. Therefore, an alternative method without requiring the use of primates will be necessary, according to our acknowledgement of the criticisms from the animal protectionists.

In this regard, engineering as well as medical experts concerned are currently working closely hand in hand in Japan. In fact, actual road traffic accidents are subjected to more detailed investigation and analyses than in the past by those experts, and such accidents are reproduced by means of experimental and numerical simulations, as an approach to the most appropriate methodology for the determination of the chains of human injury mechanisms.

In any event, steady and patient efforts are required for the implementation of studies of this nature. Global joint cooperation or exchange of information on study data will be necessary for the effective implementation of such studies.

QUESTIONS/COMMENTS for this paper follow the next talk

THE NATURE OF HEAD INJURIES

Rolf Eppinger

The work at NHTSA on this topic seems to be following a similar path to that being carried out in Adelaide and described by Tony Ryan, in terms of the following categorisation of the types of injuries that are seen in the head:

- focal injuries = depressed skull fractures, acute subdural haematomas (ASDH), other anatomical lesions;
- diffuse injuries = common in auto crashes, no gross anatomical lesion, characterised by coma and persistent neurological dysfunction.

Thus as a result of head impact there are skeletal injuries and a variety of focal types of lesions within the brain. There is also evidence of a sort of diffuse type of injury which is fairly common in automobile crashes, and has been identified from a variety of accident and clinical investigations. Neural dysfunction may occur without any gross anatomical lesions and can be characterised by persistent neurological dysfunction of the patient.

One of the landmark tests was the work of Thibault in which mechanical strain of an axon was associated with its dysfunction, that is, its ability to transmit its action potential upon stimulus. Figure 1 is a diagrammatic representation of Thibault's experiments using the squid axon, in which he established that there was indeed a relationship between the mechanical stimulus and the function of the neuron. What had not been explored in detail, and this could still be a factor, was the view that only strain was related to dysfunction. Because most biological materials are very visco-elastic, the failure conjecture gets even more complicated, that is strain and a strain/rate history might cause dysfunction of the neuron. From Thibault's very preliminary work, a relationship was proposed between strain or per cent stretch of the neuron and dysfunction, such that with a strain below 5%, there was basically no dysfunction, with a strain of between 5-10% there was reversible dysfunction which varied proportional to the strain, 10-15% strain gave irreversible dysfunction but no visible physical damage to the axon, while above 15% there was some mechanical damage of the tissue.

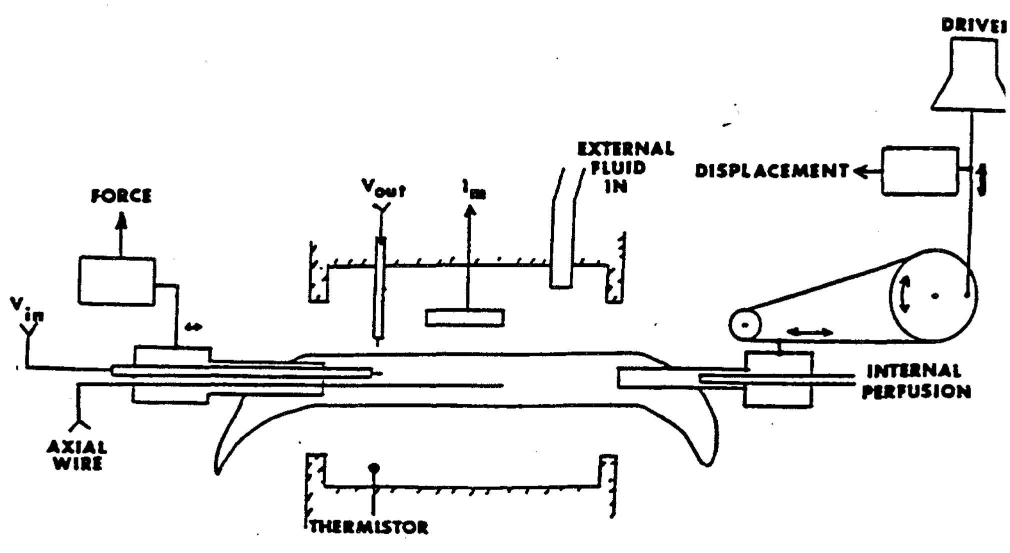


Figure 1: Diagram of Thibault's experimental set-up

If that is taken as the basic hypothesis it can then be said that neural dysfunction is proportional to the strain in the brain and then the strain is a function of the brain's geometry and forced motion. Koshiro Ono has shown how external stimuli can bring about internal motions of the brain. So to predict the brain motions, at least the complete motion that is applied to the skull must be characterised — in other words all three translational and three rotational accelerations. NHTSA's concept to develop a brain injury criteria is to use a finite element brain model to produce or determine brain strains given head motions.

Figure 2 shows some of the capabilities of the current technology to produce, given certain types of motion inputs, various types of parameters that can be examined throughout the brain in terms of either pressures or strains or stresses. This also allows the potential for examining other types of failure mechanisms. In a situation like this where the acceleration causes both positive and negative pressures it could be that in the negative pressure area, which is basically a drop from current atmospheric pressure, there could be a lot of tension going on in that part of the brain resulting in a failure due to a pressure gradient rather than just a strain gradient.

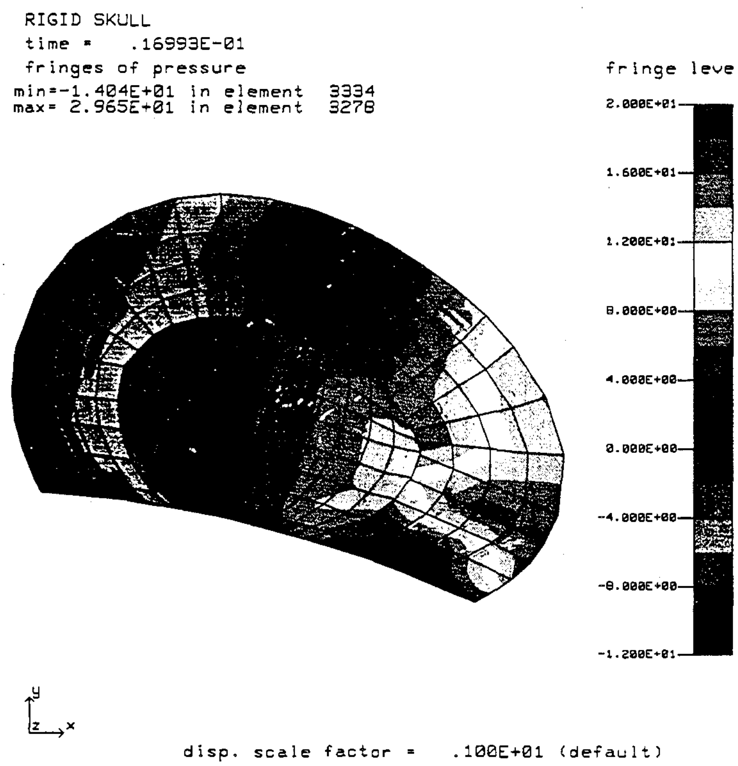


Figure 2: Rigid skull model

An example of the surface types of pressures that can be generated from the model is seen in Figure 3. The model can be sliced to see a variety of the patterns throughout the brain. If the model is reasonably accurate, it should duplicate the types of things that Tony Ryan sees upon autopsy from the actual crash environment.

Figure 4 shows the type of volume calculations made to determine how much of the volume within the brain exceeds various strain levels. The input to the model was a sine wave input, first positive and then negative, with the ratio 3:1 in time duration of the acceleration part to the deceleration part. Clearly, when the results were examined to see where these things occurred, various areas in the brain experienced various degrees of strains.

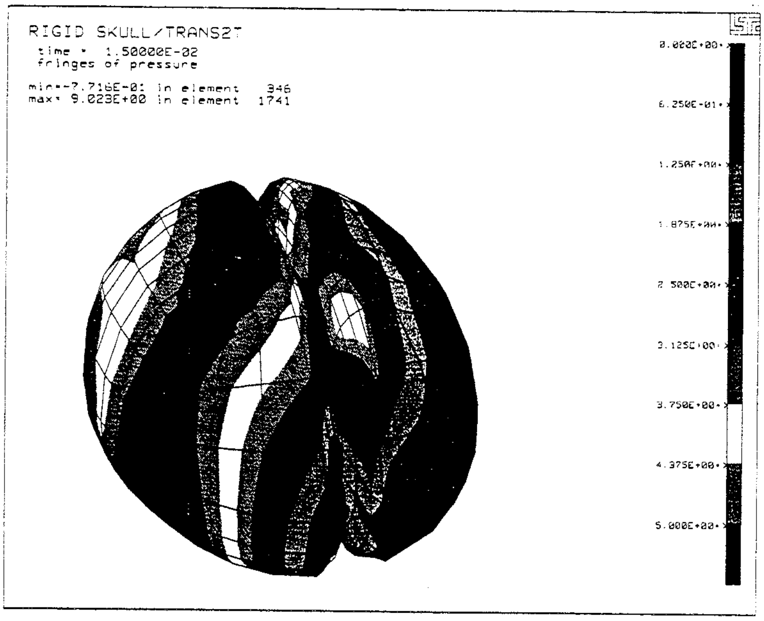


Figure 3: Surface pressure model

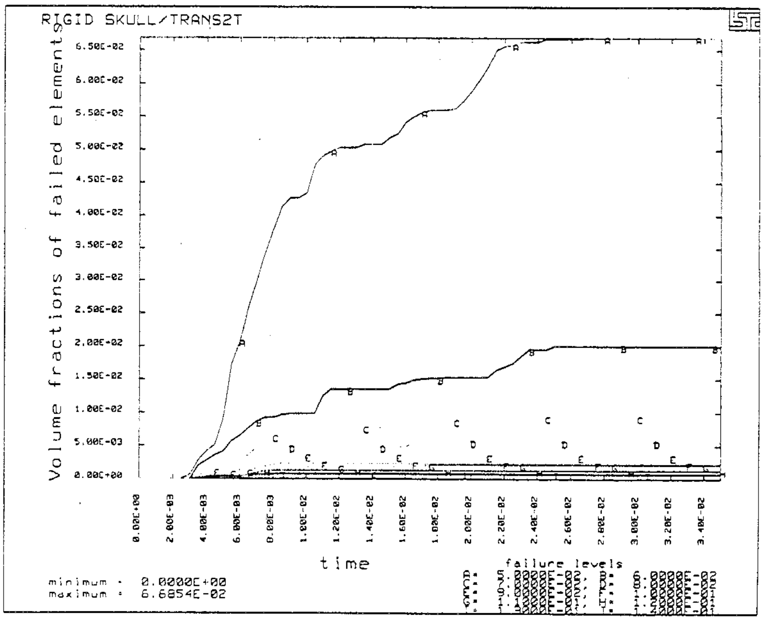


Figure 4: Volume fractions of failed brain elements over time

It is believed that the issue of subdural haematomas can also be addressed by the modelling process. The brain model is encased within a skull model and is capable of calculating the relative motion between the brain and skull. It is expected that the disruption of the bridging veins is probably proportional to the relative motion of the brain to the skull. The modelling efforts are not a finished process by any means and there are lots of challenges still ahead. Brain tissue is a very non-linear visco-elastic material and it can be subjected to very high strains (of the order of 100%). To just characterise this material correctly is a very difficult task. Historically, people have taken cadaveric samples of the brain and conducted a lot of material testing, but the material characteristics of an intact living brain have not been determined and are not known. Also the analytical techniques that display the results are not fully developed. Some good progress is being made now through contacts with some of the developers of the analytical routines. Although currently the work on strain has provided a very good viable hypothesis and the most progress has been made on that, the other potential material failure possibilities should not and cannot be discarded at this juncture.

Another aspect is that this initial hypothesis was derived from an unmyelinated neuron and so the effect of a myelinated axon must be considered. If an unmyelinated structure is stretched it is very likely there will be a uniform strain over the length of the axon, but with a myelinated fibre it is possible that the strain gets concentrated at the nodes of Ranvier, so that an overall stretch of 5% might produce 10% stretch at the nodes and very little stretch in between. The effects of that are not known. Thibault at Pennsylvania is doing some work using a sciatic nerve of a frog. Although the work is not complete, the results so far are very similar to those for unmyelinated fibres in terms of the percentage strains that seem to induce the various grades of failure.

Adjustments will have to be made for size. The making and validation of a human scale model will depend on a lot of experimental work using sub-human primates which have a much reduced brain size. Presently, issues of scaling and the extent to which geometrical detail will have to be included in the detailed modelling of the brain are not known.

It is hoped that, if this model is made correctly, it will be possible to produce a variety of different criteria for different sized occupants all the way from paediatric to adult. It should also be possible, if sub-human primate tests are conducted and the histology done afterwards, to scale from the primate back to the human, because in the modelling process the geometry can be adapted directly from the animal and the animal geometry can be duplicated. This should lead to the various criteria that should be monitored and allow rescaling of the model up to the human form.

The aim of this work was that once the model itself had been produced, it could be interrogated as an experimental entity, to see if some sort of simplified process such as an acceleration-based criterion, like the HIC, would be generated from the model itself. However, that seems to be becoming a more and more remote possibility, and since the computing powers have increased so much, it may be that if the model is developed and made robust enough, it could be used directly as the criterion. On work stations right now, with the acceleration-time histories as input, a solution can be obtained within 2 to 3 hours. That's a little bit longer than it takes to make a HIC calculation but the type of information generated is much greater. So it can be foreseen that in the near future, that would be the process by which the measured accelerations coming off the dummy would be evaluated.

The complete skull/brain model would also be able to assess various impact directions and sensitivities.

Another problem is that of the boundary conditions between the brain and the skull surface in terms of modelling. What really are those conditions? With the spinal fluid surrounding the brain inside the dura, there is a fairly slippery inter-connection. But the bridging veins and other structures that tether the brain to the skull introduce an element of uncertainty with respect to modelling. Friction or just a sort of sliding constraint between the brain and the skull could be used. Again there are all the internal membranes and the skull attachments and if the model is to be correct, the proper material characterisations of these other structures must be known so that everything works harmoniously and there is a good duplication of the real and geometric material properties of all the structures. It is hoped that in the near future this process will result in being able to process measured head accelerations through the finite element model and predict the volume of brain experiencing various maximum strains.

QUESTIONS/COMMENTS for Tony Ryan, Koshiro Ono and Rolf Eppinger:

Peter Makeham, comment: I think that Koshiro Ono's paper was very important as it was of a type that we won't see again for quite a long time. I note the comments that he made towards the end of his presentation about the ethical issues. I think that some of the information he has gained will be very hard to obtain in the future.

Peter Makeham, to Tony Ryan: I notice that you and a number of other speakers have spoken about the threshold of injury being related to strain. I can appreciate the difficulty in relating the forces applied to the head to strain experienced by brain tissue. In your own conclusions you related acceleration to the threshold and I just wondered if you have any thoughts on how you would bridge the gap between the two?

Tony Ryan: Yes, I think it's a problem for someone to work on — because that's the big gap in our knowledge. Ideally we should be able to relate a particular strain in a particular part of the brain to a particular clinical entity and outcome, but we don't have the knowledge about the physical properties of the brain substance, or about the connection between the inputs to the whole head and the exact amount of strain in a particular impact. So we're left with whole head movements, acceleration, things we can measure.

Peter Makeham: That's probably one for the visco-elastic modelist.

Ken Digges, to Rolf Eppinger: The Thibault model is certainly a very elegant and appealing model for understanding concussion and neurological damage and, as I understand it, is associated mostly with a diffuse axonal injury phenomenon, and yet there's a tremendous amount of injury that's of a focal nature where you see the haematomas and the damage to the bridging veins and indeed to the internal vasculature of the brain. Is there any corresponding theory of concussion associated with haematoma? And is there any injury criterion that goes along with the damage to the bridging veins and the focal injuries that's equivalent to the diffuse axonal injuries that Thibault has been working on?

Rolf Eppinger: They're a very active group in Pennsylvania and they've expanded their work to looking at more than just axons. They've studied both arterial and venous vessels in the brain, stimulating them by applying mechanical stress or strain. I think they found no response of the structure on the arterial side, but when little bridging veins were stretched they would actually go into spasm, shrink and stop the blood flow. So they can actually see a physiological mechanical response of the bridging vein in response to stretch so that it actually shuts down and prevents blood from exiting the brain. I'm not aware that the vessels were tested to destruction, but one can imagine that there is a destructive strain level that could be experienced with the vasculature. I would think that as we create these models, if we have a reasonable constitutive property and monitor the strain levels, we would associate the vascular damage also with strain. This is because I imagine that there are all these little tubes running in different directions, and how can I damage a tube? I surely can't damage it by squeezing it together, so I really think it's tensile strain that is the most predominant factor for failure.

Ken Digges, to Rolf Eppinger: Is there any kind of theory of concussion associated with either haematoma or damage to the vascular constituents, or is the concussion associated entirely with strain in the axonal areas?

Rolf Eppinger: That's a subject for conjecture. I think that what the researchers at the University of Pennsylvania have done, basically, is to demonstrate that you can shut down the action potential by mechanically stretching the neuron, so at least you have identified one way of causing neural dysfunction in the process of mechanical stimulus. If you now have haematomas and so forth that are adjacent to the axons and somehow that causes the same process to occur in the axon, I think that would be another possibility. Thibault's group has now continued the earlier work and watched and tried to understand what are the mechanisms that are associated with such phenomena: why does a neuron fail to function as a result of this stretching? They've postulated and, I think, have measured intercellular calcium at a 30,000:1 concentration gradient from outside to inside. Upon mechanical stimulation there is a tremendous inrush of calcium ions into the axon which starts a destructive process affecting all the internal architecture. They identified that through some sort of fluoroscopic process — I'm not familiar with it at all, but they've demonstrated that that actually occurs. And so they're actually getting down to the chemical processes that are going on as a result of strain. So we're saying that strain is the bad guy and they're saying, well, strain is just the initiator of other processes again. I think you can go on and on deeper and deeper in here. I don't know where it will end but from our point of view of attempting to come up with a criterion, we at least have a chance now in a thread of logic that says if we monitor strain and the volume of strain then we have a measure of both severity and extent of injury throughout the brain.

Peter Makeham, to all speakers: We have had three excellent papers which look at the field work that Tony Ryan has done on the non-intrusive methods, and then Koshiro Ono's very important paper work in terms of the laboratory and then your work, Rolf Eppinger, in terms of modelling. I was thinking of where this would lead us in terms of re-interpretation of the Head Injury Criterion, say, as a regulatory criterion to be used for the onset of head injury?

Rolf Eppinger: We're attempting at this point an initial assessment of something of that nature. In fact, one of my staff is now interrogating the model that we have in it's present condition and trying to apply a variety of pulses that change both in amplitude and duration and have both either just a pure rotational component or rotational plus translational component and doing it frontally and laterally, and we're trying to examine to see if we can get the iso-surfaces of volume of strain at various levels, just to get a feel of what those things are. Then we can also go back and make a comparative HIC calculation on the driving wave form that initiates that process and we can see if there is a correspondence or not in that process.

Tony Ryan: As part of that process I'd like to see Rolf Eppinger's staff person reproduce some of the injury distributions that we've found from known head impacts that happen on the road. That would give us a better idea of whether our predictions are correct or not.

Koshiro Ono: It is a little bit difficult to answer. If we look at our data, it is easy to find according to the head acceleration-duration it's simplified — it means velocity. And this curve came from the 30 km/h average and so when we apply this data, we must think about the stiffness and what kind of contact area and things like that. So the most simplified that we take into account are these three factors.

Donald Simpson, to Rolf Eppinger: I was fascinated by the finite element models. I just have a comment and a query over your problems with myelinated fibres. If I followed you correctly, you are basing your estimates on isolated un-myelinated axons and you are considering controlling that with a peripheral nerve axon — I think you mentioned a sciatic nerve axon. The problems obviously are that peripheral nerve axons are rather different from central nervous system axons in respect to the amount of connective tissue that envelopes them, and I would intuitively expect that you would find a difference between unmyelinated and myelinated axons because we have the experimental model of the baby who gets battered from time to time, whose white matter is unmyelinated and which does seem to behave differently from adult myelin. So I think you do have a problem there, and could I suggest that you get somebody who's prepared to test for you an isolated fibre bundle — the fornix would be ideal, I would have thought — from the central nervous system, where you can do mechanical studies on the fresh myelinated axon with reasonable confidence that it would approximate to a real life injury.

Rolf Eppinger: One possible alternative to that is that once we've established that this stretch really is an initiator of neural dysfunction, it's then a matter of how to go about establishing the critical stretch level. Now if we had the ability to go back and do animal testings and duplicate animal testing with the model and we could make some sort of clinical evaluation on the animal model when injury or dysfunction occurred, I could then just interrogate the model to see what strains were occurring in those same regional areas, and that would be a way of getting to that level rather than doing the isolated test.

Donald Simpson: I take your point. I think the only animal model that's gone some way to giving you what you want is the guinea pig optic nerve model and even there I would have thought that you were dealing with a structure that is a bit different from central nervous system axons.

Tony Ryan, comment: Could I refer to Ken Digges' question concerning clinical conditions related to brain injury, which I don't think was answered properly, and I think Professor Simpson could answer you. The question concerned theories of concussion associated with haematomas or damage to vascular constituents as distinct from direct neurological damage.

Donald Simpson: I don't think one really knows the answer to that. I'm sure that vascular injury is really important. I think everyone is impressed — we have 2 expert neuropathologists with us — and everyone is impressed with the frequency that small vessels are torn, and they do seem to get torn under different conditions from axons. But I think it possible that you might have to look at 2 different finite elements, one for modelling vascular injury and one for modelling axonal injury. I think they are rather different situations. I don't know if my colleagues would care to comment?

Peter Blumbergs: I think that not only do you have to model for vascular elements and axonal elements, but you possibly have to model for other structural components of the brain — the glial elements for instance and the connective tissue elements — and all of these would probably have different material characteristics. One needs to know how each of these component parts behaves. Tony Ryan's analysis really only deals with vascular injury, the vascular elements in the brain, and ignores basically the axonal components and all these other components, which are probably very important in actual neural dysfunction.

Tony Ryan: My assumption is that the haemorrhages that we record are associated with neuronal damage. It's partly because we don't have any other marker readily available and to determine diffuse axonal injury takes more time — the person has to stay alive for 12-24 hours — whereas we're dealing with people who have died within an hour, or 12 hours at the most, so you don't get any evidence of more diffuse neuronal injury. So we're left with the vascular markings which to me seems a perfectly reasonable way of working. Whether your finite element model is just using a lump of brain without differentiating between vascular, neural or connective tissue, that's another matter.

Rolf Eppinger: The modelling process in a lot of engineering techniques is not very fine: let's say if you have composite materials, you try to look at it from the gross response stand point. I'm now envisioning what the brain looks like: it's a sort of nest of axons with the glial cells and the vascular, and it's all sort of mixed together, but you have these three things and so it's a composite material. Let's compare it to, say, 3 colour spaghetti: suppose I have to model 3 colour spaghetti and each colour of spaghetti has a different type of characteristic and it's all in this big pot. What engineers tend to do is model that as a single material and squeeze it and twist it and everything and try to get that characterisation of the combined material. Then I imagine that if we would have that for brain material and it's composite, then we would have to ask the question 'when do particular types of elements fail?' Now maybe axons might fail under strain at 5% and maybe the vascular system might do it with a pressure in excess of 10 psi. In other words, the failure mechanism or the engineering parameter that describes the condition of failure might be different for each one of these intrinsic elements. But I model it as a homogeneous material and then I would say how it fails would be different. We'd have to identify those various things, of those important elements and when they fail. And that's a formidable challenge, I might add.

Ken Digges, to Koshiro Ono: One of the other very important parts of head injury is to attempt to relate the injury criteria to some kind of risk of injury and risk of AIS level. At the present time, about the only relationship that we have is one that was, I think, developed by General Motors' Dave Viano, or some of his staff, in which he examined cadaver data and developed a curve that related the risk of AIS 4+ injuries to HIC. But there's virtually nothing on AIS 3s or AIS 2s. Part of the reason, I guess, is that if you're using cadavers, the AIS 4 level is about as fine a grid as you can put the information through. I was wondering if with your animal data it would be possible to analyse results in such a way that you could come up with an injury risk function which is related to HIC for injury levels other than AIS 4?

Koshiro Ono: It is possible to analyse head injury criteria and risk of injuries by using these data, but at the end of our experiments using the primates, I was not concerned about description by the AIS. Now AIS 90 is a little sophisticated to describe the injuries of these kinds of experimental data. So at that time we didn't analyse that kind of injury risk. But now we want to try to analyse the data in this way. Basically the velocities come from head acceleration times duration — it's a very simple calculation and it's OK to describe what happened. So I did not mention about the HIC at that time, and did not analyse this kind of risk of injury, but now it's possible I think.

Michael Henderson, to Rolf Eppinger: I think that Thibault's model and the computer modelling at NHTSA is just beautiful to watch, but I worry about the absence of the involvement of fracture and the clinical importance of fracture of the skull. As Tony Ryan pointed out, the site of the impact is going to be important in determining injury, the way he is measuring it, but also important are the properties of what is hit by the head. At a given speed of impact, if you hit something small and sharp, it's far more likely to cause clinical damage than something that is broad and smooth. However, both the Thibault and the computer models show the system as a piece of jelly inside a rigid skull, so the results in terms of brain injury are the same whatever the shape of the impacted surface. Can you comment on the relationship of skull fracture to this modelling?

Rolf Eppinger: The model we've developed we use in two different ways. One way is to put it inside a rigid skull and stimulate that with rotational and translational motions. The other way is to create a skull model and put the brain model inside the skull model. That model is then capable of being propelled at an A pillar that's either padded or unpadded and the entire process then of how the A pillar and the skull interact is calculated — correctly, we hope. In theory, it's a completely deterministic model — you set up the initial conditions and you say ' I have a skull and a brain and so forth and I now also have a correct model of the object that I'm striking, whether it's a padded A pillar or whatever, and all of the interactions are calculated so that you would then see: if the skull deforms, that should be calculated: if the skull fractures, that should be calculated: that indentation and the volume change within the cranium, that would affect what would happen to the brain. Those are all interconnected and being calculated at the same time. We can run it that way, and that's a full deterministic process. The other way I was suggesting was that if we would run a dummy into something and then try to make an evaluation in terms of brain injury, I would then take the motions that I observed on the dummy head and drive the model with it. If I would want to evaluate the structure and how all that works together, I'd run the full model.

Tony Ryan: I'd add a comment there that we've found, when we've tried to attribute some of our haemorrhagic injury to fractures, because that may well be part of the local injury process, that in fact fractures contribute something less than about 10% of the injury to the brain. The fracturing process seems to be largely independent of the brain damaging process.

CLOSING REMARKS

Jack McLean

The first thing I want to say concerns the suggestion that a Panel on the Biomechanics of Trauma be formed in Australia, a suggestion which I believe has the support of most, if not all, persons at this Conference. It should be made clear that the idea that such a Panel would be under the auspices of the Institution of Engineers in no way implies that membership of the Institution would be a prerequisite for association with the Panel. I don't think that is the intention at all. I think the Panel is likely to be set up within the Institution in any event, but also there might be advantage in considering whether we're at the stage for establishing a Crash Injury Biomechanics Society in Australia, maybe a bit along the lines of IRCOBI in Europe, with a view to meeting, say, once a year or once every 2 years. In connection with that, SGIC have indicated that they have an interest in possible continuation of meetings of this type, so that this conference hopefully might not be a one-off event as far as they are concerned. I now invite Michael Griffiths to give his comments and hope that others will contribute with their ideas later.

Michael Griffiths: I don't have a lot to add, really, to what Jack McLean has just said. From time to time people have talked about some sort of Australian society — a subgroup of IRCOBI or suchlike. It's just that at the present time the Institution of Engineers has formed a College of Biomedical Engineering, giving it the same status as the main Mechanical, Civil and other Colleges in the Institution, so that has somewhat elevated the whole area of biomedical engineering in their society. They then talked about the prospects of forming various societies or panels in that College of Biomedical Engineering and one possible society was a Biomechanics and Trauma Society. They raised that with me and this conference was a possible time to discuss it with really most of the relevant people in Australia.

So I agree with Jack McLean. Let's think about it and have another discussion later in the day. It's an opportunity to have some sort of society. I'm not strongly promoting a line of affiliation with the Institution of Engineers, but it's an opportunity if people wish to use it.

LATER:

Jack McLean: Following the remarks made earlier, I'd like to propose that an ad hoc committee push along the idea of formation of a Panel — and if I could exercise Chairman's privilege and suggest that the committee be Peter Makeham, Laurie Sparke, Michael Griffiths and myself, not implying at all that any of these persons will necessarily continue in that role, but we'll put our heads together and come up with something that we think might be workable; maybe a group to organise a conference every 2 years — or every year, if we can get support — and maybe an informal newsletter we might circulate every 6 months or something of that nature.

Jack McLean's suggestion was proposed as a formal motion by Peter Caldwell, seconded by Michael Griffiths and passed unanimously.

Michael Griffiths then thanked Jack McLean for organising such a successful conference, and this received general acclaim.

Ingrid Planath: I would like to thank you all very much for being prepared to listen to this lady with the Swedish accent for three speeches, and also that we could come here and get all this interesting information from the excellent other presenters. I would also agree with Michael Griffiths that I thank Jack McLean in particular and all the organisers very much for hosting this conference and for the excellent hospitality. Kangaroo Island was one good thing; another was that we learnt the secret behind Australian Rules Football, which was a very good game we could watch on Saturday. Thank you very much.

Jack McLean: In inviting people from overseas, as you will readily understand, Jean Paul Sartre's play 'No Exit' came to mind. For those of you who know the play, you may recall that it was Sartre's concept of hell. A group of people were in an elevator on the way down to hell and they started out by telling each other what fine people they were, then they all collapsed into silence and finally began to admit that back in life they actually had blotted their copybooks one or two times; one had murdered a child, and so forth. Then they sat and stared at each other and eventually one said to the lift driver 'when are we going to get there?' and the lift driver said 'we're never going to get there: this is hell!'. Well — you may be a little puzzled as to the relevance of all this, but I was very mindful of the fact in inviting folk to come from the other side of the earth, that they sit in a Boeing 747 for a very long time, and I think if Sartre were writing that play now he'd have people sitting in a long distance jet and they'd eventually say to the stewardess 'when are we going to get there?' and she'd say 'we're never going to get there: this is hell!'.

So from that point of view, I hope that our overseas visitors have found their time here profitable and enjoyable, and I hope you don't find the trip home too onerous. I'm sure I speak for all of us when I say 'thank you all very very much for taking the time to come; in some cases it's been in vacation time, not work time, and we appreciate it very much indeed'.

I've told some of you the sequence of events by which this seminar came about. When I spent 6 months in France with Dominique Cesari in 1990, we agreed that a meeting such as this would be valuable. He indicated that he could come to Australia at this time if others prominent in the field were also available. Then in January, when I was at the Transportation Research Board meeting in Washington, I mentioned the idea to Rolf Eppinger who expressed interest. Thus it seemed that a conference was a distinct possibility and I began to think seriously about money.

Russell Cowan from SGIC came to visit the Unit and I said pretty much what I've just said now, emphasising that it would be ideal to have a conference to include most of the people with a professional interest in the area in Australia. He asked what it would cost and soon afterwards SGIC agreed to fund the seminar.

I then contacted Koshiro Ono, who very kindly said he could come — and the others — and so it went on. Ingrid Planath is here, courtesy of Volvo, so we should thank Volvo too. Ken Digges was to be in Australia, as also was Nicholas Shewchenko. Interstate contributors were very cooperative. So it all worked out astonishingly well in that regard.

SGIC really has made this event possible and with a level of support that has left nothing to chance. So even though there is no direct representative of SGIC here, I think it would be appropriate if we were to applaud their support.

Now to the real closing remarks. I think that our last session on head injury possibly more than any other showed how inadequate is our level of understanding of injury mechanisms. As far as brain injury is concerned, we're still groping towards an understanding that would be useful in practical terms, and yet we have written into legislation in the USA the Head Injury Criterion. I predict that after the Australian NCAP results are publicised, everyone will be talking about whether such and such a vehicle passed HIC or failed HIC without really having the faintest idea of what they're talking about. We are the people in Australia who do have some understanding of what HIC is about, what the limitations are, what it means if one vehicle has a HIC of more than 1,000, that it's a barrier crash test and the fact that a light weight car has the same HIC value as a heavier car doesn't mean that they both offer the same level of safety if they run into each other on the road. So I think we all have a responsibility to take every opportunity we can to try to inform people about what these magical numbers mean and about how far it is reasonable to go in using them as an index for vehicle safety. The fact that the concept of NCAP testing is very new to Australia, inevitably means that there are going to be teething troubles as we catch up with the United States in this approach to the assessment of vehicle safety. The motoring press has a very important role to play in interpreting to its reading public what's going on with NCAP testing and other aspects of crash injury biomechanics as they relate to the safety of vehicles. I certainly hope that we will manage to get through this fairly steep learning curve without too much trauma — psychological trauma, that is. Similarly, I envisage that if we do get some sort of an association off the ground, it might play a very valuable role in trying to ensure that if we get funding for another meeting like this, we'll still be talking to each other and want to come!

That is all I have to say. Thank you for coming. I'm very pleased with the way things went and I get the impression that most of you are also. We will be publishing proceedings, as soon as we possibly can, and I hope they will be a comprehensive and detailed record of what has been discussed here over the last two days.

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