Dummy Development: Spinal Injuries

TRANSPORT RESEARCH LABORATORY

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Dummy development to evaluate spine injuries

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Executive Summary

Minor injuries to the spine can have serious consequences to the quality of life and have the potential to alter significantly both behaviour and personal independence. Injury data, particularly insurance based personal injury claims show that low severity injury to the neck is a major problem with a high associative cost, which appears to be increasing. The cost of whiplash associated disorder type injury is very high with some estimates suggesting that the annual cost of whiplash associated disorder personal injury claims is as high as £2535M or 18% of the total insurance cost of road traffic accidents in the UK. Recent insurance claims data suggest that whiplash injuries account for 80% of the personal injury claims bill.

Accident data have indicated that whiplash injury can occur in impacts from all directions, but the principal direction is from the rear. No regulatory biomechanically based rear impact test procedures exist against which to assess safety systems designed to mitigate whiplash injury. The insurance industry has been supporting several avenues of research aimed at controlling whiplash associated disorder causing factors, in order to limit and control the high level of personal injury claims.

Recent in-depth UK clinically based accident studies have shown that many of the cases studied that had a whiplash injury also had another spinal injury. In a later study that focussed on spinal injury, whiplash injury was observed in all their patients who had lumbar strain injuries. The UK clinical accident data, that specifically studied whiplash associated disorder injury, suggested that any evaluation test procedure, to address low severity rear accidents with a view to reducing whiplash associated disorder type injury, should cover the whole spine and not only focus on the neck as an isolated entity.

In 1999, when the project, presented in this report, commenced no dummy existed that had been shown to be suitable for use in a regulatory rear impact test to assess rear impact spinal injury risk. It was noted that two dummies were in very early stages of development in Sweden and the Netherlands, but neither was available for evaluation. Within the European regulatory framework two crash test dummies existed: the Hybrid III, which was designed and being used for high energy frontal impact, and the EuroSID dummy designed for use in high energy side impacts. Dummy development is a costly process and the main aim of the project was to see if the Hybrid III dummy could be used or easily adapted for rear impact evaluations addressing the issues raised in the accident studies carried out for the Department.

The project was divided into several phases and the direction of the research was modified as it progressed. To initiate the study, a literature review was carried out. This review found that the data available for specifying required whiplash dummy performance was limited. However, from this review, a list of points were extracted that had implications for designing whiplash dummies. This was then used to direct the later phases of the programme.

A series of volunteer, rear impact, sled tests have been carried out in order to obtain suitable data for specifying the performance of whiplash dummies. EMG readings from the volunteer tests showed almost instantaneous tensing of major muscle groups in the back. Comparison of the EMG results with the kinematics of volunteers who exhibited some tensing with those who exhibited little muscle tensing, showed that bracing significantly affected motion of the head. This strongly suggests that PMHS data is not suitable for developing whiplash performance corridors. Also from the tests to the volunteers, a series of performance corridors have been derived. It is suggested that these corridors can be used to specify the low severity response of whiplash dummies and new dummy developments should be evaluated against these targets.

Within this project, a Hybrid III dummy/human mathematical (FE) model has been developed. The model is useful in predicting kinematics and load concentrations. However, further development is needed before it can be used to predict injury risk. The potential value of mathematical modelling has
been established. Extension of the work described in this report should include the incorporation of better biomechanical data and enhanced validation.

Rear impact sled tests have also been carried out with three different anthropomorphic test dummies and two dummy variants with alternative necks. Two of the dummies were revised during the project duration, so the upgraded versions were also tested, as they became available.

The dummy results have been compared with the volunteer performance corridors and this has shown that no version of the Hybrid III dummy tested gives a satisfactory performance despite it meeting some of the proposed performance criteria. This is fundamentally due to the dummy having a rigid thoracic spine. Although the THOR has flexible joints in both its lumbar and thoracic spine, it is still not considered suitable for use as a whiplash dummy because it does not demonstrate the required biofidelic flexibility of the spine and torso. The BioRID appears to show potential for use as a whiplash dummy. However, the torso together with the spine require greater flexibility and further tuning of the individual spine sections’ stiffness is needed. The BioRID demonstrated rearward, T₁ marker motion most like the volunteers. However, inconsistencies within the set-up were noticeable. Before a rear impact test procedure is established better definition of dummy positioning techniques should be described.

Although it appears to be possible to compensate for a non-biofidelic rigid thorax, found in many dummies, by adjusting the neck stiffness, this method will only produce reliable results in the impact situation for which the neck was adjusted. A test device should produce sensible results over a range of impact conditions. It is strongly recommended that future regulatory dummies should have a full biofidelic spine. The current research only examined two-dimensional dummy behaviour. In real world impacts, occupants will be loaded from a number of different directions. Future research should examine oblique behaviour to ensure that the injury predicting devices (the dummies) work in impacts other than pure rear.
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1 Introduction

1.1 Background

Historically the direction of biomechanical research, dummy development and associated test procedures has been directed towards the reduction of serious and fatal injuries. This strategy has been very successful and has led to significant reductions in all levels of injury - fatal, serious, moderate and slight. Over recent years attention has been drawn to the consequences of clinically less serious injuries and to the disabling nature of some injuries, which in themselves are not life threatening as assessed by the injury coding system AIS (Abbreviated Injury Scale). Minor injuries to the spine, in terms of risk to life, can have serious consequences to the quality of life and have the potential to alter significantly both behaviour and personal independence. Injury data, particularly insurance based personal injury claims show that low severity injury to the neck is a major problem with a high associative cost, which appears to be increasing. The cost of Whiplash Associated Disorder (WAD) type injury is very high with some estimates suggesting that the annual cost of whiplash associated disorder PI (Personal Injury) claims is as high as £2535M or 18% of the total insurance cost of road traffic accidents in the UK. Recent insurance claims data suggest that whiplash injury accounts for 80% of the PI claims bill. The reasons for this level of PI claims are complex and may be influenced by a number of factors ranging from ‘insurance claim awareness’ to ‘changes in vehicle accident patterns’ to ‘changes in vehicle design’ and ‘societal pressures’.

Low impact severity neck injuries are often called Whiplash, or WAD injuries. Other researchers also classify whiplash associated disorder injuries as CSD (Cervical Strain Disorder) injuries. It is thought that whiplash injury can encompass a number of different injuries and outcomes. A new scaling system for Whiplash injury has been developed in North America called ‘QTF’ - Quebec Task Force. It is a new clinical scaling system that some researchers believe can be used to assess the severity of whiplash associated disorder injury [1].

Recent in-depth UK clinically based accident studies have shown that injuries to the cervical (neck) and lumbar (lower) spine [Section 13], for restrained occupants, are more prominent and more significant than previous analyses had indicated. The UK clinical accident data, that specifically studied whiplash associated disorder injury [2, 3], suggested that any evaluation test procedure, to address low severity rear accidents with a view to reducing whiplash associated disorder type injury, should cover the whole spine and not only focus on the neck as an isolated entity. This was because many of the cases studied who had a whiplash injury also had another spinal injury. In a later study that focussed on spinal injury [4] they observed 100% whiplash injury in all their patients who had lumbar strain injuries, thus confirming some of the observations made in their former studies.

Accident data have indicated that whiplash injury can occur in impacts from all directions, but the principal direction is from the rear. No regulatory biomechanically based rear impact test procedures exist against which to assess safety systems designed to mitigate whiplash injury. The insurance industry has been supporting several avenues of research aimed at controlling whiplash associated disorder causing factors, in order to limit and control the high level of PI claims. Recent developments have seen the publication of a rating system for seat head restraints, based on a North American head restraint position test procedure [5]. This procedure assesses the position of the head restraint using a headform extension fitted to the H-point manikin [6]. Head restraint ratings are being published via an insurance based web site [7]. The purpose of this ratings system is to encourage better positioning and properties of head restraints with a view to limiting head excursion in rear impacts, making the pragmatic assumption that motion limitation will reduce the incidence and severity of whiplash associated disorder injury. In addition the insurance industry is proposing a dynamic sled based test procedure to test vehicle seats in low severity rear impacts [8]. In support of these activities the automotive industry have been developing a new test dummy, focussed on rear impact loading to the spine, the BioRID.
In 1999, when the project presented in this report commenced, no dummy existed that had been shown to be suitable for use in a regulatory rear impact test to assess rear impact spinal injury risk. It was noted that two dummies were in very early stages of development in Sweden and the Netherlands, but neither was available for evaluation. Within the European regulatory framework two crash test dummies existed: the Hybrid III, which was designed and being used for high energy frontal impact, and the EuroSID dummy designed for use in high energy side impacts. Dummy development is a costly process and the main aim of the project was to see if the Hybrid III dummy could be used or easily adapted for rear impact evaluations addressing the issues raised in the accident studies carried out for the Department.

The project was divided into several phases. Firstly a review of the literature on whiplash injury. This was then used to direct the later phases of the programme. The second phase was an evaluation of existing test dummies and thirdly their modification as appropriate. Within the programme, phases of crash testing and occupant simulation were planned with scope to modify the direction of the research, if found necessary.

2 Methodology

The project originated with a methodology consisting of the evaluation and adaptation of an existing crash test dummy, to make it suitable for use in evaluating spinal injury in a rear impact. The project commenced with a broad review of the available literature [9]. This was then used to focus the practical areas of the research effort. Following the literature review the focus of the project was amended, as the base data on which to evaluate a rear impact dummy for spinal research was not available. The data that was available was largely inappropriate having been derived in test procedures and test conditions that could not easily be replicated or was considered inappropriate for biofidelity evaluation. The research program then undertook a programme of low-severity rear impact testing with volunteers in two main objectives. Firstly to understand how a human would behave in a well controlled rear impact test, at sub-injury levels and secondly to define a set of high quality biofidelic design and evaluation target corridors for dummy design. To obtain the necessary ethical approvals for such work some preliminary dummy testing, using the Hybrid III dummy, was undertaken to give the ethical committee guidance as to the expected magnitude of forces and accelerations that a volunteer would be subjected to. Other testing was also carried out to define an appropriate test protocol, prior to application for ethical approval.

Two phases of volunteer testing were carried out. The first programme was a half speed test, to inform the volunteer of what was going to happen and to aquatint them with the type of research and the severity of the testing that they would be undertaking. Following the initial acclimatisation test all of the volunteers were happy to continue with the research programme, having suffered no adverse side effects. A full impact severity test was carried out on each of the volunteers after a period of seven trouble free days. Following the volunteer testing programme, appropriate dummy design and assessment target corridors were developed from both transducer and kinematic data.

Early testing with the Hybrid III dummy and information gleaned during the literature study strongly suggested that the basic Hybrid III dummy was totally inappropriate for low-severity rear impact research without significant modification, which was not possible with the resources available within the project. However, as it is currently one of the most widely used dummies for vehicle tests, it was kept in the assessment programme and used to establish a ‘baseline’ against which any alternative dummy performance could be evaluated. As noted earlier, outside of the spinal research program presented in this report two specialised ‘Whiplash’ dummies were under development in Sweden and the Netherlands. The development of these two new dummies went through many evolutionary stages where various combinations of new parts (initially new necks) were combined with standard or modified parts from existing dummies. Some of this work has not been formally reported. At the time of completing the TRL project the Swedish work had lead to a new dummy the BioRID and the Dutch work to an improved neck and pelvis flesh for use with the Hybrid III dummy. In addition, the THOR
(Test device for Human Occupant Restraint), a more advanced frontal impact dummy with some spinal flexibility, was also being developed in the US, which was thought to possess some positive features, not present in the Hybrid III dummy. Five different dummy options were assessed in the main dummy assessment programme, the BioRID, the standard Hybrid III dummy, the Hybrid III with the improved TRID neck, the standard THOR dummy and the THOR with a EuroSID neck. The last option was devised by TRL, because the EuroSID neck was known to have good flexibility. These dummies were evaluated and compared to the volunteer data that had been generated.

During the latter part of the project and independent of it, two of the dummies selected, the Swedish BioRID dummy and the US Frontal impact dummy THOR, had been further developed and were commercially available. Seeing that both dummies had been updated, further tests were performed on these modified versions to determine whether the improvements had in fact made them more appropriate for use as rear impact spinal research dummies.

This report reviews the development of the dummy design and assessment targets and the performance of the various dummies making proposals as to where future effort may be needed.

3 Literature Review

A wide-ranging review of the literature was carried out in the initial phases of the study to put the project into perspective and to form a focus for the later research program. The literature study clearly showed that much work had been undertaken to identify the basic causes of whiplash injury, the development of appropriate human surrogates and the development of appropriate testing procedures. Unfortunately most of the effort had been focused on the neck rather than the whole of the human spine, which the accident studies had suggested should be considered along with the neck. The literature study is included in Annex 1.

3.1 Summary

The literature review report presented a comprehensive review of the published literature on the biomechanics and injury mechanisms of the spine in relation to motor vehicle accidents and dummy development. A brief survey of the anatomy of the spine revealed that it has an intricate structure and that the interaction between the various ligaments, muscles, discs and bony structures is complex. However, a thorough understanding of the construction and biomechanics of the spine is required before it can be accurately modelled either physically or mathematically. Such knowledge is also essential to the understanding of injury mechanisms.

In the past, dummy design engineers have tended to look at parts of the human skeleton, such as the spine, mainly from a mechanical perspective of supporting other dummy elements with little reference to any medical information. There is a huge quantity of medical literature concerning injuries to the spine, especially the neck, and a full perusal of it was beyond the scope of the project. Nonetheless, the review focussed on some of the more relevant technical papers.

Much of the medical literature dealt with the diagnosis, treatment, rehabilitation and prognosis of spinal injury and at first sight appeared to be of limited use in the design of an improved dummy spine. However, the content of this literature highlights the fact that some of the long term disabling injuries (e.g. behavioural dysfunction) may not be predicted from instrumentation measurements made using a lifeless dummy, but may be a result of another injury to another part of the body. The review benefited from the input of spinal surgeons in Manchester and Nottingham. These valuable collaborations should be maintained in future work.

The subject of cervical spine injuries was discussed with various research groups. There was and is still no overall agreement on the mechanisms and injuries that cause whiplash associated disorders. Each medical expert has his or her own favourite theory about how these types of injury are produced. It is also noted that some research groups suggest that injuries may not only be caused during the
impact loading phase but also during rebound. This is quite different from normal injury producing theories that suggest that major traumatic injury is caused during the loading phase. The leading theories were highlighted in the review. It seems to be most likely that whiplash injuries result from many different mechanisms, and different occupant loading sequences could cause different whiplash associated disorder injuries. Thus, the design of a biofidelic neck and spine must allow the measurement of a number of potentially relevant parameters for research into injury mechanisms, accompanied by a continual monitoring of the importance of those parameters to current whiplash research.

Severe spinal injury, which may involve damage to the spinal cord, ligaments and vertebral bodies, is better understood than whiplash associated disorder injuries, particularly in the thoracic and lumbar spine regions. However, more recently this section of the spine has been rather neglected in favour of research on the neck and whiplash injuries with their high associated cost. Thus there is a shortage of data on the biomechanics of the thoracolumbar (Section 13) region of the spine and it’s tolerance to impacts and static loading. There was also a paucity of information on the physical properties of the vital components of the thoracolumbar spine with which to develop refined mathematical and physical models of the human spine. What little data there has tended to be ill-defined and disparate with large variations in the results reported by different researchers.

It was argued that the bias of research effort towards study of the neck could be justified as accident statistics showed that this was where most spinal injuries occurred. However, serious injuries to the rest of the spine have also been reported and are also important. The disabling effects of those injuries can have considerable financial implications for society at large, not to mention the cost in terms of human suffering. A brief survey of spinal injuries in traffic accidents was detailed in the review. The results of the survey suggested that the occurrence of thoracolumbar injuries were much rarer than cervical spine injuries.

The survey showed that mathematical models could be cost effective and a valuable and flexible research tool for the study of the response of the spine to impact. These models could be particularly powerful when supported and validated by a complementary experimental test programme. Thus, it was expected that finite element modelling could have a prominent role to play in any future work on dummy spine development at TRL.

The limitations of the Hybrid III dummy spine were discussed. These deficiencies are well known and were widely reported in the literature. The original design of the Hybrid III neck is now twenty-five years old and knowledge of its behaviour is that more detailed. Some of the theories which circulated at the time of its development, such as whiplash injury is caused by hyperextension [Section 13], have subsequently been discredited as being the sole cause. It has been reported that the Hybrid III neck was far too stiff for low severity impacts, which may not be surprising as it was designed for high-energy frontal impact conditions. The rest of the Hybrid III spine is also clearly lacking in human biofidelity for rearward flexure [Section 13] as within the dummy it is a rigid structure on which to support a rib cage compliant with frontal deflection. It is possible that current measurements of head acceleration using the Hybrid III may yield misleading results because the head sits on top of a rigid spine and stiff neck whose dynamic behaviour does not match that of a human spine. The new THOR dummy, developed in the USA, was said to mark an important stage in anthropomorphic dummy design, having the potential to offer improved biofidelity of the neck. In addition it incorporated some degree of flexibility in the thoracic spine.

Engineers and scientists in Sweden have been very active in designing a new dummy spine that replicates the flexible behaviour of the human spine in extension. Much of the Swedish research was founded on a hypothesis first proposed by Professor Aldman in 1986 who postulated that sudden and violent pressure changes, which occur within the spinal canal of the neck, during a rear impact may be sufficient to cause damage to the spinal ganglia. Subsequent experiments with pigs demonstrated that this was a feasible injury mechanism. It appeared to be all the more plausible because the typical
symptoms of whiplash (e.g. dizziness, radiating pain etc) are consistent with the suggested neurological damage.

3.1.1 Implications for dummy requirements
• The spine of a whiplash dummy should ideally be flexible from lumbar to head.
• There are insufficient data to specify the stiffness of the spine.
• A flexible dummy spine might need a system to stabilise it, to prevent lateral slumping, similar to the main muscle groups in the human back (the erector spinae).
• A measurement system is required to record the amount of neck distortion.
• A system may be required to replicate the spinal canal so that pressure changes can be measured, alternatively it may be possible to infer pressure changes from other measurements.

4 Impact testing
A number of different test methods have been used in the study of rear impact by TRL, within the reported project and by other research groups carrying out whiplash-based studies. Within the research programme three test methods were used, each method having both positive and negative features. The first method was based on the conventional deceleration impact sled, whereby an impact sled is accelerated up to a desired velocity followed by a period of rapid but controlled deceleration – the impact pulse. Normally the sled runs horizontally with impact energy being supplied from a variety of sources. A modification of this pure horizontal sled is the ‘inclined plane rig’ in which both horizontal and vertical motion is combined via gravitational effects. The third technique, used within the main body of the projects research, is a twin sled rig with one sled being a powered bullet mass that impacts a free moving unrestrained sled through a controlled energy transfer system thus developing the desired crash pulse.

4.1 Single sled
4.1.1 Horizontal
The initial baseline trials with a Hybrid III dummy were performed in the TRL Dynamic Restraint Testing Facility (DRTF). The DRTF makes use of a number of elastic ropes connected via a roller system to a sled mounted on a pair of horizontal rails. The rails restrain the sled such that it can only move in one direction, i.e. it cannot lift off the rails. The sled is accelerated up to the required speed by the elastic ropes which then go slack leaving the sled free to impact an arresting system which slows the sled at the required rate to simulate a vehicle crash. The tests made use of the TRL modified ECE Regulation 44 test bench reversed to generate a rear impact to the seated rider. The TRL modifications to the test bench were to make the seat more suitable for this type of test and are described in Appendix 2. The main disadvantage of this type of test is that the positioning of the dummy may be disturbed before impact when the sled is accelerated up-to-speed. However, for high-energy impacts this is a valid test method, where gross dummy displacement may be expected, as any slight variation in the position of the dummy at \( T_0 \), the point of impact, would not be significant. At low impact severities dummy displacements will be small, thus small changes in initial positioning could have more significant consequences on overall dummy trajectories, and associated measurements.

When the project commenced other sled options were not available and the noted deficiencies were deemed to be tolerable. Following the initial project research, when it was identified that appropriate dummy design targets were needed against which to assess dummy behaviour, it was concluded that an alternative more controllable and representative test method was needed.

4.1.2 Inclined plane
The inclined plane is a simple low cost test system and one that has been widely used for low severity impact testing with volunteers, with no reported problems. A version of this test device has been called the ‘Seat Belt Convincer’ for the promotion of adult seat belt use. This test device consists of a simple inclined plane down which a small trolley runs freely, on which is mounted a car seat and seat belt. In use a subject sits on the seat. The sled is then pulled up the slope. At the top of the slope the sled is
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released allowing the seat, sled and rider to travel down the slope until the end when it impacts an appropriate energy attenuation device. Impact velocities of about 7 km/hr are achieved with a ‘short sharp stop’ sufficient to demonstrate to the rider the value of a seat belt restraint.

TRL examined the potential of this test device for use in the rear impact study in a short evaluation programme using the Hybrid III dummy but with the vehicle seat facing backwards, up the slope. This was found to be a useful low cost test device but it was finally thought that combining vertical and horizontal accelerations could compromise features that could be important to the study and would not replicate well the in-car accident situation. In addition the amount of available test energy was limited by the height of the slope, which meant that long-duration controlled decelerations were not possible. A further problem with this type of test facility, when used by volunteers, is the degree of anticipation which could lead to the subjects being tensed, since an impact would be ‘expected’ after a known period of time. The seat belt convincer was therefore not used for the main evaluation programme. Simultaneously, TRL had been developing a twin sled system in their DRTF, under a separate project.

This type of facility had already been used for volunteer evaluations [10] and the literature survey had identified it as providing good data.

4.2 Dual sled

The dual or twin sled test facility was an adaptation of the TRL DRTF test rig, described in Section 4.1.1. Instead of accelerating the dummy or rider up to the desired velocity and then performing a controlled deceleration, the dummy or rider is seated on a stationary free running sled. A second sled, moving at the selected velocity, is then used as an impacting energy source with an appropriate energy absorber placed between them. The use of this twin sled system overcomes many of the problems identified with the usual horizontal accelerating test configuration, Section 4.1.1, and the inclined plane, Section 4.1.2. The twin sled system simulates the mechanism of a rear end impact but to obtain a realistic impact requires the correct combination of sled masses, bullet sled velocity and energy absorber. A reverse acceleration rig (‘high g’ rig) can also be used to achieve a realistic test more simply but this type of facility was not available to this project.

The TRL DRTF dual sled test again used the TRL modified UN-ECE Regulation 44 test bench [11], as used in the first testing phase of this research programme. Instead of reversing the R44 seat on the sled and driving it backwards it stayed facing forward. A second sled of equal mass was used as the impacting energy source. The target sled was permitted to move forward freely when impacted, being slowly brought to rest once the primary impact had taken place.

4.3 Test seat

Both the DRTF test phases used the TRL modified EC Regulation 44 test bench, as described in Annex 2. The higher support to the occupants back and head restraint were introduced to make it more representative of a vehicle front seat and to reduce the risk of injuring the volunteers by providing better support than the standard bench. The seat back cushion was made from a 70 mm thick layer of closed cell polyethylene foam (EV30)\(^1\) of density 30 kgm\(^3\), chosen for its ability to support the force of the tested occupant without bottoming out (Appendix 2). In addition the back support cushion reacted via force transducers, so that occupant back force could be measured.

4.4 Impact test phases

4.4.1 Hybrid III tests

Several phases of testing with the Hybrid III dummy were undertaken during the project, as it was initially envisaged that this dummy would form the basis for modification(s), if deemed necessary, to make it satisfactory for rear impact spinal injury evaluation. As well as the Hybrid III baseline test described below, the dummy was also evaluated in all of the test configurations detailed in Sections 4.1 and 4.2 for comparison.

\(^1\) Supplied by Zotefoams Plc.
4.4.1.1 Baseline tests
A programme of baseline tests was carried out using the Hybrid III dummy on the TRL DRTF in its normal single sled configuration, Section 4.1. These results were reported in 1999 in an unpublished progress report [12]. Rear impact tests were performed at 4, 8, 16 and 32 km/h to establish mechanical and kinematic performance data. The nominal sled acceleration in each test was 2 g, 3 g, 6 g and 10 g respectively in the 4, 8, 16 and 32 km/h tests. It was planned to compare this data with that obtained from volunteers, and use it to refine and validate the FE (finite element) model of the human spine, Section 5. Figure 1 and Figure 2 show the initial test configuration with a reversed and modified Regulation 44 test bench seat. As already discussed, the seat back was made taller than that used in the Regulation in order to give some further support to the upper body, as experienced in a vehicle seat.

![Figure 1 Hybrid III baseline test, side view](image1)
![Figure 2 Hybrid III baseline test, frontal view](image2)

The data from this programme of testing was used to support the ethical approval process needed in order to undertake volunteer testing and support the computer modelling programme.

4.4.2 Volunteer testing
Prior to this project no rear impact volunteer testing had been performed in the UK. In order to undertake such work it was necessary to obtain permission from the local Ethical Committee. TRL with the assistance of the TRL Doctor, Andrew Whitfield, the Occupational Health and Medical Research Advisor to TRL, procured such permission via a literature review and informative testing with the Hybrid III dummy. As well as obtaining ethical permission and recruiting the volunteers, it was necessary to develop volunteer information sheets and medical assessment forms. Follow up studies were carried out by the TRL Medical Advisor which showed that no injuries had become apparent several months after the programme had been completed.

4.4.3 Whiplash research dummies
Since no dummy has been universally recommended for rear impact research and injury risk predictions, dummies designed for other impact directions have been used, throughout the research community. In recent years some special, low severity, rear-impact dummies have been developed but only one has been developed to a usable form and none have yet been accepted for regulatory applications.

When the project commenced the Hybrid III dummy was the ‘recognised’ frontal impact test device and since rear, in planar terms, is directly opposite to front, it was initially thought that it would be an appropriate test device. However, as the other new dummies became available they were evaluated for their suitability. At the end of the project two of these dummies had been further developed. These
were evaluated to determine if the changes made to them had made them any better for rear impact use.

It should be noted that many of the efforts to develop whiplash dummies have concentrated on producing just a neck to be used with existing frontal dummies. As these existing frontal dummies have rigid thoracic and lumbar spine sections, the modified dummy can only be used to investigate neck injuries. Only the BioRID has a fully flexible 2D spine suitable for assessing injury risk to the full length of the spine. However, no associated injury measurement capability was being proposed for the BioRID below the neck.

4.4.3.1 Hybrid III
The Hybrid III dummy is extensively used as a frontal impact dummy and it has been enhanced with neck force measuring transducers. Some groups have suggested that it is a suitable whiplash dummy since the neck load cells can detect neck forces when it is subjected to a rear impact. Other researchers have strongly suggested that this may only be true for high-energy impacts, when serious neck injury is likely, but not for the low-energy impacts normally associated with whiplash injuries.

4.4.3.2 TRID neck
A special rear impact, whiplash dummy neck, the RID neck, has been developed in Sweden. The RID neck was specifically designed to give a better simulation of the shearing behaviour of the human neck and the ‘S’ movement of the neck described in the literature, whereby the head initially translates with respect to the neck and then rotates as the neck comes under tension. Early use of the RID neck in experimental work confirmed that it was a significant improvement over the standard Hybrid III neck, but that there was still room for improvement. Subsequent development work on the RID neck has been undertaken by TNO in the Netherlands, and is described by Thunnissen et al [13]. The result of that development work was the TRID II neck. The new design was said to give good reproducibility and to agree with published head and neck rotation and other parameters. However, Thunnissen et al. also pointed out that bending of the thoracic spine and the interaction of the torso with the seat back heavily influence head and neck kinematics in rear impacts and what is needed is a biofidelic model of the complete spine [13].

4.4.3.3 BioRID
Sweden then went on to develop a dummy called BioRID. They extended the principles of the RID neck to the whole length of the spine (Figure 3). The dummy has undergone several design iterations and is now commercially available. Studies comparing the performance of the BioRID and other available dummies have been running in parallel to the later stages of this project. A conclusion from a recently published paper was that the BioRID kinematics were different to the Hybrid III with TRID neck in low speed rear impacts when seated in a variety of car seats [14].

Some vehicle manufactures have used the BioRID in the development of specialised rear impact car seats, aimed to minimise whiplash type injuries.
4.4.3.4 Hybrid III/TRID
As noted above TNO developed further the Swedish RID neck into the TRID neck. They then, within the EC Whiplash-1 project, further developed the Hybrid III dummy to give it more human like behaviour. The areas initially developed were a soft foam back, to improve interfacing with the seat, and modified pelvic flesh. The modified pelvic flesh allowed the legs to move at the hip more easily permitting the dummy to straighten allowing the dummy to ‘slide up’ the seat back. This version of the dummy was never evaluated outside of the Whiplash-1 group. By the end of the Whiplash-1 project the dummy had been further developed to incorporate several of the elements of the THOR dummy. In the TRL assessment the TRID neck was tested fitted to a standard Hybrid III dummy.

4.4.3.5 THOR
The THOR dummy [15], was again developed as a step forward in frontal impact protection and many of its features were based on a UK dummy developed in the 1970s, called OPAT [16]. The dummy has been tested in both lateral and rear impacts. If found to be appropriate for the evaluation of improved injury risk in frontal impacts, it is likely that it could become the replacement dummy for the Hybrid III.

5 Mathematical modelling
The literature study, Section 3, clearly showed that knowledge on low-severity rear impact injury was broad but without an in-depth knowledge of injury mechanisms etc. Computer modelling can be a valuable aid in investigating behaviour in areas that can not be adequately investigated through mechanical means, if the model is of sufficient complexity and suitably validated. This is particularly true of biological materials and structures which can not be measured or studied in-vivo for injury or possible injury mechanisms.

Within this project FE methods have been applied to the simulation of the rear impact event, in particular the response of the Hybrid III dummy, as this was the dummy that was the target for possible improvement. The Hybrid III FE model used was that developed by FTSS, the manufacturers of the dummy. The purpose of this part of the study was to investigate the level of correlation that could be achieved between the mathematical model and the Hybrid III dummy in the rear impact tests described in the earlier phases of this project, Section 4.4.1, and to allow closer examination of dummy kinematics not afforded by the test instrumentation. Details of this phase of the TRL project were reported in the first interim report [17]. The severity of the baseline sled tests used to study the correlation between the FE model of the dummy and the real dummy were higher than that which any volunteer could be subjected to, since the object was to study dummy behaviour at expected injury levels. The initial FE study showed that at the higher velocity (32 km/hr) dummy correlation was reasonably good but it was poor at the lower impact severity levels, which were the focus of the research programme. At the lower impact severities the dummy model initial conditions were seen to
be very important and it was found to be difficult to set the FE model of the dummy as per the actual comparative test.

Independently of this project, TRL had developed a FE model of sections of the human thoracic and lumbar spine as part of its human body modelling research [18]. Within this research programme a FE model of the neck was procured and combined with the ‘human spine’ model. The whole spine model was merged into the Hybrid III FE dummy model replacing the original dummy spine elements. This hybrid dummy/human model was then compared against the dummy and volunteer test data.

For any mathematical model to be useful it must incorporate appropriate material characteristics and be validated against appropriate test data. The procurement of appropriate human tissue characteristic data, at the loading rates one might experience in a whiplash-inducing event, was outside of the work programme of Sampson’s project, thus the model was based upon interpreted published data [18].

The hybrid model behaved as one might have expected of a more biofidelic model, with the characteristic ‘S shaped’ curvature of the spine flattening out as the dummy loaded and penetrated the flat seat back.

With the hybrid model it is possible to study predictions of ligament and vertebral disk forces as well as strain at the various spinal levels. These simulation results were reported in the second interim report [17]. Unfortunately it was not possible to validate these predictions and conclusions but it is felt that the sequence of events that they suggest could occur in the real human spine. They suggest relatively high strains at various lumbar and cervical positions, above that predicted at other spinal levels. If strain is a predictor of injury risk then injury in the neck and lumbar region could occur, in the situation evaluated. The high strains appear to be related to localised changes in spinal curvature thus to minimise these ‘excessive strains,’ forced changes in spinal curvature should be avoided.

It can be seen from the above that the hybrid mathematical model is already useful in predicting kinematics and load concentrations. However, the model needs to be developed further, with improved material characteristics appropriate to the strained elements being modelled and at relevant loading rates, before it can be used to predict injury risk. In addition the model does not incorporate muscle reactions, which may well be important in controlling factors present during the development of whiplash injury, so consideration should be given to obtaining the necessary data on muscle reactions and including it in the model.

6 Biofidelity studies

6.1 Spinal behaviour

For a dummy to replicate human behaviour it should interact with the vehicle seat in a human like manner, taking up the natural curvature(s) of the vehicle seat under both normal and impact loading. This aspect is now even more important as some vehicle manufactures are developing active head restraint systems that sense the presence, position and motion of the occupant [19].

In a vehicle seat, different human occupants or dummies could induce seat yielding in different ways thus making comparisons very difficult to quantify. For this study a special seat was constructed to develop dummy design target corridors and assess different test devices. The seat was designed to be featureless, unyielding and instrumented in such a way as to give information that could be used to compare dummy and human behaviour. The seat is described in Section 4.3.

6.1.1 Pressure mapping

A number of advanced occupant position sensing systems are being developed to determine the size, shape and seating posture of the occupant, as a means of optimising the deployment of active restraint systems. It is therefore necessary that a dummy interacts with the vehicle seat in the same way as an occupant would. Pressure distribution or pressure mapping could be one of the sensing systems used.
Although not designed as a crash sensor a large multi-channel Tekscan\(^2\) pressure mat was used by TRL to map dynamically the interaction between the seat back and both the volunteers and the dummies in most of the sled tests. The mat was placed at the interface of the volunteer or dummy and the seat back. The pressure mat that was used had an active area of 480 x 520 mm utilising 1596 pressure nodes and was mounted 50 mm above the seat bite line. Pressure mats of this type are not intended for use in impact studies of this nature, however, they were the best technology available to the study. The maximum sampling rate of the system was 125 Hz which is much lower than ideal for impact sensing and impact studies. As well as being a low frequency sampling system it also has a capacitive delay, therefore synchronising its time history with that of other crash sensors was not possible. Even at this low sampling rate a general indication of differences between humans and anthropometric devices can be observed since the capacitive delay effect and sampling frequencies will be common for all tests.

### 6.1.2 Seat back load cells

In this study the basic seat structure was rigid but the rigid seat framework was covered with slabs of padding material of equivalent stiffness to that used in real world vehicle seats (properties of the foam are given in Appendix 2). Behind the seat padding, mounted on the rigid structure, three specially constructed load plates were placed, dividing the occupants back area into three equally sized horizontal slices, 600mm wide and 195 mm, high. Each of the plates was simply supported on four load cells, as shown in Figure 4. The load cells were slightly modified for the final dummy tests by incorporating an accelerometer at the centre of the plates, between the four force transducers in order to inertia compensate the recorded force measurement. Thus the target corridors produced from the volunteer tests were based on non-inertia compensated data and the final dummy assessment on compensated data. However, these inertia compensations were small in comparison with the measured forces.

![Figure 4: Seat back load cells](http://www.tekscan.com/)

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\(^2\) [http://www.tekscan.com/](http://www.tekscan.com/)
6.2 Volunteer tests

As already explained the volunteers were tested using a dual sled system. The volunteers were seated on a stationary free running sled and a second moving sled was impacted into the rear volunteers sled. An aluminium honeycomb block was placed between the two sleds to limit the amplitude of the decelerating pulse experienced by the volunteer to about 2g. This arrangement in effect replicates a rear-end shunt type accident which is the classic cause of whiplash injuries. The ten volunteers used were male with an average age 26.5 years, height 1.785m and average mass of 77.5kg. After a series of familiarisation tests at a velocity of about 3.5km/h the each volunteers was subjected to a test at 7km/h. Appendix 3 gives details of the honeycomb blocks used and the subjects sled accelerations achieved in the main test programme.

The volunteers were fitted with accelerometers on the head and body (T1 and pelvis), details of the instrumentation used are given in Appendix 3. The head accelerometers were fitted to the volunteers via a close fitting webbing harness with sufficient adjustment to ensure that the transducers were always mounted at a common point. The T1 accelerometers were fitted to a small aluminium bracket and surgically taped to the subject at the T1 location as firmly as could be achieved. The pelvic accelerometer was similarly fitted at the base of the spine above the bony protrusion of the sacrum. All tests were filmed with high-speed cine cameras to obtain kinematic data.

Muscle activity and neck bracing are factors that could influence neck injury risk. Therefore all of the volunteers were equipped with surface mount EMG sensors mounted adjacent to the primary neck muscle groups, sternocleidomastoideus, erector spinae and trapezius, to try to determine if muscle activity could be a factor in the whiplash process.

6.2.1 Head and neck behaviour

All of the volunteers were asked to sit on the test seat in a normal relaxed manner with their heads facing forwards, visually focussing on an object in front of them, which was positioned at eye height. The height of the head restraint was then adjusted to give the occupant support at the level of the posterior pole of the occipital and parietal bones (Section 13). The fore and aft separation was set to approximately 75 mm in the higher speed tests, to minimise gross motion of the head and neck strain. Unfortunately the volunteers were aware that an impact was about to take place due to visual and audible changes in the working environment about them e.g. the turning of the high-speed lights and the noise of the impact sled travelling down the track. These stimuli and the expectation of an imminent impact meant that the distance of the head to the head restraint, when the impact occurred, was not as well controlled as might have been desired. Knowledge of an imminent impact allowed the volunteers to have the opportunity to brace themselves prior to the actual impact, thus controlling free impact kinematics.

One of the features often referred to in the literature, in discussions related to occupant kinematics is the ‘lifting up’ of the occupant in the seat, due to a combination of the straightening up of the spine and the sliding up of the torso on the inclined seat back. In the volunteer programme straightening up of the spine was noted with the occupant’s shoulders moving backwards and upwards. No noticeable motion of the thorax moving up the seat was noted.

The kinematic target corridors, Section 7.2 in general terms show how the volunteers behaved. The rebound behaviour of the subjects was found to be quite varied, although the entire rebound phase is not shown in these curves. In particular, during rebound, some subjects moved forwards well into the seat belt restraint, obviously not tensed, whilst others were clearly tensed and forward excursion was limited. It is not clear how this post impact behaviour could influence whiplash injury risk. The severity of rebound will be a combination of seat frame and cushion stiffness as well as physiological conditioning. Some researchers have suggested that rebound may be the phase of the impact that causes some of the whiplash injuries. If whiplash injury is caused during the rebound phase then the subjects who braced most may in fact be more at risk of sustaining an injury.
Noting the variation in kinematic response within the volunteer data, one can only query the validity of PMHS (post-mortem human surrogate) testing for low severity whiplash type testing, when body tone is absent.

6.2.2 Muscle action
During the volunteer studies the action of the main muscle groups in the neck were studied using surface EMG (electromyography, Section 13) measurements. Preliminary results from this phase of the study were reported in one of the interim project reports [17]. The results were subsequently reviewed with similar observations but with less confidence than that expressed in the preliminary review. This later analysis was presented in the Vehicle Safety 2002 Conference paper [20].

From the recordings that were considered to be the most robust, in the revised analysis [20], it was found that involuntary responses in some of the major muscle groups in the neck were potentially rapid enough to affect the head and neck motion during the impact. The sternocleidomastoideus muscle group (Section 13), according to the EMG recordings, was the most active muscle group during the impact. Initial activity was noted at about 60ms and peak activity at about 80ms. There is evidence to suggest that both the trapezius and the erector spinae (Section 13) also became active during the whiplash motion, at similar time frames. In one subject one of the trapezius muscles activated much earlier (within 10ms) and its opposite muscle at about 30ms. This was not observed in the other volunteer tests but one might hypothesise that the subject may not have been facing fully forward at the time of impact and that these muscles fired very quickly in order to stabilise the head from lateral or torsional movement. These observations support the comments of other researchers, that all the neck muscle groups work in unison, in a purely rear impact, [21].

It is not easy to come to firm conclusions on the potential influence of muscle activity and the risk of whiplash associated disorder injury, based on these limited EMG evaluations. The results suggest that muscle activity could be a factor in stimulating or controlling whiplash injury. Muscle activity does occur very early in the impact suggesting that it is not a slow unimportant process. This observation suggests that whiplash studies with cadaveric material may be misleading, due to the lack of muscle tone. The current study does not give any understanding of what physiologically stimulates the very rapid EMG reactions.

It is noted, when examining time history acceleration measurements for both dummies and volunteers, that the first impact is felt in the lower body at the pelvis. Acceleration pulses are then sequentially observed in the thorax, neck and head. Based on this time sequence one might hypothesise that the early neck muscle activity might be stimulated from the pelvis, the first body part to be physically excited. One might also hypothesise that controlling pelvis acceleration might be a way of attenuating the speed or magnitude of upper body muscle activity.
7 Producing dummy design and assessment targets from biomechanical data

The volunteer’s test data can be used to develop a range of different dummy assessment targets based on the transducer outputs and analysis of the visual media.

A number of techniques can be used to develop dummy performance design targets. Unfortunately this can lead to some confusion and wide variation in the quality of the targets. Some research workers draw a corridor around all of the available data but this tends to produce wide corridors, if one or two responses do not follow the general trend of the data set, increasing the standard deviation. This skew would also make it difficult to identify the mean of the general behaviour without seeing all of the responses. A wide test device corridor can have a number of detrimental effects depending on how they are used. They can lead to the approval of a test device that is not truly representative of the human population. Such a device could lead to the development of safety systems that are inappropriate for the majority of the population. Alternatively they can lead to several test devices being approved which will give different results when used to test the same seat and head restraint system. This second situation would be a serious problem if there is a regulatory approved test.

Therefore EEVC Working Group 9 advocated a statistically based method when they developed the design responses for side impact dummies. The procedure consisted of developing a mean response from the PMHS or volunteer data and then developing a corridor about the mean response with a width of plus and minus one standard deviation at the peak mean value. The curved standard deviation corridor this method produced was then re-defined by an appropriate number of straight lines. This technique clearly defined the mean trends, but elements of the corridor at lower amplitudes were relatively wide. A refinement of this technique, not previously used in the development of target corridors, would be to base the corridor width limits on the continuous mean value, plus and minus one standard deviation. Such a corridor would have a wide band at peak values and of smaller width at low values. One of the big advantages of this latter technique is that it clearly shows the mean target response at all amplitudes along with confidence values. If the origin is easy to define and it is an important feature to be able to define a parameter from a fixed position then the latter technique may be more appropriate, than the former method.

Corridors can be derived from time-shifted data as well as ‘as recorded’. If there is any variation or uncertainty in the initial time \( T_0 \) then it is appropriate to adjust the timing of \( T_0 \). Within the test programme, changes in initial position were observed, particularly with the volunteer’s head to head restraint position. To minimise these positional variations time shifting has been used in developing the following corridors, where appropriate. The following sections show how the data from the volunteer tests have been used to derive proposed whiplash dummy performance corridors to require suitable biofidelic responses. Since all these corridors are derived from the volunteer tests, it follows that these are intended for use in tests with dummies subjected to the same loading regime.

7.1 Transducer targets

As already noted, the volunteers were fitted with accelerometers on the head and body and force transducers were fitted in the seat back, see Appendix 4.

Dummy design and assessment targets or corridors for a whiplash dummy were developed from the acceleration profiles for the volunteer’s head, \( T_1 \) and pelvis level as well as the distribution of force exerted on the seat back. To develop the corridors Roberts used the EEVC WG9 method of plus or minus one standard deviation from the mean \([20]\). The work of Roberts is summarised below.

The data shown in Figure 5 and Figure 6 are for one parameter (\( T_1 \) fore and aft acceleration) and show the variability in volunteer data within this parameter. It should be noted that this was one of the most variable of the volunteer measurements. Figure 5 presents the mean volunteer response with the plus and minus one standard deviation at the peak mean value corridor. Figure 6 shows the straight-line
conversion corridor and all of the volunteer data from which it was derived. Due to the statistical technique used to derive the corridors it is not surprising that some of the volunteer data occur outside it.

![Graph showing志愿者平均响应，正负一个标准差和生成的走廊
for T1前/后向加速度](image)

Figure 5: Volunteer mean response, plus and minus one standard deviation and the generated corridor for T1 fore/aft acceleration

![Graph showing所有志愿者数据和生成的走廊
for the T1 fore/aft acceleration](image)

Figure 6: All the volunteer data and the generated corridor for the T1 fore/aft acceleration

Using the same method, straight-line, volunteer corridors were derived for:
- Head acceleration in the fore and aft (x) and vertical (z) directions as well as the resultant.
- T1 acceleration in the fore and aft direction
- Pelvis acceleration in the fore and aft direction.
Figure 7 and Table 1 show the straight-line corridor for pelvis acceleration with the volunteer responses overlaid and the numerical points defining the corridor. Note that the corridor width for the pelvis fore and aft acceleration increases after the first main peak to plus and minus two standard deviations of that first main peak. This is to encompass the increased variability observed after the first main peak.

**Table 1: Fore and aft pelvis acceleration corridor values**

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<th>Time (ms)</th>
<th>Lower (g)</th>
<th>Upper (g)</th>
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<tr>
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Figure 7: Fore and aft pelvis acceleration

Figure 8 and Figure 9 show the T₁ acceleration corridors, plotted with the volunteer responses from which they were derived, for the fore and aft and vertical directions respectively. Table 2 and Table 3 present the points defining the two corridors.

**Table 2: Fore and aft T₁ acceleration corridor values**

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</table>

Figure 8: Fore and aft T₁ acceleration

Figure 9: Vertical T₁ acceleration
The corridors for fore and aft and vertical head acceleration are shown in Figure 10 and Figure 11 respectively. Again these figures also show the volunteer responses used to develop the corridors. The points used to draw the corridors are shown in Table 4 and Table 5.

**Table 3: Vertical T1 acceleration corridor values**

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**Table 4: Fore and aft head acceleration corridor values**

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Figure 11: Vertical head acceleration

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Table 5: Vertical head acceleration corridor values

Figure 12: Resultant head acceleration

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</tr>
<tr>
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<td>-0.4</td>
<td>0.5</td>
</tr>
<tr>
<td>33</td>
<td>0.6</td>
<td>1.5</td>
</tr>
<tr>
<td>68</td>
<td>0.2</td>
<td>1.1</td>
</tr>
<tr>
<td>80</td>
<td>1.0</td>
<td>1.9</td>
</tr>
<tr>
<td>111</td>
<td>6.6</td>
<td>7.5</td>
</tr>
<tr>
<td>120</td>
<td>5.8</td>
<td>6.8</td>
</tr>
<tr>
<td>145</td>
<td>1.4</td>
<td>2.4</td>
</tr>
<tr>
<td>160</td>
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<td>1.7</td>
</tr>
<tr>
<td>500</td>
<td>-0.3</td>
<td>0.5</td>
</tr>
</tbody>
</table>

Table 6: Resultant head acceleration corridor values

Figure 12 and Table 6 show the corridor and corridor values for the resultant, of the fore and aft and vertical directions, head acceleration. Also plotted in Figure 12, are the volunteer responses.

7.2 Kinematic targets

All tests were filmed from the side using two high-speed high-resolution cine cameras running at 400Hz and a high-speed low-resolution digital camera, which ran at a frame frequency of 1000 Hz. One of the cine cameras was positioned several metres from the target trolley to give a wide-angle view of the whole trolley, minimising parallax distortion problems. The second film camera was mounted on the impacted trolley, in order to give a lateral, close-up view of the subject’s head and neck motion. The kinematic data were derived from this close-up, on-trolley view. This analysis has not previously been published. The proposed corridors were derived using the alternative statistical method described in Section 7 with a width of plus and minus one continuous standard deviation about the mean of the data. This method was used as it appeared better suited to the data.
Targets were attached to the volunteer or dummy at various strategic positions. On the side of the head, small 28 mm targets were placed at the centre of gravity. Targets were also placed on the front, top and back of the head medially on the instrumentation straps and one or two other targets on the side of the head, on the Frankfort plane when possible (Section 13). For T1 targets, two small circular targets were stuck to an aluminium strip, which in turn was attached to the accelerometers at the external T1 position.

Note:- The tests were recorded on high quality film based cameras but due to difficulties in analysing this high quality film media, the kinematic corridors are derived from an analysis of much lower quality, reduced resolution digitised compressed .avi files produced from the films. This method of analysis limits the quality of the corridors when motion is not large compared to the whole picture area. A comparison was carried out between the two analysis sources and it was found that the general shape of the responses were comparable and valuable for this kinematic investigation. However, the oscillations visible in some of the responses in smaller displacement graphs are a product of the lack of resolution within the digitised image.

Figure 13 and Figure 14 show the T1 displacement corridors, plotted with the volunteer responses from which they were derived, for the fore and aft and vertical directions respectively. Table 7 and Table 8 present the points defining the two corridors.
The rotation corridor for T1 is shown in Figure 15 with the volunteer responses. The corridor limits are given in Table 9.

![Figure 14: Vertical T1 displacement](image1)

![Figure 15: T1 rotation](image2)

**Table 8: Vertical T1 displacement corridor values**

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>Lower (mm)</th>
<th>Upper (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
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<tr>
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<td>-10</td>
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<tr>
<td>175</td>
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<tr>
<td>185</td>
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<td>23</td>
</tr>
<tr>
<td>240</td>
<td>-16</td>
<td>-1</td>
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<tr>
<td>355</td>
<td>-4</td>
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<td>375</td>
<td></td>
<td>24</td>
</tr>
<tr>
<td>500</td>
<td>-10</td>
<td>11</td>
</tr>
</tbody>
</table>

**Figure 14: Vertical T1 displacement**

**Table 9: T1 rotation corridor values**

<table>
<thead>
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<th>Time (ms)</th>
<th>Lower (°)</th>
<th>Upper (°)</th>
</tr>
</thead>
<tbody>
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<td>0</td>
<td>0</td>
</tr>
<tr>
<td>25</td>
<td>-5</td>
<td>0</td>
</tr>
<tr>
<td>60</td>
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<td>5</td>
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<td>95</td>
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<tr>
<td>100</td>
<td></td>
<td>23</td>
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<td>115</td>
<td>18</td>
<td></td>
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<tr>
<td>130</td>
<td>28</td>
<td></td>
</tr>
<tr>
<td>150</td>
<td>14</td>
<td></td>
</tr>
<tr>
<td>185</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>280</td>
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</tr>
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<td>305</td>
<td>-5</td>
<td></td>
</tr>
<tr>
<td>500</td>
<td>-5</td>
<td>12</td>
</tr>
</tbody>
</table>

**Figure 15: T1 rotation**

**T1 vertical limits**

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>Lower (mm)</th>
<th>Upper (mm)</th>
</tr>
</thead>
<tbody>
<tr>
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<td>355</td>
<td>-4</td>
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<tr>
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<td>24</td>
</tr>
<tr>
<td>500</td>
<td>-10</td>
<td>11</td>
</tr>
</tbody>
</table>
The corridors for fore and aft and vertical head acceleration are shown in Figure 16 and Figure 17 respectively. Again these figures also show the volunteer responses used to develop the corridors. The points used to draw the corridors are shown in Table 10 and Table 11.

**Head fore and aft limits**

<table>
<thead>
<tr>
<th>Time (ms)</th>
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<th>Upper (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0</td>
<td>-9</td>
</tr>
<tr>
<td>35</td>
<td>-10</td>
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<tr>
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<td>0</td>
<td>-86</td>
</tr>
<tr>
<td>100</td>
<td>-115</td>
<td>-86</td>
</tr>
<tr>
<td>110</td>
<td>-115</td>
<td>-86</td>
</tr>
<tr>
<td>125</td>
<td>-115</td>
<td>-86</td>
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<td>135</td>
<td>-115</td>
<td>-86</td>
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<tr>
<td>205</td>
<td>-41</td>
<td>30</td>
</tr>
<tr>
<td>235</td>
<td>-17</td>
<td>69</td>
</tr>
<tr>
<td>335</td>
<td>-12</td>
<td>55</td>
</tr>
</tbody>
</table>

Figure 16: Fore and aft head displacement

Table 10: Fore and aft head displacement corridor values

**Head vertical limits**

<table>
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<th>Time (ms)</th>
<th>Lower (mm)</th>
<th>Upper (mm)</th>
</tr>
</thead>
<tbody>
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<td>0</td>
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<tr>
<td>20</td>
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<td>35</td>
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<tr>
<td>125</td>
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</tr>
<tr>
<td>130</td>
<td>23</td>
<td>23</td>
</tr>
<tr>
<td>150</td>
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<tr>
<td>190</td>
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<tr>
<td>245</td>
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<td>385</td>
<td>-12</td>
<td>8</td>
</tr>
<tr>
<td>455</td>
<td>-15</td>
<td>8</td>
</tr>
</tbody>
</table>

Figure 17: Vertical head displacement

Table 11: Vertical head displacement corridor values
Figure 18 and Table 12 show the corridor and corridor values for head rotation. Overlaid on the corridor are the individual volunteer responses.

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>Lower (°)</th>
<th>Upper (°)</th>
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</tr>
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<td></td>
</tr>
<tr>
<td>80</td>
<td>0</td>
<td></td>
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<tr>
<td>125</td>
<td>10</td>
<td>20</td>
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<tr>
<td>145</td>
<td>10</td>
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<tr>
<td>195</td>
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<tr>
<td>220</td>
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<td>0</td>
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<tr>
<td>285</td>
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</tr>
<tr>
<td>475</td>
<td>-15</td>
<td>5</td>
</tr>
</tbody>
</table>

Table 12: Head rotation corridor values

The T1 subtracted from the head displacement corridors are shown in Figure 19 and Figure 20, for the fore and aft and vertical directions respectively. Table 13 and Table 14 contain the values used to define the corridor limits. It should be noted that the head displacements were measured from the head centre of gravity marker and the T1 displacements from a marker attached to the accelerometer positioned at the back of the neck. Therefore the differences between head vertical displacement and T1 shown in Figure 20 are due to a combination of neck extension, change in nominal neck angle and the off-set between the two markers. This graph is not implying that the necks of the volunteers stretched by about 30 mm.

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>Lower (mm)</th>
<th>Upper (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>30</td>
<td>0</td>
<td>5</td>
</tr>
<tr>
<td>40</td>
<td>-25</td>
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<tr>
<td>115</td>
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<td>-46</td>
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<td>125</td>
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<td>270</td>
<td></td>
<td>55</td>
</tr>
<tr>
<td>455</td>
<td></td>
<td>-30</td>
</tr>
</tbody>
</table>

Table 13: Fore and aft head – T1 displacement corridor values
The angle through which the head moves with respect to T1 response for each of the volunteers is shown in Figure 21 along with the corridor. The corridor limits are given in Table 15. The T1 to head angle is not corrected for rotation of the occupant overall and as such, incorporates the influence of any rotation from the pelvis up.

Figure 20: Vertical head – T1 displacement

Table 14: Vertical head – T1 displacement corridor values

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>Lower (mm)</th>
<th>Upper (mm)</th>
</tr>
</thead>
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<tr>
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<tr>
<td>25</td>
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<td>-1</td>
</tr>
<tr>
<td>35</td>
<td>-2</td>
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<td>9</td>
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<tr>
<td>115</td>
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<tr>
<td>140</td>
<td></td>
<td>35</td>
</tr>
<tr>
<td>155</td>
<td>12</td>
<td></td>
</tr>
<tr>
<td>180</td>
<td>-4</td>
<td>10</td>
</tr>
<tr>
<td>235</td>
<td></td>
<td>20</td>
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<td>315</td>
<td>-34</td>
<td></td>
</tr>
<tr>
<td>500</td>
<td>-34</td>
<td>20</td>
</tr>
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</table>

Figure 21: T1 to head angle

Table 15: T1 to head angle corridor values

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>Lower (°)</th>
<th>Upper (°)</th>
</tr>
</thead>
<tbody>
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<td>0</td>
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<td></td>
</tr>
<tr>
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<td>-1</td>
<td>0</td>
</tr>
<tr>
<td>60</td>
<td></td>
<td>5</td>
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<td>80</td>
<td>5</td>
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<tr>
<td>105</td>
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<td>190</td>
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<tr>
<td>280</td>
<td></td>
<td>3</td>
</tr>
<tr>
<td>285</td>
<td>-15</td>
<td></td>
</tr>
<tr>
<td>500</td>
<td>-15</td>
<td>10</td>
</tr>
</tbody>
</table>
Trajectory corridors were also derived for head minus T₁ displacement in the vertical direction against the fore and aft, Figure 22 and Table 16, as well as the T₁ to head angle against fore and aft displacement, as shown in Figure 23 and Table 17. The volunteer trajectories for each of these graphs are shown with the corridor and also include some of the rebound phase although the corridor only considers the rearward loading phase.

![Figure 22: Head – T₁ displacement trajectory](image)

<table>
<thead>
<tr>
<th>Head – T₁ displacement trajectory limits</th>
</tr>
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<tbody>
<tr>
<td>Fore and aft (mm)</td>
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<tr>
<td>-------------------</td>
</tr>
<tr>
<td>0</td>
</tr>
<tr>
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<tr>
<td>-79</td>
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<tr>
<td>-79</td>
</tr>
<tr>
<td>0</td>
</tr>
</tbody>
</table>

![Table 16: Head – T₁ displacement trajectory corridor values](image)

![Figure 23: Head – T₁ angle and displacement trajectory](image)

<table>
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<tr>
<th>Head – T₁ angle/displacement trajectory limits</th>
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</thead>
<tbody>
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<td>Fore and aft (mm)</td>
</tr>
<tr>
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</tr>
<tr>
<td>-78</td>
</tr>
<tr>
<td>-78</td>
</tr>
<tr>
<td>0</td>
</tr>
</tbody>
</table>

![Table 17: Head – T₁ angle and displacement trajectory corridor values](image)
7.3 Seat based assessment targets

7.3.1 Seat forces

As noted in Section 6.1.2 the design of the seat back load cells changed slightly during the test programme with the target corridors being set with non-inertia compensated load cells and the final dummy assessments with inertia compensated data. The dummy force measurements are therefore of a higher quality than that obtained from the volunteer analysis. Since the inertia compensation was small and the volunteers, by their very nature, are ‘variable’ and the target corridors have been derived statistically from several tests it is felt that the comparison between the two types of measured data is valid.

Seat back panel force corridors are shown in the following four figures and accompanying tables with the mean and the mean plus or minus one standard deviation at peak, derived from the volunteer data. Figure 24 and Table 18 show the corridor for the top seat back panel force.

![Figure 24: Top seat back panel force](image)

<table>
<thead>
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<th>Time (ms)</th>
<th>Lower (kN)</th>
<th>Upper (kN)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.12</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>0.3</td>
<td></td>
</tr>
<tr>
<td>89</td>
<td>0.65</td>
<td>0.89</td>
</tr>
<tr>
<td>99</td>
<td>0.65</td>
<td>0.89</td>
</tr>
<tr>
<td>110</td>
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<td>132</td>
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<td>138</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>168</td>
<td>0.1</td>
<td></td>
</tr>
<tr>
<td>200</td>
<td>0.1</td>
<td></td>
</tr>
</tbody>
</table>

Table 18: Top seat back panel force corridor values
Figure 25 and Table 19 show the corridor for the middle seat back force.

![Figure 25: Middle seat back panel force](image)

**Middle seat back panel force limits**

<table>
<thead>
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<th>Time (ms)</th>
<th>Lower (kN)</th>
<th>Upper (kN)</th>
</tr>
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<td>0.7</td>
<td>0.7</td>
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<tr>
<td>5</td>
<td>0.7</td>
<td>0.7</td>
</tr>
<tr>
<td>13</td>
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<td>50</td>
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<td>0.29</td>
<td>0.46</td>
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<td>106</td>
<td>0.12</td>
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</tr>
<tr>
<td>140</td>
<td>0</td>
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</tr>
</tbody>
</table>

Table 19: Middle seat back panel force corridor values

Figure 26 and Table 20 show the corridor for the bottom seat back panel force.

![Figure 26: Bottom seat back panel force](image)

**Bottom seat back panel force limits**

<table>
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<th>Time (ms)</th>
<th>Lower (kN)</th>
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<td>0</td>
<td>0.12</td>
<td>0</td>
</tr>
<tr>
<td>27</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>85</td>
<td>0.62</td>
<td>0.9</td>
</tr>
<tr>
<td>95</td>
<td>0.62</td>
<td>0.9</td>
</tr>
<tr>
<td>120</td>
<td>0.55</td>
<td></td>
</tr>
<tr>
<td>135</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>200</td>
<td>0.1</td>
<td></td>
</tr>
</tbody>
</table>

Table 20: Bottom seat back panel force corridor values
The corridor for the sum of the forces from the three individual seat back panels is shown in Figure 27 and Table 21.

<table>
<thead>
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<th>Time (ms)</th>
<th>Lower (kN)</th>
<th>Upper (kN)</th>
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</thead>
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<td>0.27</td>
<td>0.27</td>
</tr>
<tr>
<td>20</td>
<td>0</td>
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<td>59</td>
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<td>1.47</td>
<td>2.1</td>
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<tr>
<td>145</td>
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<td>0.34</td>
</tr>
<tr>
<td>200</td>
<td>0.09</td>
<td>0.09</td>
</tr>
</tbody>
</table>

Figure 27: Total seat back panel force

**Table 21: Total seat back panel force corridor values**

7.3.2 Pressure mapping

Pressure mapping data are visually very informative, when comparing different test subjects or dummies. However, the volunteer’s pressure mapping responses can not be readily converted into design performance corridors as there are four variables to display; position in both x and y, pressure and time. The best that might be achieved is to develop an area criterion but this has not been undertaken within the project. In addition the high rate sensitivity of the system’s loading capabilities are not constant with the requirements of biomechanical impact, thus the data can only be used for general comparative purposes.

Figure 28 shows the typical pressure map data for two of the subjects. Only small variations are noted from these typical traces which are thought to be due to slight changes in initial seating posture. The scale, relating colour in the Tekscan images to pressure, in kPa, is shown in Figure 29.

![Figure 28: Typical pressure map data for volunteers and dummies](image-url)
8 Dummy evaluations

Three phases of dummy evaluation took place within the research programme. The first single sled phase provided the baseline data for the Hybrid III dummy and was used to help select the severity of the volunteer tests. The second dual sled phase examined the performance of the, then available, advanced dummies that had been or could have been used for rear impact studies. The third phase of testing was also dual sled testing and was to evaluate the latest versions of the previously evaluated dummies and the original Hybrid III dummy.

The second phase of dummy comparisons, based on transducer data was published in the Vehicle Safety 2002 paper [20]. No kinematic data were presented in this publication. The following shows the results included in the Vehicle Safety paper along with the kinematic assessments and the performance of the latest versions of the dummy, which were tested towards the end of this project.

The following sections present all of the dummy test data from the second and third phases compared with the volunteers, which were all performed using the dual sled facility, described in Section 4.2. All of the tests reported below were of the same severity as used in the main volunteer test programme. A few higher severity tests were carried out with some of the dummies, however, the results have not been reported due to the lack of suitable higher severity biomechanical tests with which to compare them.

In order to make the comparison as unbiased as possible, the same external head accelerometers harness and external T1 accelerometers mounting were used on the dummies as had been used on the volunteers, instead of the normal internal dummy mountings.
### 8.1 Seat based assessments

#### 8.1.1 Pressure map

The Tekscan pressure mat readings from each test give a qualitative representation of how each dummy interacts with the seat back. Images from each pressure-mapped sequence have been extracted and four images from the time histories of one test with each different dummy are compared with a typical volunteer’s pressure map in Figure 30 to Figure 32. Due to synchronisation differences between tests the pressure mapping snap-shots show general comparative trends for common time intervals. The shades and colours are equivalent for all tests, see the colour key in Figure 29.

<table>
<thead>
<tr>
<th>Time (ms)</th>
<th>0</th>
<th>40 - 80</th>
<th>64 - 104</th>
<th>96 - 152</th>
</tr>
</thead>
<tbody>
<tr>
<td>Typical Volunteer</td>
<td><img src="image1.png" alt="Image" /></td>
<td><img src="image2.png" alt="Image" /></td>
<td><img src="image3.png" alt="Image" /></td>
<td><img src="image4.png" alt="Image" /></td>
</tr>
<tr>
<td>Hybrid III</td>
<td><img src="image5.png" alt="Image" /></td>
<td><img src="image6.png" alt="Image" /></td>
<td><img src="image7.png" alt="Image" /></td>
<td><img src="image8.png" alt="Image" /></td>
</tr>
<tr>
<td>Hybrid III with TRID neck</td>
<td><img src="image9.png" alt="Image" /></td>
<td><img src="image10.png" alt="Image" /></td>
<td><img src="image11.png" alt="Image" /></td>
<td><img src="image12.png" alt="Image" /></td>
</tr>
</tbody>
</table>

**Figure 30: Volunteer and Hybrid III Tekscan pressure mat comparison.**

The Tekscan images clearly show changes in pressure distribution from one time period to another and between dummies. The volunteer pressure maps indicate a relatively flat pressure distribution with the seat with the higher-pressure areas corresponding to the shoulder/scapula and the pelvic wings.
Generally, the Hybrid and THOR dummies show high pressure areas along the line of the spine with some late pressure build up around the pelvis. An obvious feature of the THOR $\alpha$, is the curved spine shape. The diagonal line of pressure comes from the instrumentation umbilical cable exiting from under the THOR vest. This effect can be seen with all three THOR versions tested. The offset shoulder pressure seen in the THOR $\alpha$ test may be due to the channelling of this cabling behind the vest. It was not possible to position the THOR to achieve a vertical spine and equal shoulder pressures, however, the set-up shown in these pictures was repeatable. The seat back interaction from the THOR $\alpha$ does show an improvement over the previous versions tested with the pressure in the thorax region spread over a greater area, although still not as large an area as seen with the volunteers. There are still points of high pressure in the lumbar region, mainly due to the course taken by the instrumentation umbilical.

Figure 31: Volunteer and THOR Tekscan pressure mat comparison.
The BioRID dummy shows a relatively flat pressure mapping much closer to that of the volunteers compared to either the Hybrid III or THOR profiles. The lateral pressure line with the BioRID is due to the instrumentation umbilical cable, which in these tests was placed behind the dummy. This is not observed in BioRID 2a, when the cable was repositioned, providing a smoother pressure distribution over the whole surface of the thorax. The localised, higher pressure, regions remain close to the posterior iliac spine position, as seen with the volunteers.

8.1.1.1 Seat back forces

8.1.1.1.1 Individual seat back panels
The load on the three separate force plates in the seat back was calculated by summing the outputs from the four load cells mounted at the corners of the load plates. The mean results from the final dummy tests for each individual force plate are shown Figure 33 to Figure 35. It was not possible to calculate the load recorded on the bottom force plate for the Hybrid III tests due to a load cell failure, also preventing the summation of forces over all three plates. It should also be noted that the dummy results have had the seat back forces inertia compensated, whereas this was not performed for the volunteer data from which the corridors were derived. However, due to the variation inherent with volunteer tests and the ten tests sample size from which the mean was calculated, it is felt that the comparatively small effects of the inertia compensation have largely been negated. Therefore, whilst comparison of the individual responses of the volunteers may not be valid, the force corridors derived from the volunteer tests are still valuable as dummy performance targets. The following figures give the force time histories for the latest versions (at the time of the study) of the advanced dummies compared with the standard Hybrid III and the volunteers’ corridors.
Figure 33: Top seat back panel force.

Figure 34: Middle seat back panel force.

Figure 35: Bottom seat back panel force.
An obvious feature displayed by the individual force plate results is that all three dummies exhibit peak forces on the middle back plate at least twice as high as that observed in the volunteer tests exceeding the upper corridor maximum value. It is suggested that a variation in initial seating postures may be the reason for the differences seen between the human and dummy results. Both the BioRID and THOR have total seat back forces similar to those of the volunteers. These results appear to be confirmed by the Tekscan pressure map, which shows the volunteer to have high pressure points at the top and bottom as opposed to the middle.

### 8.1.1.1.2 Combined seat back panels

The total force recorded by all three seat back force plates is shown in Figure 36.

![Figure 36: Total seat back panel force.](image)

The mean total seat back forces recorded in tests with the THOR α and BioRID 2a shows too rapid a rise in seat force when compared with the volunteers’ corridor. The THOR α also gives slightly too high a peak force on the seat back whereas the BioRID 2 has a peak force value within the peak values from the upper and lower corridor limits.

Whilst the THOR α records too high a total force compared with the volunteer corridor, of the three dummies investigated in this manner, it shows results closest to the volunteers, based on the individual force plates. However, this is only a marginal performance difference when considered with the general trend of all the dummies giving too great a load on the middle force plate.

### 8.2 Transducer data

As the dummy and occupant is driven backwards into the seat the thorax and pelvis will be the first body parts to be loaded, followed by the head. If the torso behaves poorly then it is likely that the neck and head will not behave in a human like manner, even though the relative body parts response may appear to be good.

#### 8.2.1 Fore and aft pelvis accelerations

At the pelvis level, the Hybrid III exhibits properties that might be associated with the pelvis and lower back being too stiff for biofidelic interaction with the seat back. Figure 37 shows the pelvis fore aft acceleration for both the Hybrid III and also a Hybrid III with the alternative TRID neck. The
Hybrid III version, with an original neck, peaks too early and also has a greater maximum acceleration. After the initial contact with the seatback, it also continues to rock forward and back. The modification incorporating a TRID neck appears to have improved the pelvis behaviour significantly, although the effects of variations in test set-up are not known. The TRID neck modified Hybrid III has a peak in fore and aft pelvis acceleration only 5 ms too early, compared with the corridor. Apart from the peak occurring too early, its value is between the peak corridor limits and the rates of increase and decrease of acceleration are similar to those given by the volunteer results. Also, unlike the unmodified version, the rebound stays within the corridor.

![Graph showing Hybrid III fore and aft pelvis acceleration](image)

Figure 37: Hybrid III fore and aft pelvis acceleration.

The pelvis fore aft responses for the THOR dummy, THOR fitted with the EuroSID neck and THOR $\alpha$ are shown in Figure 38. All three dummies have a peak value close to the corridor limits, none of them however, show the stepped acceleration increase as seen in the volunteer corridor profile. The THOR fitted with the EuroSID neck has a response similar in general shape to that required, but it is consistently behind the volunteer corridor in time. The THOR $\alpha$ has the least promising response with the initial acceleration rise being too fast suggesting that it is too stiff.

Unfortunately the BioRID and BioRID 2a were not fitted with a pelvis accelerometers, therefore no evaluation can be made with respect to the volunteers or any improvements from version to version at this level within the dummy.
8.2.2 Fore and aft T₁ accelerations
Both the original Hybrid III and the Hybrid III with the TRID neck were seen to have fore and aft accelerations that peak too early, as measured at the T₁ height on the spine. Figure 39 shows the fore aft accelerations at T₁ compared with the corridor. It can be seen that the original Hybrid III neck has a maximum value closer to that of the volunteer corridor, whereas the Hybrid III with a TRID neck peaks below the specified range. The peak for the Hybrid III T₁ fore and aft acceleration occurs at about the same time as the peak in fore and aft pelvis acceleration and the Hybrid III with a TRID neck has a peak before its pelvis peak, see Figure 37.

Figure 40 shows the neck fore aft accelerations for the three THOR dummy configurations. Only the THOR with the EuroSID neck has a measured peak acceleration value that is outside the limits given by the volunteer corridor. All of the THOR versions peak too early by about 50 ms, when compared with the corridor. As with the Hybrid III, the THOR dummies’ peaks in T₁ fore and aft acceleration occur around the same time as the pelvis peaks, therefore the problem with the response can not be said to follow up the spine from the pelvis. Following the peak acceleration, the acceleration gradient for all dummy versions is consistent with the corridor, although the EuroSID neck version shows some oscillation, taking it outside of the corridor.

The T₁ fore and aft acceleration for the BioRID and BioRID 2a is shown in Figure 41. At the T₁ level, the BioRID 2a has a peak value closer to the volunteer data than the BioRID. However, the profile of the BioRID acceleration is closer to the volunteer corridor. It shows a slower increase to its peak acceleration value and then a sharper decrease than the version 2a, fitting more closely with the corridor profile. Unfortunately no pelvis acceleration data were available for the pelvis of the BioRIDs with which to compare temporal aspects of the pelvis and T₁ acceleration responses.
Figure 39: Hybrid III fore and aft T₁ acceleration.

Figure 40: THOR fore and aft T₁ acceleration.

Figure 41: BioRID fore and aft T₁ acceleration.
8.2.3 Head accelerations
The fore and aft head accelerations for the Hybrid III and Hybrid III fitted with a TRID neck are shown in Figure 42. Figure 43 and Figure 44 show the vertical and resultant Hybrid III head accelerations. Lateral accelerations are not presented, as this is a simple fore/aft impact with negligible lateral acceleration components.

![Figure 42: Hybrid III fore and aft head acceleration.](image)

![Figure 43: Hybrid III vertical head acceleration.](image)
The fore and aft peak acceleration value for the Hybrid III is within the range generated by the volunteer corridor, although slightly too early by about 5 ms. The Hybrid III with a TRID neck has a lower peak value, about two thirds of that of the normal Hybrid III. It also peaks after the volunteer corridor by about 20 ms. The TRID neck version does see a decreased rebound peak, delaying that feature about 60 ms after the volunteers.

The fore aft head accelerations of the three versions of the THOR are compared with the volunteer corridor and shown in Figure 45. Vertical and resultant head accelerations can be found in Figure 46 and Figure 47.

Figure 44: Hybrid III resultant head acceleration.

Figure 45: THOR fore and aft head acceleration.
With the exception of the THOR and THOR α vertical, all of the THOR dummies tested have peak head acceleration values that are approximately correct in terms of time when compared with the corridor requirements. The increase to and decrease from these peaks also occur at approximately the right time. As with the Hybrid III dummies this represents a delay, produced by the neck, from the time of the peaks in the T1 accelerations, with respect to the volunteer corridor timings. However, the values of the head acceleration peaks, for all versions of the THOR, fail to fall within the range specified by the volunteer corridor, with the EuroSID neck version being too low and the supplied THOR necks being too high. Coupled with the vertical acceleration results where the EuroSID neck version has too high a peak value, the implied difference between this and the other two THOR versions is that the head was allowed to rotate rearwards more before contacting the head restraint. A slightly greater head rotation can be seen in the kinematics from the films, which may confirm this, although it is interesting to note that the THOR with an EuroSID neck also has a lower resultant head acceleration than the other two versions suggesting an overall head acceleration attenuation effect.

The BioRID head acceleration in the fore aft direction is shown in Figure 48 and the vertical and resultant head accelerations are shown in Figure 49 and Figure 50.
Figure 48: BioRID fore and aft head acceleration.

Figure 49: BioRID vertical head acceleration.

Figure 50: BioRID resultant head acceleration.
The original BioRID is seen to have a peak fore and aft and resultant acceleration values just greater than the volunteer corridor and about 20 ms too late. Whilst the vertical acceleration peak is slightly lower than the corridor and at the same time. The alterations made to the BioRID version 2a appear to have brought this response more in line with the target requirements. The peak fore and aft acceleration is now within the corridor limits and occurs at the correct time. The only minor deviation from the corridor is in the width of the peak fore aft and resultant acceleration, having a slightly longer duration, but this is not considered to be a significant problem. However, the 2a positive vertical acceleration is now slightly too early and too low.

8.3 Kinematic data

The dummy’s movements were recorded using the same method as was used for the volunteer tests, with the same analysis software and resolution of avi files, produced from cine films, as described in Section 7.2. The close-up view was analysed because it gave the greatest accuracy, however, this limits the possible comparisons to just T1 and head parameters, as the pelvis was out of shot.

The results are presented with T1 parameters first, as the performance at the head is dependent on the neck and the performance of the neck dependent on the input from the thorax.

8.3.1 T1 displacements

With either neck the Hybrid III produces a much smaller rearward displacement, at T1, than the volunteers, but the Hybrid III with TRID neck is closer to the volunteer corridor, see Figure 51. For T1 vertical displacements, it is also the TRID neck variant which is closer to the volunteer corridor, shown in Figure 52. However, neither Hybrid III version produced displacements that were very similar in shape to those of the volunteers.

![Figure 51: Hybrid III fore and aft T1 displacement.](image-url)
For T₁ fore and aft displacement, Figure 53, none of the three THOR dummy versions tested show the desired amplitude of peak motion in the fore and aft direction or the timing of the peak given by the volunteer corridor. However, the fore and aft displacement of the THORs were similar in form to the volunteers. The closest to the corridor is the THOR α, which peaks at around 80% of the lower corridor limit. In the vertical direction, Figure 54, none of the THOR versions tested gave negative and then positive peaks of correct magnitude to fit within the volunteer corridor between 100 and 200 ms. With the THOR and THOR with EuroSID neck, it appears that the responses are both too small in amplitude and too early, although the small scale of the graph must be considered.
It can be seen in Figure 55, that the BioRID 2a has slightly too small a rearward movement at T₁, compared with the volunteers. The BioRID peak value fits within the magnitude limits of the corridor, despite the peak occurring slightly after the volunteer’s. The fore and aft displacements of both versions of BioRID are of a very similar form to that of the volunteer corridor. BioRID displacements at T₁ in the vertical direction, Figure 56, show very little similarity with the volunteer corridor. The BioRID 2a shows negligible movement until well into the rebound phase.
8.3.2 Head displacements

Figure 57 shows the fore and aft displacement of the Hybrid III head with both the original and TRID necks. The Hybrid III with its original neck gives fore and aft displacements that are smaller than that given by the upper limit of the volunteer corridor in the first phase of loading, but they are similar in form. The addition of the TRID neck has increased the peak recorded displacement value, bringing it closer to the volunteer corridor. However, the result of this increase, has delayed the timing of the peak, reducing the velocity of rearward head motion. This is the expected behaviour from the TRID neck, which is less rigid than the Hybrid III neck.

In the vertical direction, Figure 58, neither of the Hybrid III versions demonstrated the required vertical head displacement. However, the Hybrid III with the original neck shows greater vertical head movement peaking at the correct time but with a peak value less than half of the corridor lower limit.

Figure 56: BioRID vertical $T_1$ displacement.

Figure 57: Hybrid III fore and aft head displacement.
The fore and aft head displacement graph for the three versions of the THOR dummy is shown in Figure 59. The THOR and THOR $\alpha$ have negative peak values within the limits of the volunteer corridor although the velocity of forward motion is too great. The THOR with the EuroSID neck has a rebound velocity more like the volunteers but a peak rearward displacement that is too small. Some of this decrease in rearward displacement may be produced by the extra rotation of the head with the EuroSID neck as identified from the transducer data in Section 8.2.3.

The three THOR dummy responses for vertical head displacements are shown in Figure 60. None of the three show the required vertical displacement seen with the volunteers. The THOR and THOR $\alpha$ show the opposite of what is required with significant negative displacements, the THOR $\alpha$ peaking at $-14$ mm. A negative vertical displacement would be expected if the head was pivoting rearwards on a neck of almost constant length. In this way it can be seen that the THOR neck offers no, or very little, vertical motion.
Figure 60: THOR vertical head displacement.

Figure 61 shows the fore and aft displacements for the BioRID head. The rearwards head displacement of the original BioRID has a greater peak value than the volunteer corridor, whereas the BioRID 2a has a smaller negative peak. This change was not expected when considering the changes in the dummy designs. The difference in magnitude of the peaks can be explained through the starting position of the head with respect to the head restraint, with the BioRID sitting further from the restraint than the BioRID 2a. The head to head restraint distance difference is confirmed by the head contact times which were, for the BioRID, 95 to 98 ms, as opposed to 68 to 73 ms for the BioRID 2a, which is more like the head contact times of the volunteers between 58 and 88 ms. Even the 70 mm difference between peak values, as observed, could be accounted for by this difference between test set-ups, with both peak values occurring approximately 50 mm after head contact.

A significant positive vertical head displacement is seen with the BioRIDs in Figure 62. Both dummies produced a vertical velocity comparable with that from the vertical rise of the volunteers, however, the magnitude of the BioRID vertical displacement peak is still only about 40 % of the volunteer lower corridor limit. The BioRID response is almost identical to the Hybrid III vertical head motion, whereas the BioRID 2a maintains a positive vertical head displacement for a longer duration. The reason for this is not fully understood, at present, and requires further investigation of the head to head restraint interaction and possibly an in-depth knowledge of the changes made in the production of the BioRID 2a. Overall it appears that the two versions of the BioRID follow the form of the corridor most closely of all the dummies, even though the peak values do not match the corridors. The fact that the BioRID gave too large fore and aft displacements in the first phase and the BioRID 2a gave too little suggest that a smaller adjustment would have resulted in it matching the corridors.
8.3.3 Fore and aft displacements of heads related to T₁

From consideration of the head motion with respect to the marker mounted on the T₁ accelerometers, it is possible to investigate the performance of the dummies’ necks, more closely.

The delay observed in the Hybrid III rearward head displacement peaks is also evident in the relative motion of the head with respect to T₁, Figure 63. With both necks, the peak rearward displacement of the Hybrid III head relative to the T₁ marker is within the corridor limits, however, the TRID neck has moved the response to occur after the corridor, whereas the original Hybrid III very closely follows the leading edge of the corridor.
Figure 63: Hybrid III fore and aft displacement of the head relative to T1.

Figure 64 shows the relative displacement of the head with respect to the marker mounted on the T1 accelerometers for the THOR, THOR with EuroSID neck and the THOR α compared with the corridor as generated by the volunteer behaviour. All three dummies have peak rearward displacements within the corridor limits, however again the increase in positive displacement is outside of the corridor, as their rebound velocity is too great. This performance is very similar to that shown by the Hybrid III neck, only with a smaller peak value for relative rearward displacement. The construction of the THOR neck is very similar to that in the Hybrid III, so similar performance between T1 and the head was expected.

Figure 64: THOR fore and aft displacement of the head relative to T1.

Considering the relative x-direction displacement of the BioRID heads with respect to the neck then the BioRID has too great a peak value and too late, the BioRID 2a fits best within the corridor, see Figure 65.
8.3.4 T1 to head angles and trajectories

Figure 66 shows the T1 to head angle for the Hybrid III and the Hybrid III fitted with a TRID neck. The peak in T1 angle is closer to the timing of the volunteer corridor for the Hybrid III, but closer to the peak range of the corridor with the TRID neck version. As with the volunteers, the T1 to head angles are not corrected for general rotation of the occupant, so the angle incorporates the influence of torso rotation.

Figure 67 shows the trajectory of the vertical relative motion of the head with respect to the marker mounted on the T1 accelerometers plotted against the horizontal fore and aft displacement. When considering the relative vertical displacement of the centre of gravity of the head with respect to the T1 marker at the back of the neck, it should be noted that this measurement is a function of complex motions as already discussed in Section 7.2. From Figure 68, showing the angle of T1 to the head plotted against the relative head to T1 marker displacement in the x direction, it can be seen that the Hybrid III with a TRID neck stays within the volunteer corridor. The Hybrid III although possessing too great a T1 marker to head angle, does show a fore and aft direction displacement peak closer to that given by the corridor.
All three of the THOR dummy versions have a peak in $T_1$ to head angle just before the volunteers, see Figure 69. This indicates that the neck of the dummies allows the rearward movement of the head before that seen with the volunteers. However, the THOR has a peak value that is too low and the THOR $\alpha$ has a peak value that is too high at 120% of the upper limit peak. The THOR with a EuroSID neck has a peak value within the corridor limits for $T_1$ to head angle. This may illustrate the variability in the performance of the dummy necks and shows the expected result when considering the THOR $\alpha$’s lack of positive vertical displacement.
The lack of upward vertical motion from the THOR dummies is illustrated further by the horizontal and vertical displacement trajectory of the head with respect to the T1 marker, Figure 70.

The trajectory described by the T1 to head angle plotted against the fore and aft relative displacement of the head with respect to T1, is shown in Figure 71.
Figure 71: THOR T₁ to head angle against fore and aft displacement.

Figure 72 shows the T₁ to head angle for the BioRID and BioRID 2a compared with the volunteer corridor. Figure 73 plots the relative head centre of gravity to the marker mounted on the T₁ accelerometers displacement trajectories of vertical displacement against fore and aft displacement and Figure 74, the T₁ to head angle plotted against T₁ to head fore and aft displacement.

Figure 72: BioRID T₁ to head angle.
From the T₁ to head angle for the BioRID and BioRID 2a, Figure 72, the peak angle is greater than the volunteer corridor upper limit. Also the BioRID peak occurs slightly after the volunteers and the BioRID 2a peaks before. The BioRID 2a also demonstrates an angular velocity both rearward and forward that is greater than the volunteer’s.

The extent of the lack of vertical movement of the head centre of gravity relative to the marker mounted on the T₁ accelerometers is again illustrated by the trajectories of relative head to T₁ displacements, see Figure 73. The BioRID also exhibits too great a rearward relative displacement of the head whereas the BioRID 2a does not move backwards far enough compared with the corridor as defined by the volunteers. This difference in rearward displacement, as identified from the vertical to fore and aft trajectory, is also evident in the plot of T₁ to head angle against fore and aft displacement, see Figure 74. Both dummies have a peak T₁ marker to head angle greater than that seen with the volunteers but because of the further rearward translation of the head on the BioRID, the gradient of the trace follows that from the volunteer corridor. The BioRID gradient is expected to appear more like the BioRID 2a’s, had the head to head restraint distance been smaller. The BioRID dummies can therefore be seen to encourage too great a rotation of the head about T₁ for the amount of rearward motion they produce. In this respect they are the least biofidelic of all the dummies tested here.
8.4 Further observations

The BioRID and THOR α exhibited comparatively large rearward displacements, at the head and T₁ respectively. This may be due to the dummy positioning not being well enough defined. For comparisons with volunteers this variation in initial position will not invalidate the results, as the exact position of the volunteers at the time of impact is unknown. However, it becomes more difficult to relate dummy to dummy performance. Should a low severity rear impact test procedure be proposed as a regulatory test, it will need to have a very well defined dummy positioning specification, noting that the potential variations in dummy positioning will only be increased in vehicle seats as opposed to the simplified test bench used in this study. For this reason, flexibility of the dummy in all areas might not be desirable as this might increase variability in the initial position and set-up. However, such dummy flexibility may be required to find a seating position more similar to a human’s than could be provided by the existing dummies.

9 Injury Risk - NIC

The evaluation of injury risk, as a discrete study, was not undertaken within this project but information from it can be used to make some comments on injury risk criteria proposals being put forward by other research groups. As hoped none of the volunteers tested reported any form of injury, thus one must conclude that any measurement made at this level of impact severity would be below any critical injury threshold for the volunteer group, who represent the young adult end of the population (between 20 and 29 years). Injury criteria are normally based on dummy transducer measurements and what can be measured in a dummy in many instances will be very different to that which can be used with a volunteer. e.g. a dummy can incorporate a force transducer in the neck – a measurement that can not be made with the volunteer. It is possible to compare acceleration based criteria, making the assumption that the external acceleration measurement positions are similar to the external dummy ones. For two-dimensional motion they will be assessing the same motion as would internal transducers in the dummies head.

This study has been able to review the NIC (Neck Injury Criterion) acceleration based criteria [22] that has been proposed as an appropriate Whiplash injury criteria. The NIC was developed to reflect transient pressure changes in the cervical spinal canal.

Low severity design target corridors from the volunteer testing are inappropriate for determining injury risk but it is possible to compare the proposed performance criterion with the volunteer tests by calculating the assessment parameter for the volunteers. For the volunteers, the NIC has been calculated from the head and T₁ accelerations and a fixed width corridor of plus and minus one standard deviation has been produced, Figure 75 and Table 22. However, it is not proposed that this corridor be used to assess a whiplash dummy, it has just been used to extract a plus or minus one standard deviation maximum and minimum for the volunteer data.

The neck injury criterion (NIC) has been calculated for the volunteers and the three dummies tested here, using the equation:

\[ NIC = a_{rel} \times 0.2 + (v_{rel})^2 \]

where \( a_{rel} \) is the acceleration of the neck relative to the head \((a_{T1} - a_{head})\) and \( v_{rel} \) is the velocity of the neck relative to the head \((v_{T1} - v_{head})\).
The neck is bent into a \( \uparrow \) shape in the first stage of the impact and it is suggested by Boström et al., that this is the stage when spine pressure transients are at their maximum [23]. Therefore the NIC\(_{\text{max}}\) values reported in Table 23 were the maximum positive NIC value occurring within the first 150 ms of the test, corresponding to the initial rearward motion of the head with respect to T1. It was possible to test the dummies at impact severities greater than with the volunteers. The opportunity was taken to subject the THOR \( \alpha \) and BioRID 2a dummies to a 4 g impact and a stepped 2 then 4 g pulse. Although the results are presented here, limited discussion is offered at this stage as only two dummies have been tested at these conditions.

In the literature related to NIC injury it has been hypothesised that injury would not occur at values of NIC < 15 m\(^2\)s\(^{-2}\). According to this criterion none of the impacts were severe enough to cause neck injury.

The Hybrid III gives a NIC\(_{\text{max}}\) value below the other two dummies but above the volunteer range. It is therefore the closest of the dummies to the volunteer results. Under the 2 and 4 g impact conditions the BioRID 2a records a similar NIC\(_{\text{max}}\) to the THOR \( \alpha \), yet the THOR \( \alpha \) gives a higher value than the BioRID 2a in the stepped deceleration pulse test.

Figure 76, Figure 77 and Figure 78 show the Hybrid III, THOR and BioRID dummy’s NIC time histories, respectively, from the 2 g deceleration impacts, compared with the corridor generated from the volunteer tests. The upper limit of the volunteer corridor gives a NIC\(_{\text{max}}\) value of 5.3 m\(^2\)s\(^{-2}\). Only the THOR fitted with the EuroSID neck, Hybrid III fitted with a TRID neck and the BioRID peak under this value. Of these three dummy variants, it is the THOR fitted with an EuroSID neck that remains within the corridor for the longest period. All of the dummies give a similar peak value for NIC\(_{\text{max}}\), but then drop out of the corridor approximation of the volunteers’ responses, due to the larger negative peak.
Figure 76: Hybrid III NIC.

Figure 77: THOR NIC.

Figure 78: BioRID NIC.
10 Discussion
Within this project, there have been many phases of work all contributing towards the specification and evaluation of a crash test dummy, suitable for use in evaluating the risk of spinal injury in a rear impact. The individual phases are discussed in the following sections:

10.1 Literature study
A number of important issues, have been identified relating to rear impact injury risk assessment. Most research into rear impact purely focuses on neck injury and the injury commonly called Whiplash. Little effort appears to be directed towards addressing injury risk to the whole spine.

Information against which to develop and assess a rear impact dummy has been found to be rather limited. Data that have been developed focus solely on the performance of the head and neck with little comment on the behaviour of the rest of the spine. It has therefore been necessary to develop a range of good quality dummy design and assessment targets that could be replicated at any time in the future.

10.2 Volunteer tests
Dummy design and performance targets have been developed based on a series of volunteer tests. A suite of transducer and kinematic targets has been developed as well as an assessment of the pressure distribution interface between the volunteer and the seat back. These have then been applied to evaluate existing dummies that might be used in low severity rear impacts.

10.3 Dummy evaluation
Within this project three markedly different dummies were evaluated, some with different necks. The Hybrid III has a rigid thoracic spine section. The THOR has a similar flexible lumbar section to that in the Hybrid III but in addition, introduces a flexible thoracic section. The BioRID has a fully articulated spine, from head to pelvis, as can be seen in Figure 3. This difference in the dummy spines may explain, or contribute towards explaining almost all of the features found in the results.

From consideration of the dummy kinematics related to the volunteer behaviour, the Hybrid III suffers from having too small T1 and head movements. Initially it appears to be quite biofidelic and the performance of the neck as evaluated from relative fore and aft head to neck motion is almost within the corridor (Figure 63). The seat back interaction may explain the poor biofidelity of the torso. In the Tekscan pressure mat image results, almost exclusively the pressure is exerted along the line of the spine (Figure 30). This is probably due to a combination of factors: The lack of flexibility in the spine with the long rigid thoracic section. The shape and rigidity of the Hybrid III back and torso flesh and the mass distribution within the body, with the Hybrid III having a heavy spine and relatively light flesh as opposed to the volunteers with a light spine and heavy flesh and fluid filled organs.

To improve upon the performance of the Hybrid III head motion, the TRID neck was developed. This gives a better head displacement and neck deflection at the expense of the timing of the response. It has to generate this head performance from the same small T1 movements.

The THOR neck also manages to generate biofidelic fore and aft head displacements from too small T1 movements in the x and z-directions. A side effect of this is that the head translates down instead of up as the neck extends backwards without any ramping up of the thorax at T1. As with the Hybrid III, the poor biofidelity as measured at the base of the neck, is likely to be a direct result of poor seat back forces and pressure profile, features of a spine and torso which do not behave in a biofidelic manner. In these parameters, the THOR shows only a small improvement with the THOR α, and performs in a similar manner to the Hybrid III. The peak pressure can still be seen to run along the spine with very little distributed elsewhere over the dummy’s back.

With the BioRID, both the seat back pressure profile and the performance as evaluated at T1 showed a significant improvement over the Hybrid III and THOR dummy versions. The design features that
may have brought about this overall improvement are the more flexible spine and the greater transference of effective mass away from the heavy spine during the impact.

The BioRID 2a head accelerations have shown that it is possible to generate biofidelic head responses from less biofidelic T₁ responses. The only potential problem with this T₁ to head linkage is that should the seat back to dummy interaction be altered to be closer to that of the humans, then the neck will once again need revising. That is, if the T₁ acceleration in the fore and aft direction is delayed and the vertical component increased then the reverse will need to be designed into the neck to achieve the correct head acceleration.

10.4 Injury risk
The NIC was investigated as a parameter for dummy comparison. Based on the poor biofidelity performance of all of the dummies tested in producing spinal kinematics. The validity of using this criterion to assess injury risk must be questioned. Should only neck criteria be adopted for evaluating rear impact severity, then rear impact safety features may solely target reduction of these parameters. It is conceivable that without a dummy that demonstrates biofidelic spinal behaviour, neck injury prevention may be at the expense of thoracic and lumbar region injuries. The absence of injuries to the volunteers at NIC values below the proposed performance requirement suggests that the limit is not too low, but it cannot be used to set a new value.

10.5 Summary
The test devices available for evaluation within this project all have reasonably biofidelic necks. The BioRID 2a neck, as a refinement of the BioRID, shows the most biofidelic performance overall. However, the issue of the biofidelity of rear impact dummies should not focus solely around the motion of the head and the performance of the neck, as the transmission of forces between the occupant and the seat back, appears to present more problems. Although it appears to be possible to compensate for a non-biofidelic rigid thorax by adjusting the neck stiffness, this method will only produce reliable results in the impact situation tested. Ideally a test device should produce sensible results over a range of impact conditions. It is strongly recommended that future whiplash dummies should have a full biofidelic spine.

The rigid or semi-rigid spines of the Hybrid III and THOR fail to encapsulate the rolling behaviour of the human spine as it straightens and adapts to the profile of the seat back. None of the dummies go so far as to generate the same lift, as observed at the head, as from the complete lumbar, thoracic and cervical spine extension (straightening) in the human volunteers. This is accompanied in the humans by the rolling of the shoulders, whereas the dummies have a much more rigid collar and torso in general. Therefore before the neck design is finalised, development should be directed towards improving on the performance produced by the BioRID 2a torso, since the BioRID, with its flexible spine, is the only dummy where the whole spine biofidelity has the potential to be made to match a human in a rear impact situation. The BioRID 2a, as evaluated by this project allows some straightening, effective extension, of the spine and the best pressure interface with the seat back, which are the features evidently lacking from all current dummies.

Assessing dummy performance and injury risk is very difficult. It is conceivable that injury risk assessments made by some of the proposed dummies could be misleading and predict savings that would not occur in real world accidents due to lack of whole spine biofidelity and whole spine injury risk assessment.

As the human behaviour in our rear impact tests is so dependent on the movement of the whole spine, the parameters that are the most important for evaluating spinal injuries will need to reflect this and not just concentrate on the neck. It is difficult to propose alternative parameters on which to base injury risk, as the dummies and volunteers do not have comparable instrumentation.

The muscle activity observed in the EMG results occurred very early in the impact. It is assumed that this tensing was due to the stimulation of the first shocks of the impact coming through the seat. One
might also hypothesise that controlling pelvis acceleration might be a way of attenuating the speed or magnitude of upper body muscle activity. Following on from this, it is possible that seat cushions that transmit the force more gently would reduce occupant tensing causing them to react with the seat as intended.

11 Conclusions

1) A series of volunteer, rear impact, sled tests have been carried out in order to obtain suitable data for specifying the performance of whiplash dummies.

2) EMG readings from the volunteer tests showed almost instantaneous tensing of major muscle groups in the back. Comparison of the EMG results with the kinematics of volunteers who exhibited some tensing with those who exhibited little muscle tensing, showed that bracing significantly affected the head motion. This strongly suggests that PMHS data is not suitable for developing whiplash performance corridors.

3) A series of performance corridors have been derived from the tests with the volunteers. These corridors can be used to specify the low severity response of whiplash dummies.

4) Although it appears to be possible to compensate for a non-biofidelic rigid thorax, found in many dummies, by adjusting the neck stiffness, this method will only produce reliable results in the impact situation for which the neck was adjusted. A test device should produce sensible results over a range of impact conditions. It is strongly recommended that regulatory whiplash dummies should have a full biofidelic spine.

5) A series of rear impact sled tests have been carried out with three different anthropomorphic test dummies. The dummy results have been compared with the volunteer performance corridors and this has shown:
   a) No version of the Hybrid III dummy tested gives a satisfactory performance despite it meeting some of the proposed performance criteria. This is fundamentally due to the dummy having a rigid thoracic spine
   b) Although the THOR has flexible joints in both its lumbar and thoracic spine, it is not considered suitable for use as a whiplash dummy because it still does not demonstrate the required biofidelic flexibility of the spine and torso.
   c) The BioRID appears to show potential for use as a whiplash dummy. However, the torso and spine together require an increased flexibility and further tuning of the individual spine sections’ stiffness is needed.

6) A literature study has been carried out and has shown that only limited data are available for specifying required whiplash dummy performance. From this review a list of points were extracted that had implications for designing whiplash dummies.

7) A Hybrid III dummy/human mathematical (FE) model has been developed. The model is useful in predicting kinematics and load concentrations. However, further development is needed before it can be used to predict injury risk.
12 Future Research

The BioRID demonstrated rearward, T₁ marker motion most like the volunteers. However, inconsistencies within the set-up were noticeable. Before a rear impact test procedure is established, better definition of dummy positioning techniques should be described.

The potential value of mathematical modelling has been established. Extension of the work described in this report should include incorporating better biomechanical data and enhanced validation.

Within this project many dummy design targets have been developed. It is suggested that any new dummy should be evaluated against these targets.

The current research only examined two-dimensional dummy behaviour. In real world impacts, occupants will be loaded from a number of different directions. Future research should examine oblique behaviour to ensure that the injury predicting devices (the dummies) work in impacts other than pure rear.

13 Glossary

**Anterior**
Nearer to or at the front of the body.

**Anthropometry**
The study of human body measurement for use in anthropological classification and comparison.

**Anthropomorphic**
Suggesting human characteristics for animals or inanimate things.

**C₁ to C₇**
Cervical vertebrae (neck).

**CR line**
The CR line is the intersection line between the top plane of the seat base and the front plane of the seat back.

**Electromyography**
Evaluation of the electrical activity of resting and contracting muscle.

**Erector spinae**
The erector spinae muscle runs longitudinally up the back. It is the largest muscle mass of the back and consists of three groups of muscles.

**Extension**
The act of straightening.

**Flexion**
The act of bending a joint or limb in the body by the action of flexors.

**Flexure**
The act of bending or flexing.

**Frankfort plane**
A standard reference plane approximately passing from the middle of the ear to the bottom of the eye socket.

**In vivo**
In the living body.

**L₁ to L₅**
Lumbar vertebrae (loin).

**Medial**
Nearer the midline of the body or structure.

**Occipital bone**
The occipital bone forms the posterior part and most of the base of the cranium.

**Parietal bones**
The two parietal bones form the greater portion of the sides and roof of the cranial cavity.

**Posterior**
Nearer to or at the back of the body.

**Sternocleidomastoideus**
The sternocleidomastoid muscle, is one of two muscles located on the front of the neck which serve to turn the head from side to side.

**T₁ to T₁₂**
Thoracic vertebrae (chest).

**Thoracolumbar**
Of or relating to the thoracic and lumbar parts of the spinal column.

**Trapezius**
Either of two large flat triangular muscles running from the base of the occiput to the middle of the back that support and make it possible to raise the head and shoulders.
References


11. UN-ECE. Regulation 44: Child restraint system.


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The authors would like to acknowledge TNO for the use of the RID neck and thank Mr. Mat Phillippens (TNO) for his practical support in the volunteer study. The authors are grateful to Mr. Mathew Avery and the Motor Insurance Repair Research Centre (Thatcham) for the use of the BioRID dummies and guidance in using them. We would also like to acknowledge Dr. Andrew Whitfield (Healthnet) for his great help in obtaining ethical approval for the volunteer study and in his practical help during the tests. Finally we would like to thank all of the TRL Limited staff who made valuable contributions towards the project.

Appendices

Dual sled impact testing

Appendix 1: Sled test procedure
For biofidelity assessment it is important to have a simple easily reproducible test procedure. Many previous volunteer studies have either been based on vehicle seats or seats with vertical rigid backs at 90° to the base, which do not replicate vehicle posture. The former test condition is not easily reproducible and the latter not very vehicle like. In this study a surrogate vehicle seat was based on the bench seat used in ECE Regulation 44 to assess child restraints.

Following the preliminary Hybrid III tests and before the volunteer test programme began, it was felt to be inappropriate to accelerate and then decelerate the volunteer backwards for safety reasons as well as the fact that it could stimulate unwanted muscle reactions as well as variable initial positions. To overcome these concerns a twin sled system was used. The second sled was used as bullet mass to impact the R44 sled.

Appendix 2: Sled seat design
The seat used in this study is geometrically similar to the R44 seat, except that the height of the R44 seatback was increased to support the shoulders. The back cushion was replaced by a stiff polyurethane foam and the base cushion has been replaced by high density, closed cell, polyethylene foam (EV30) manufactured by Zotefoam and had a height of 590mm above the CR line of the R44 bench [Section 13]. The front of the seat squab was 580mm forward of the CR line, both measurements being made along the plane of the surface. Connected to the back cushion were three steel plates, which divide the cushion into three equal areas. Each of these steel plates faced onto 4 load cells, which monitored the force on each plate during impact. In addition a head restraint was attached to the seat to limit gross motion of the occupant's head. The head restraint contained a portion of the high density EV30 Zotefoam polyethylene with some softer packaging foam in front.

Figure 79 shows the impactor acceleration results from four drop tests onto a 70 mm thick section of the high density, closed cell, Zotefoam EV30 polyethylene. The impactor mass was 6.8 kg and it was 248 mm in diameter. The impact speed was 2.2 ms⁻¹. Prior to these tests a small section of the same material was tested at a variety of different impact velocities. The impactor acceleration results from tests at 4, 8 and 16 kmh⁻¹ are presented in Figure 80, the stress strain outputs can be found in Figure 81. These earlier tests were carried out using 70 mm square sections of foam, the sample thickness varied between 27 and 34 mm. These smaller sub systems tests involved a flat impactor face of equal dimensions to the test sample of mass, 1.76 kg, reaching the specified velocities for impact.
Figure 79: Repeatable impact performance Zotefoam EV30 polyethylene

Figure 80: Acceleration impact performance of Zotefoam EV30 polyethylene at varying velocities

Figure 81: Stress/strain impact performance of Zotefoam EV30 polyethylene at varying velocities
Appendix 3: Impact pulse

An aluminium honeycomb block was placed between the two sleds to limit the amplitude of the decelerating pulse experienced by the volunteer and dummy to 2 g, over a period of approximately 100 to 120 ms, in effect replicating a rear-end shunt type impact. The change in velocity, experienced by the subjects, was 2.0 m/s\(^2\) (rounded to 7 km/h\(^1\), in some descriptions). Figure 82 shows the acceleration pulses from the volunteer tests. For the tests of this nature the Aluminium blocks were of dimensions 200 x 200 x 250 mm and of a constant crush strength of 0.342 MPa ± 10 %.

![Figure 82: Acceleration pulses as measured on the target trolley in volunteer tests](image)

Appendix 4: Instrumentation

External instrumentation

Pairs of small Endevco 7264B accelerometers fixed to biaxial mounting blocks recorded linear horizontal and vertical head accelerations. These mounting blocks were attached to thin aluminium plates held in position on either side of the head by the head straps. The accelerometers were located as close as possible to the centre of gravity of each subject’s head. The sensing axes of the accelerometers were parallel and perpendicular to the Frankfort plane and these axes defined the local x and z axes respectively for each volunteer.

A small biaxial arrangement of accelerometers was mounted at the T\(_1\) location. For the volunteers, a small sheet of malleable thermoplastic was mounted on top of the T\(_1\) spinous process and held in place on the subject’s skin using elastic medical tape. Double-sided adhesive tape was then used to fix the accelerometers onto the thermoplastic. As the dummies were not likely to feel any discomfort, the accelerometer arrangement was attached directly using double-sided adhesive tape. The sensing directions of the accelerometers, so mounted, were normal and tangential to the surface of the volunteer’s skin or dummy’s neck at T\(_1\). The initial angle of orientation of the accelerometers was measured relative to the vertical before each test with volunteers. Care was taken to ensure that the T\(_1\) accelerometers would not foul against the top of the seat or bottom of the head restraint during any of the impact tests. The locations of the head and T\(_1\) instrumentation can be seen in Figure 83 and Figure 84 showing these instruments set-up on both a volunteer and the THOR α respectively.
The seventh and final accelerometer used with the volunteers was mounted at the base of the spine above the bony protrusion of the sacrum. Once again elastic medical tape was employed to hold the accelerometer in place. This accelerometer recorded acceleration in the fore and aft x-direction.

For the dummy tests the six head and T₁ external accelerometers were used and these provided the data used to compare dummies with volunteers. Further to these channels, the internal dummy instrumentation was also recorded.

Internal instrumentation

Hybrid III

Standard Hybrid III instrumentation was used:

- Head acceleration
- Chest acceleration
- Pelvis acceleration
- Upper neck force
- Upper neck moment

All tri-axial, as well as chest deflection.

THOR

The 8 accelerometers used for the THOR instrumentation are as follows:

- Head fore aft and vertical.
- Chest fore aft and vertical.
- Pelvis fore aft and vertical.
- Lower neck (T₁) fore aft and vertical.

BioRID

The BioRID had accelerometers measuring the following:
• Head, fore aft, lateral and vertical.
• 4th Cervical, fore aft and vertical.
• 1st Thorax, fore aft and vertical.
• 8th Thorax, fore aft and vertical.
• 1st Lumbar, fore aft and vertical.

Also recorded were the Neck load forces, fore aft and vertical as well as the neck load moment about the y-axis.

All recorded signals from both external and internal accelerometers were processed in accordance with ISO 6487.

Instrumentation results are compared in Section 8.2.

Appendix 5: Test set up

The computer modelling work used a spinal posture similar to the UMTRI (University of Michigan Transportation Research Institute) position. It made sense for the volunteers to adopt the dummy model position rather than vice versa. Anatomical points such as hip, head centre of gravity and knee point, etc. could not be matched due to variation in human beings, therefore only external reference distances were matched, such as head to head restraint, shoulders to seat back, pelvis to seat back and angle of lower leg.

The restraining seat belt(s) was set tight by hand. Leaning the occupant forward could ensure contact between the lumbar and the seat back, before relaxing the torso back. The head was set to be approximately 75 mm from the head restraint. Arms were positioned with the upper arms adjacent to the torso and with the lower arms resting such that the hands were lying palms down on the top of the thighs. The knees of the Hybrid III and THOR dummies were specified to be 269 mm apart, where the distance is to the outboard knee clevis flange. The legs were in the vertical longitudinal plane as far as possible. Feet were placed comfortably out in front with the ankle angle at 90 degrees.

Figure 85: Volunteer test set-up
Figure 86: THOR α test set-up, oblique view

Figure 87: THOR α test set-up, lateral view
The following annexes are comprised of copies of new and previously published documents produced during the course of the project for DfT. Only one has been available to the public.

**Annex 1: Literature review**
Biomechanics, injuries and modelling of the spine in motor vehicle accidents

by P R Dixon, R Minton, M LeClaire and G G Edwards
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EXECUTIVE SUMMARY

This literature review has been funded by the Vehicle Standards and Engineering Division of the Department of the Environment, Transport and the Regions. It has been conducted to obtain an understanding of the current knowledge of spinal injuries and the mechanisms which cause them with a view to developing an improved biofidelic spine for the Hybrid III dummy. With this aim in mind a survey of the most recent dummy spine developments has also been undertaken. During the course of the study, the ideas of biomechanical experts at Volvo and Chalmers University, Gothenburg, Sweden and at TNO, Holland have been sought. Furthermore, the opinions of medical experts have been gained via a series of meetings with orthopaedic surgeons from Hope Hospital, Manchester and the Queens Medical Centre, Nottingham. All of these various medical and technical viewpoints have contributed to the content of this report.

The review starts with a brief description of the basic anatomy of the human spine. A familiar knowledge of this is essential to the understanding of the biomechanics and injury mechanisms of the spine. Chapter 3 deals with the subject of spinal biomechanics. A quantitative knowledge of this is required for the future development of both physical and mathematical models of the spine. The subject inevitably involves the use of some medical vocabulary and therefore a glossary of medical terms has been included in Appendix A to assist the reader.

In chapter 4 the various mechanisms of cervical spine injury are discussed. It is apparent from this section of the review that the so called ‘whiplash’ injury and whiplash associated disorders are still not well understood and there is much debate about their actual cause. Further research is needed in this area before good understanding of the injury mechanisms can be obtained and injury criteria developed which will have the capability of predicting ‘whiplash’ injury.

More severe spinal injury which may involve damage to the cord, ligaments and vertebral bodies is better understood, particularly in the thoracolumbar spine (chapter 5). However, it would appear that most research on spinal injury has focussed on the neck and that, in comparison, the thoracolumbar spine has been rather neglected. Hence, there is a shortage of tolerance and physical properties data for the latter region of the spine. Perhaps the future work programme should include some study of the biomechanical characteristics of spinal tissue which could then be used in finite element modelling of the spine.

Within the time scale of this review, it has only been possible to conduct a brief survey of spinal injuries in traffic accidents. The results of this are presented in chapter 6. It may be worthwhile extending this accident analysis to examine the relative importance of thoracolumbar spine injuries compared with cervical spine injuries. Other factors, which affect spinal injury, such as impact direction, severity of impact, type of vehicle could also be explored in more detail.

Chapter 7 is a summary of mathematical modelling techniques and illustrates how such models can be cost effective and flexible research tools for the study of the response of the spine to impact. These models are especially powerful when supported and validated by a complementary experimental test programme. It is expected that finite element modelling will have a prominent role to play in any future design and development work at TRL.

Chapter 8 discusses the limitations of the existing Hybrid III spine. These well known deficiencies are widely reported in the literature and gave rise to the initial impetus for the current project. Essentially, the Hybrid III spine is too stiff and not very humanlike in its response to impact. This may result in misleading measurements being made from instrumentation in the dummy’s neck and head. Several investigators around the world are attempting to improve the biofidelity of the Hybrid III spine and the most recent developments are reviewed in chapter 9. However, it is clear that even these latest designs have serious limitations.
This literature review has demonstrated the complexity of spinal injuries in humans and spinal developments in dummies. The exact causes of whiplash associated disorders are still poorly understood and current test hardware is limited in its biofidelity and performance. There still appears to be a need for a biofidelic omnidirectional dummy spine and neck for the assessment of vehicle seats and the prediction of spinal injury in impact tests. Biomechanics of the spine is a very vast subject but this report has highlighted those areas in which TRL can make a valuable and worthwhile contribution to current research in this field. The report concludes with proposals for future work.
BIOMECHANICS, INJURIES AND MODELLING OF THE SPINE IN MOTOR VEHICLE ACCIDENTS

ABSTRACT

This report is a comprehensive review of the published literature on the biomechanics and injury mechanisms of the human spine in relation to motor vehicle accidents. The study was undertaken with a view to the future development of an improved biofidelic spine for the Hybrid III crash test dummy. The principle mechanisms of injury to the spine are summarised. Of the various types of injury possible, ‘whiplash’ is the one which is least well understood. A number of plausible theories of the cause of ‘whiplash’ and whiplash associated disorders are described. The results from a preliminary survey on spinal injuries in vehicle accidents are presented. There is also a chapter on mathematical modelling of the human spine. The limitations of the existing Hybrid III spine are discussed and the latest developments in dummy spine design are reviewed. Finally, the review identifies gaps in current knowledge needed for the design and development of a biofidelic dummy spine and makes recommendations for future work.

1 INTRODUCTION

The last ten years have witnessed a remarkable increase in the number of safety features installed in motor vehicles. Nowadays, these features are used by manufacturers in advertisements to enhance the sales of their cars and, consequently, the general public is much more aware of safety issues. With the improvements in vehicle safety, there has been a steady decline in the number of fatalities recorded annually in traffic accidents. Thus, in more severe accidents there has been a general shift from fatal injuries to survivable serious injuries. Spinal injuries can fall within this group.

Although serious spinal injuries (AIS ≥ 2) are not as common as serious injuries to other parts of the body sustained in road accidents, they still account for nearly half of the casualties admitted to spinal injury trauma units. They frequently result in a high level of disablement for the victims and thus high financial costs for society, not to mention the cost in terms of human suffering.

Even minor injuries to the spine (AIS = 1), although they may not be life-threatening, can have long lasting and debilitating effects for the individuals concerned. Of these, neck strain injuries or ‘whiplash’ injuries, as they are commonly called, are the most significant. ‘Whiplash’ is currently a topical subject of research world-wide and there is much debate as to what the actual causes of this type of injury are. Indeed, it is shown in chapter 4 of this document that the term ‘whiplash’ is a very poor description of the injury itself.

The study of the biomechanical behaviour of the human spine in motor vehicle accidents is clearly very important and more research is required before the causes and mechanisms of spinal injuries are fully understood. Such an understanding is needed for the development of biofidelic spines in anthropomorphic test devices which will allow the accurate prediction of spinal injuries in vehicle impact tests. Currently, there are no crash test dummies which adequately fulfill this capability. The complexity of the anatomy and biomechanics of the human spine has meant that the progress in dummy spine development has been slow. The existing Hybrid III dummy was designed principally for frontal impacts and shows poor biofidelity in rear impacts in which ‘whiplash’ injuries are common. Furthermore, the standard Hybrid III instrumentation is of limited use in assessing the likelihood of injury to the spine.
This literature review has been funded by the Department of the Environment, Transport and the Regions (DETR) as the first stage of a research project to improve the biofidelity of the Hybrid III spine. The subsequent stages of the project have not been defined in detail and are dependent on the findings of this review. Thus, the following chapters represent a comprehensive investigation of the published literature on the vast subject of the biomechanics and injuries of the human spine in motor vehicle accidents. This report has been produced with a view to the future development of a biofidelic dummy spine and so a survey of the latest antropomorphic test hardware has been included.

In producing this work, the opinions of orthopaedic surgeons at Hope Hospital, Manchester and the Queens Medical Centre, Nottingham have been sought. Also, the views of biomechanical engineers at TNO, Holland and at Chalmers University and Volvo, Gothenburg, Sweden have been obtained. The ideas of all of these experts have been valuable and are discussed within the main text. It is hoped that this review will be the first step in the future development of an improved dummy spine. This in turn should lead to safer car seats and greater protection for vehicle occupants in traffic accidents.
2 ANATOMY

2.1 GENERAL DESCRIPTION

The human spine is a strong flexible column which extends from neck to coccyx and consists of a series of bones known as vertebrae. In an average adult male it measures about 71cm in length and about 61cm in the average adult female. The spine or vertebral column, as it is often called, encloses and protects the spinal cord, supports the head, and provides a point of attachment for the ribs and the muscles of the back. It transmits body weight in walking and standing, is capable of movement in the anterior (forward), posterior (rearward) and lateral directions and can also rotate. Between the vertebrae are openings called intervertebral foramina through which pass nerves connecting the spinal cord to various parts of the body.

The adult spine typically contains 26 vertebrae (see figure 2.1). These are grouped as follows; 7 cervical vertebrae (C1 to C7) in the neck region, 12 thoracic vertebrae (T1 to T12) at the rear of the thoracic cavity, 5 lumbar vertebrae (L1 to L5) supporting the lower back, 5 sacral vertebrae which are fused in the adult to form one bone called the sacrum and usually 4 coccygeal vertebrae fused together to form the coccyx. Thus, the human spine has a total of 33 vertebral bodies prior to the fusion of the sacral and coccygeal vertebrae which occurs between birth and adulthood. From an injury and dummy development point of view it is the cervical, thoracic and lumbar regions of the spine which are most important because damage to the spinal cord in these areas may result in paralysis. Occasionally, the thoracic and lumbar regions are treated as a single unit referred to as the thoracolumbar spine.

A concise overview of the anatomy of the spine is presented below. A more detailed description can be found in the excellent texts of Kapandji (1998) and Tortora (1989).

2.2 INTERVERTEBRAL DISCS

Between adjacent vertebrae are fibrocartilaginous intervertebral discs. Each disc is composed of two parts; a soft centre (nucleus pulposus) surrounded by a tough fibrous flexible ring (annulus fibrosus). Figure 2.2 shows how a disc is situated between two vertebral bodies. The nucleus pulposus is a transparent jelly containing 88% water and maintains an internal pressure of approximately 10N/cm². The two main functions of intervertebral discs are to allow movement between pairs of vertebrae and to act as buffers against the shocks caused by running, jumping and other stresses which may be applied to the spine. Each disc behaves as a fluid and thus under a compressive load it will shorten in height and expand laterally. After arising in the morning, and during the day, a person decreases in height because of the compression of the discs under body weight. In a healthy individual this may amount to 1mm per disc giving a total of 23mm of shortening for the whole vertebral column. The spine lengthens again, of course, during sleep.
Figure 2.1: The vertebral column. (a) Anterior view. (b) Right lateral view (From Principles of Human Anatomy 5th edition by G Tortora. Copyright (C) 1988 by Harper & Row Publishers. Reprinted by permission of Addison-Wesley Educational Publishers)

Figure 2.2: Intervertebral disc in its normal position (left) and under compression (right). The relative size of the disc has been enlarged for emphasis. A “window” has been cut in the annulus fibrosis so that the nucleus pulposis can be seen. (From Principles of Human Anatomy 5th edition by G Tortora. Copyright (C) 1988 by Harper & Row Publishers. Reprinted by permission of Addison-Wesley Educational Publishers)
2.3 CURVATURE

In figure 2.1 it can be seen that the spine is not actually a straight column but is rather a sort of spiral spring having the shape of a letter S. A newborn child has a backbone curved in a single arc which is anteriorly concave. During the transition to adulthood, the spine develops four distinct curves (two convex and two concave) as the supporting functions of the vertebral column are acquired. The S-curvature enables the spine to absorb the shocks of walking on hard surfaces; a straight column would conduct the jarring shocks directly from the pelvis to the head. Curvature of the vertebral column in the direction of primary curvature (concave anteriorly) is known as kyphosis, while that in the direction of secondary curvature (concave posteriorly) is termed lordosis.

The weight of the whole body is more evenly distributed along the length of an S-shaped spine than it would be if the spine were a straight column which would concentrate the greatest load at its base. Furthermore, the S-curvature protects the vertebral column from breakage. The doubly bent spring arrangement is far less vulnerable to fracture than would be a straight column.

2.4 VERTEBRAE

Figure 3 illustrates the structure of a typical vertebra. Although there are variations in size, shape and detail in the vertebrae in different areas of the spine all of them, apart from C1 and C2, share the same basic structure. There are three principal components to a vertebra:

1. The vertebral body is the thick, disc-shaped anterior (front) portion which has a load bearing function. Its superior (upper) and inferior (lower) surfaces are roughened for their connection with the intervertebral discs.

2. The vertebral (neural) arch extends posteriorly (rearwards) from the vertebral body forming an almost ‘horse-shoe’ shape. It is formed by two short, thick sections of bone, known as the pedicles, which project posteriorly from the body to unite with the laminae. The laminae are the flat parts that join to form the posterior portion of the vertebral arch. The space between the vertebral arch and body contains the spinal cord and is known as the vertebral foramen. Collectively, the vertebral foramina of all the vertebrae form the vertebral (spinal) canal. The pedicles are notched superiorly and inferiorly in such a way that, when they are arranged in the column, there is an opening between vertebrae on each side of the column. This opening, the intervertebral foramen, permits the passage of a single spinal nerve.

3. Seven bony projections or processes emerge from the vertebral arch. At the junction of a lamina and a pedicle, a transverse process extends laterally on each side. A single spinous process projects posteriorly (rearwards) and inferiorly (downwards) from the junction of the laminae. These three processes act as points of muscular attachment. The remaining four processes form joints with neighbouring vertebrae. The two superior articular processes of a vertebrae articulate with the vertebrae immediately superior to them. Similarly the two inferior articular processes of a vertebra articulate with the vertebra inferior to them. The articulating surfaces of the articular processes are usually referred to as facets. The processes and their relative positions can be more clearly visualised in figure 2.3.
Figure 2.3: Typical vertebrae. (a) Diagram of superior view. (b) Diagram of right lateral view. (From Principles of Human Anatomy 5th edition by G Tortora. Copyright ©1988 by Harper & Row Publishers. Reprinted by permission of Addison-Wesley Educational Publishers)
2.5 REGIONAL VARIATIONS IN VERTEBRAE

2.5.1 The Cervical Spine

The seven cervical vertebrae are numbered from the top; C1, also called the Atlas, is the topmost vertebra, followed by C2, the Axis, then the lower cervical vertebrae, C3-C7. The atlas is so called because it supports the head. Essentially, it is a ring of bone having anterior and posterior arches and large lateral masses. It lacks a body and a spinous process whereas the axis has both. The most visually obvious difference between the lower cervical vertebrae and those of the thorax is that the cervical vertebrae are smaller and their transverse processes are much less pronounced.

This feature facilitates lateral flexion of the neck, and allows roll motion for the head. The top two vertebrae differ considerably from the rest of the spinal column. The Axis (C2) features a vertical “peg”, the odontoid process, sometimes simply called the “dens” (Latin, tooth), projecting upwards from its anterior portion. The Atlas (C1) fits over this peg, which thus acts as a pivot for rotation of the Atlas in a horizontal plane. This Atlanto-axial joint is responsible for much of the yaw motion of the head (approximately 40-60%), with rotation of the lower cervical vertebrae contributing the remainder. The upper facet surfaces of the Atlas form a joint with two bony protuberances at the base of the skull - the Occipital Condyles. This Atlanto-occipital joint allows rotation about a lateral axis only, and, in combination with the fore-and-aft bending of the rest of the cervical column, gives the head its pitch motion. In physiological parlance, forward pitch of the head (looking down) is called flexion, while backward pitch (looking up) is called extension.

2.5.2 The Thoracic and Lumbar Spine

Thoracic vertebrae are considerably larger and stronger than those of the cervical region. Also, the spinous process on each vertebra is long, laterally flattened, and directed inferiorly. Furthermore, thoracic vertebrae consist of longer and heavier transverse processes which have facets for articulating with the ribs.

The lumbar vertebrae (L1-L5) are the biggest and strongest in the column. Their various projections are short and thick and are well adapted for attachment to the large back muscles and ligaments.

2.6 LIGAMENTS

There are a number of ligaments attached to the spine which play an important part in its overall stability and movement. The anterior longitudinal ligament runs continuously down the front of the vertebral column and is attached to the discs and not the vertebral bodies. In a similar manner, the posterior longitudinal ligament runs continuously down the inside of the spine between the spinal cord and the back of the vertebral bodies. Again, this ligament is only attached to the discs and prevents the discs from coming out into the spinal canal. The ligamentum flavum is a separate sheet of elastic connective tissue which passes between the neural arches and connects one lamina to the next. The supra-spinous ligament joins all the spinous processes together.

2.7 MUSCLES

As well as its role in support and protection, the vertebral column is important in the anchoring of muscles. Many of the muscles attached to it are so arranged, in fact, as to move either the column itself or various segments of it. There is a complex network of superficial and deep lying muscles. Probably the most important muscle is the large erector spinae which keeps the spine in tension and, as the name implies, holds the spine erect. This muscle can also initiate movement of the whole column. It begins on the sacrum and passes upward forming a mass of muscle on either side of the
spinous processes of the lumbar vertebrae. It then divides into three columns ascending over the back of the chest. Although slips (narrow strips) of the muscle are inserted into the vertebrae and ribs, it doesn’t end here, as fresh slips arise from these same bones and continue on up into the neck until one of the divisions, called the longissimus capitas, finally reaches the skull.

Another important large muscle is the ilio psoas which connects the transverse processes of the lumbar spine with the pelvis and plays a significant role in walking, running etc. Smaller muscles run between the transverse processes and between the vertebral spinous processes of adjacent vertebrae and from transverse processes to spinous processes thereby giving great mobility to the segmented bony column.

In the cervical spine, relative motion of the head and neck is accomplished through muscle pairs which are attached to the skull, individual vertebrae, and the torso via tendons. These pairs, which are symmetric on the left and right sides of the body, respond in various group actions to produce the desired movement of the neck and head. Muscle pairs which produce voluntary flexion are the ones which resist extension and vice versa.

The muscles lying behind the head/neck are more massive than those lying in front and they are also located further from the head-neck pivot (the occipital condyles). This means that larger moments can be developed for resisting flexion than extension. Furthermore, a lower resultant muscle force is required to produce the same magnitude of resisting bending moment in flexion than would be required in extension.

2.8 REFERENCES


3 BIOMECHANICS

3.1 NATURAL RANGE OF MOVEMENT

As a whole, the vertebral column from sacrum to skull is equivalent to a joint with three degrees of freedom; it allows flexion and extension in the sagittal plane, lateral flexion right and left and axial rotation. The range of these basic movements at each vertebra is very small but because the spine has many vertebrae the cumulative effect is quite significant. Table 3.1, below, shows the natural ranges of angular movement for the three regions of a typical adult spine. It should be noted that the figures quoted are approximations as there is no agreement amongst authors regarding the range of movement at different levels of the column. These values are highly dependent on the individual and vary enormously with age. Hence, the figures in table 3.1 should be read as maximum ranges for a particularly supple person.

Table 3.1: Natural ranges of angular movement for the three regions of the human spine (from Kapandji, 1974).

<table>
<thead>
<tr>
<th></th>
<th>Lumbar</th>
<th>Thoracic</th>
<th>Cervical*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion</td>
<td>60°</td>
<td>45°</td>
<td>40°</td>
</tr>
<tr>
<td>Extension</td>
<td>35°</td>
<td>25°</td>
<td>75°</td>
</tr>
<tr>
<td>Lateral Flexion</td>
<td>±20°</td>
<td>±20°</td>
<td>±(35° - 45°)</td>
</tr>
<tr>
<td>Rotation</td>
<td>±10°</td>
<td>±35°</td>
<td>±(80° - 90°)</td>
</tr>
</tbody>
</table>

*includes C1 and skull

In the lumbar spine, the range of flexion and extension is maximal between L4 and L5 and decreases progressively at higher levels. Thus the lower lumbar column is more active in flexion and extension than the upper column. Flexion is limited by the tension generated in the ligaments of the joints between the superior and inferior articular processes as well as by tension in the ligaments of the vertebral arch, i.e. the ligamentum flavum, the interspinous ligament, the supraspinous ligament and the posterior longitudinal ligament. Extension is limited by the interaction of bony structures on each vertebral arch with those of a neighbouring vertebral arch. Tension developed in the anterior longitudinal ligament also limits the range of extension.

The range of rotation in the lumbar spine is limited by the orientation of the articular facets of the vertebrae and by shearing forces on the intervertebral discs. From table 3.1 it can be seen that for the lumbar column as a whole the range of rotation is only 5E in either a clockwise or anti-clockwise direction, i.e. an average of 1E per segment. In comparison, the range of rotation in the thoracic spine is appreciably greater at "35E which amounts to about 3E for each segment. This greater range of rotational motion is due, in part, to the different orientation of the articular processes in the thoracic region. Furthermore, improved rotational mobility occurs because the intervertebral discs rotate and twist and do not undergo shearing movements as in the lumbar region.

Extension in the thoracic spine is limited by the impact of the articular processes and the spinous processes which, being sharply inclined inferiorly and posteriorly, are normally almost touching. During extension, the anterior longitudinal ligament is in tension while the posterior longitudinal ligament, the ligament flavum and the interspinous ligaments are relaxed. On the other hand, flexion is limited by the tension developed in the interspinous ligament, the ligament flavum, the capsular
ligaments of the joints between the articular processes and the posterior longitudinal ligament while the anterior longitudinal ligament remains relaxed.

Lateral flexion is limited by the impact of the articular processes on the side of the movement and also by the contralateral ligamenta flava and intertransverse ligaments. The bony and cartilaginous attachment of the rib cage to the thoracic column limits its basic movements in all directions.

In the cervical spine, extension is limited by the tension developed in the anterior longitudinal ligament and by the impact of the superior articular process of the lower vertebrae on the transverse process of the upper vertebra and in particular by the impact of the posterior arches through the ligaments.

Flexion is not limited by bony impact but by the tension developed in the posterior longitudinal ligament and other ligaments. During car accidents these ligaments can be severely stretched or even torn and in extreme cases this can lead to anterior dislocation of the facet joints which endangers the spinal cord with the risk of death, quadriplegia or paraplegia.

Table 3.1 shows that a healthy cervical spine allows the head a considerable degree of movement in the three modes - flexion/extension, lateral flexion and rotation. Other modes of movement are rather less well accommodated by the neck. Vertical motion of the head involving axial tension or compression of the cervical spine is hardly allowed at all (the vertical motion experienced when one attempts to see over an obstacle is achieved primarily by straightening the curvature of the spine, rather than stretching it). Axial tension is limited by the stretching of the intervertebral ligaments, while compression is limited by the crushing together of the vertebral bodies. The remaining type of head motion consists of horizontal displacement without rotation. Experience indicates that this is allowed to a small extent in both the sagittal and, to a lesser extent, the lateral directions. It is achieved by means of a shearing motion between the individual vertebrae.

It is reasonable to assume that in motor vehicle accidents in which one or more of the three main areas of the vertebral column are forced to move beyond their natural limits, as indicated in table 1, then damage to the spine is likely to occur. However, in the cervical spine the vertebrae do not necessarily have to move beyond their natural limits in order for disabling injury to result from a motor vehicle accident. The complicated and fascinating subject of ‘whiplash’ injury to the neck will be discussed in later sections.
3.2 STRENGTH OF THE CERVICAL SPINE

3.2.1 Measurement Techniques

Obtaining physical measures of the forces required to produce neck injury or to produce failure of the vertebral components is a difficult, if not impossible task. Human volunteers cannot be exposed to injury-producing situations and relevant in vivo measurements cannot usually be made. Thus, indirect approaches have to be adopted to obtain data which can be related to the overall strength of the neck.

A number of different approaches have been used to obtain neck strength data. A considerable amount of work has been reported on the static and dynamic mechanical properties of cadaveric neck components. The results from such studies can indicate the mechanical performance of a living human neck.

Static strength tests on necks have been conducted with volunteers resisting static loads applied to their heads. Dynamic tests have been performed where volunteers have been subjected to controlled non-injurious acceleration environments. In these latter tests, the torso is restrained and the head is accelerated by the neck. In a similar manner, dynamic tests may be performed using human cadavers with the advantage that the severity of exposure may be increased until physical damage to the neck structure is produced.

In dynamic neck tests it is common practice to measure accelerations of the head relative to the torso whereas in static tests the usual measurement is force applied to the head. Investigators have observed deficiencies in relating injury severity to these measurements. In static tests, the applied load does describe the force level the neck must resist, but doesn’t directly define the resisting bending moment that the neck must develop. The same is true of the accelerations measured in dynamic tests. In order to minimize these deficiencies, Mertz and Patrick (1971) developed a method for calculating the resultant reactions developed between the top of the neck and the base of the skull (occipital condyles) for both static and dynamic tests. Hence direct comparisons of static and dynamic neck reactions could be made.

3.2.2 Static Strength of the Neck

The pioneering work of Mertz and Patrick (1967 & 1971) and Patrick and Chou (1976), using human volunteers, determined the neck’s reaction on the head for statically applied loads to the head. The main results of these studies are presented in table 3.2 which summarises the maximum forces and bending moments developed at the occipital condyles. None of these figures should be read as being upper bounds of non-injury load reactions between the head and neck at the occipital condyles. Tests were usually terminated due to discomfort with the straps used to apply the load to the head and no neck pain or injury resulted. Thus, they are considered to be non-injurious neck reactions and correspond to an Abbreviated Injury Scale (AIS) rating of zero.

Gadd et al (1971) subjected human cadavers to static rearward and lateral neck bending loads. They observed that minor ligament injury occurred 80° of rearward neck bending and 60° of lateral neck bending.
Table 3.2: Maximum static forces and bending moments developed at the occipital condyles by human volunteers (from Mertz & Patrick, 1967, 1971)

<table>
<thead>
<tr>
<th>Bending Torque (Nm)</th>
<th>Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward flexion</td>
<td>50.2</td>
</tr>
<tr>
<td>Extension</td>
<td>20.3</td>
</tr>
<tr>
<td>Lateral flexion</td>
<td>47.5</td>
</tr>
</tbody>
</table>

Force (N)

Anterior-posterior shear | 845
Posterior-anterior shear | 845
Lateral shear | 400
Axial tension | 1134
Axial compression | 1112

3.2.3 Dynamic Strength of the Neck

Mertz and Patrick (1971) and Patrick and Chou (1976) have conducted tests on volunteers and human cadavers to determine the neck’s reaction on the head under dynamic conditions. The main results of these studies are presented in table 3.2 for volunteers and table 3.3 for human cadavers. The bending moment for forward flexion includes the moment of the chin force, at chin contact, taken with respect to the occipital condyles.

Table 3.3: Tolerable neck reactions calculated at the occipital condyles for dynamic volunteer tests (From Mertz and Patrick (1971) and Patrick and Chou (1976))

<table>
<thead>
<tr>
<th>Loading Configuration</th>
<th>Neck Bending Moment (Nm)</th>
<th>Neck Shear Force (N)</th>
<th>Axial Force (N)</th>
<th>Head Angle Relative to Torso (degrees)</th>
<th>AIS</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward bending 0°</td>
<td>88.2</td>
<td>787</td>
<td>---</td>
<td>70</td>
<td>1</td>
<td>Pain, no injury</td>
</tr>
<tr>
<td>Rearward bending 180°</td>
<td>30.5</td>
<td>231</td>
<td>249</td>
<td>68</td>
<td>0</td>
<td>No injury</td>
</tr>
<tr>
<td>Lateral bending 90°</td>
<td>45.2</td>
<td>792</td>
<td>---</td>
<td>43</td>
<td>0</td>
<td>No injury</td>
</tr>
<tr>
<td>Lateral bending 135°</td>
<td>18.0</td>
<td>311</td>
<td>356</td>
<td>---</td>
<td>0</td>
<td>No injury</td>
</tr>
<tr>
<td>Lateral bending 45°</td>
<td>31.2</td>
<td>440</td>
<td>165</td>
<td>---</td>
<td>0</td>
<td>No injury</td>
</tr>
</tbody>
</table>
Table 3.4: Neck reactions calculated at the occipital condyles for dynamic human cadaver tests
{From Mertz and Patrick (1971) and Nahum et al (1968)}

<table>
<thead>
<tr>
<th>Loading Configuration</th>
<th>Neck Bending Moment (Nm)</th>
<th>Neck Shear Force (N)</th>
<th>Axial Force (N)</th>
<th>Head Angle Relative to Torso (degrees)</th>
<th>AIS</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward bending 0°</td>
<td>190</td>
<td>1588</td>
<td>---</td>
<td>88</td>
<td>0</td>
<td>No damage</td>
</tr>
<tr>
<td></td>
<td>176</td>
<td>1944</td>
<td>---</td>
<td>69</td>
<td>0</td>
<td>No damage</td>
</tr>
<tr>
<td>Rearward bending 180°</td>
<td>47</td>
<td>---</td>
<td>---</td>
<td>---</td>
<td>0</td>
<td>No damage</td>
</tr>
<tr>
<td></td>
<td>57</td>
<td>---</td>
<td>---</td>
<td>---</td>
<td>3</td>
<td>Ligamentous damage</td>
</tr>
</tbody>
</table>

Mertz and Patrick (1967 and 1971) found that the resultant bending moment was a good indicator of neck strength. Based on their cadaver data, they proposed tolerance levels for the 50th percentile adult male. For flexion a resultant bending moment of 190Nm was suggested as a lower limit for an injury tolerance level. This bending moment did not produce any discernible ligamentous damage to a human cadaver. For extension, they proposed, an injury tolerance level of 57Nm. Although this level was associated with ligamentous damage in a human cadaver, the cadaver used was relatively old and therefore it is likely that there would have been some degeneration of the strength of the ligamentous tissue in comparison with living tissue. The results of Mertz and Patrick suggest that the neck is three times stronger in resisting flexion than extension.

The response curves and tolerance limits determined by Mertz and Patrick (1971) were later used in the development of the Hybrid III dummy (see chapter 8). They concluded that the angle between head and torso is not a good indicator of injury risk because, in hyperextension or hyperflexion, a small increase in angle produces a large increase in resisting torque, presumably as the ligaments start stretching beyond their normal range. They point out that there is no unique neck response curve for an individual because of the degrees of muscle tone that a person is capable of generating. The response is most repeatable with the neck muscles tensed, so they say that this condition should be used to specify dummy neck responses.

In contrast to this, Kallieris et al (1996) report sled tests on cadavers (front, side and rear impacts), in the mid to severe range (25kph Δv for rear impacts - higher for other impact directions). The neck loads at the occipital condyles were calculated from the observed head kinematics in the same way as Mertz & Patrick, and they related these loads to the observed injuries. The calculated moments in the two rear impact tests were 20 and 32Nm. One of these gave an AIS3 neck injury, contradicting the non-injury threshold of 48Nm suggested by Mertz & Patrick. They comment that the muscular and neurological injuries associated with whiplash are not very well reproduced by cadavers.

Yoganandan et al (1995) used cadaver tests to look at severe neck injuries, especially spinal cord damage. They used a device to measure spinal cord pressure during rear impacts (impact speeds of 11 and 22kph). Injuries observed included vertebral fractures. They suggest a possible injury threshold for spinal cord damage at 0.35MPa.

Ewing and Thomas (1973) have also conducted dynamic forward neck bending tests with instrumented volunteers. Their testing was focussed mainly on determining the inertial response of the neck to accelerations of between 2 and 7g. In their more severe tests, they calculated the maximum forward neck bending moments relative to the occipital condyles. Three of the volunteers developed maximum bending moments of 35Nm, 45Nm and 50Nm without any pain. These values are consistent with the result for forward bending moment in table 2 which yielded an AIS 1 level neck injury.
The results from cadaver tests should always be viewed with caution because there are significant differences in the inertial response of a cadaver and that of a live human to impact. The neck structure of a cadaver is flaccid. Its neck muscles cannot transmit any significant load even when the neck is hyperflexed. All the neck loads must be transmitted by the bony vertebrae, the intervertebral discs and the surrounding ligaments which retain their mechanical properties reasonably well for a short time after death. In the living human, neck muscles can transmit loads and thereby share the load distribution with other neck structures. The main implication of this difference is that while the cadaver will mimic the neck damage patterns of the human, the cadaver damage will occur at lower collision severity levels. Thunnissen et al (1995) compared the head and neck motion of human volunteers with that of cadavers and, hence were able to formulate the ideal performance requirements for a dummy neck - head system. Comparing the performance of the Hybrid III neck with these newly defined performance requirements, they concluded that the Hybrid III neck is too stiff.

3.3 THE KINEMATICS OF THE HUMAN SPINE

Much of the early work on the dynamic behaviour of the human spine was conducted at the Naval Biodynamics Laboratory (NBDL) in New Orleans by Ewing et al (1977,1978) who subjected volunteers to sled test accelerations from 2g to 15g. These early researchers investigated the way in which the head and neck motion related to the sled pulse. It was soon realised that the response of the thoracic spine to an impact influences the type of response and the damage patterns produced in the cervical spine.

A large quantity of the NBDL data was subsequently analysed by Wismans et al (1983) at TNO who investigated the relationship between the neck motion and the T1 acceleration. As a result of their work, T1 acceleration is now used as the input in neck response tests. Wismans et al (1987) subsequently compared the volunteer data with corresponding data from cadaver tests and observed a larger amount of head rotation in the cadaver tests because of the lack of muscle tone. Bendjellal et al (1987) have conducted sled tests on cadavers to investigate the kinematic response of the neck to high g-level lateral accelerations with a view to defining a specification for dummy neck design.

Some researchers have used imaging techniques to study the movement of the human spine. Margulies et al (1992) used magnetic resonance imaging to track the normal motion of the head and neck in extension and flexion. They concluded that flexion is characterized by a slight lengthening of the spinal cord relative to its neutral position accompanied by a variable displacement of the upper cervical cord and motion of the lower cord towards the head. Conversely, extension causes a shortening and movement of the spinal cord away from the head.

Ono and Kaneoka (1997), in Japan, used cineradiography (90 frames/s) to record visually the response of the cervical vertebrae of human volunteers in sled tests which were designed to simulate car rear-end collisions. They found that the cervical spine tended to form a characteristic ‘S-shape’ between 50 and 100ms after the start of the impact. This has been identified as a point at which injurious damage may occur to the neck {Svensson et al (1993), Panjabi et al (1998)}. Complementary high speed video of the tests conducted by Ono and Kaneoka (1997) showed that the subject’s upper torso is pushed against the seatback and is forced to move upwards along the incline of the seatback. This behaviour is a characteristic feature of rear impacts and is often observed in high speed films of sled tests. It was reported by Begeman et al (1980) in a comparative study of sled tests using human volunteers, cadavers and dummies. The effect has also been replicated in simulation models ( van den Kroonenberg et al, 1997). The Japanese study demonstrated the importance of the T1 motion which initiates the axial compression of the neck.
3.4 MECHANICAL TESTING OF CADAVERIC SPINAL COMPONENTS

Static and dynamic testing of cadaveric vertebral components has been widely reported. Much of this work has been conducted on motion segments (also known as functional spinal units, FSU) which consist of two adjacent vertebral bodies and the intervening disc. Other researchers have chosen to work with complete head and neck structures {e.g. Yoganandan et al (1995)}.

The vertebral bodies are stronger than the discs in tension while the reverse is reported in compression (Yamada, 1973). In tension, failure is routinely observed at the disc end plate (i.e. the junction between the disc and the vertebral body) at forces from 863N at cervical levels to 3000N at lumbar levels. In compression, the vertebral bodies fail at 3000N at cervical levels to 5000N at lumbar levels.

More recent investigations into the mechanical strength of cadaveric components may be found in the literature. Yoganandan et al (1990) have compared their measurements of vertical failure loads in compression of a head-neck complex with the results from equivalent cadaveric tests by other researchers. Moroney et al (1988) have determined the load versus displacement characteristics of lower cervical spine motion segments in compression, shear, flexion, extension, lateral bending and tension. They found that stiffness decreases with degeneration. Schultz et al (1979) have also reported on the mechanical properties of human lumbar spine segments. Their work was based on tests on 42 fresh cadaver segments in flexion, extension, lateral bending, torsion, compression, anterior shear, posterior shear and lateral shear. Yoganandan et al (1991) obtained force versus deflection data from 8 fresh human cadaveric head/neck complexes from compression tests to failure at 2.5mm/s. Failure forces ranged from 1.3kN to 3.6kN.

The tensile properties of ligaments have also been investigated {Chazel et al (1985) and Myklebust et al (1986)} In tension, the load versus elongation curve is non-linear and inconstant.

Osvalder et al (1992) and Osvalder and Neumann (1993) at Chalmers University, Gothenburg have conducted dynamic pendulum tests on motion segments of the lumbar spine. Failure of the average segment was observed at a bending moment of 151Nm and a mean shear force of 481N and a flexion angle of 19E. It was found that the response of the lumbar motion segment to a dynamic flexion shear is dependent on the bone mineral content in the vertebrae and the size of the segment.

Pintar et al (1995) have measured spinal cord pressures by placing instrumented artificial spinal cords within human cadaveric head/neck complexes. The specimens were loaded dynamically in compression. Below a spinal cord pressure of 30N/cm² no failure of the neck occurred. Minor cervical spine injuries were observed at pressures between 35 and 75N/cm² while fractures and dislocations were associated with pressures of 50 to 200N/cm².

Further data on the mechanical properties of the cervical spine in relation to mathematical modelling is presented in sections 7.4.4 and 7.4.5.

3.5 BIOMECHANICS OF THE THORACIC AND LUMBAR SPINE

Injury to the bony portions of the thoracic and lumbar spine (often referred to collectively as the thoracolumbar spine) is relatively rare in automotive accidents. In such incidents injury to the neck is more likely and consequently most of the literature on the biomechanics of the spine deals with the cervical region rather than the thoracic and lumbar regions. Nevertheless, the response of the thoracolumbar spine to an impact determines the loading in the neck and hence the eventual injury outcome. This point indicates the need for biofidelity in the thoracic and lumbar regions in the design of any new dummy spine.

King (1993) has produced a review of the biomechanical response of the thoracolumbar spine. In his review, King stresses the important role of the facet joints in providing a load path for compressive
axial loads applied to the vertebral column. In compression the facet joints behave like a stiffening spring while in tension they offer very little resistance. Thus in tension most of the mechanical resistance is provided by the ligamentum flavum and the interspinous and supraspinous ligaments (Yang and King, 1984). It should be noted that for individual facets axial compression is equivalent to spinal extension and axial tension to spinal flexion. This behaviour has an important bearing on the understanding of injury mechanisms of the spine.

There have been several investigations of the mechanical behaviour of isolated thoracolumbar spines and thoracic and lumbar motion segments under vertical loading (Sances et al (1983) and Myklebust et al (1982)). It was found that the isolated thoracolumbar spines failed under compression loads from approximately 800N to 5100N. With vertical forces applied to the upper thoracic region of intact seated cadavers, thoracolumbar fractures were observed with forces from 1554N to 2750N. The fractures were predominantly in the low thoracolumbar areas but more injuries in the upper thoracic spine were observed when the cervical elements were included (Myklebust et al, 1982).

Studies of single thoracolumbar vertebral bodies in compression have indicated a failure range between 1900 and 15,700N {Kazarian and Graves (1977) and Lin et al (1978)}. Freeman (1998) quoted mean compressive fracture loads for thoracolumbar vertebral bodies of 7600N for humans below the age of 40 and 4200N for those over 60. The thoracic spine is stiffer than the lumbar spine and this is reflected in the respective ranges of motion presented in table 3.1.

3.6 CONCLUDING REMARKS

A good knowledge of the biomechanical properties of the human spine is crucial in any project to develop a physical or mathematical model of the spine. Thus, the data on the natural ranges of motion, stiffnesses and yield strengths of the various parts of the spine, described above, will have an important bearing on the current project to design a more biofidelic spine for the Hybrid III dummy. The preliminary perusal of the literature reviewed in this chapter has revealed a shortage of good data on the mechanical properties of spinal tissue. In the future phases of the project, it may be worthwhile conducting some experimental tests on human spines. The results of such tests could then be used to enhance the accuracy of the mathematical models developed during the course of this work. However, the matter is not straightforward since experiments using human volunteers may involve a risk of injury and tests on cadaveric spines carry ethical implications.

3.7 REFERENCES


Dummy Development: Spinal Injuries


4 CERVICAL SPINE INJURIES

4.1 INTRODUCTION

From a mechanical and structural point of view, the cervical spine is a very complex mechanism. The human neck contains vital neurological, vascular and respiratory structures as well as the cervical vertebrae and spinal cord. Although injury statistics only attribute two to four per cent of serious trauma to the neck (McElhaney & Myers, 1993), any neck injury can have debilitating, even if not life-threatening, consequences. Permanent paralysis is a particularly devastating and costly injury. Neck injuries are frequently associated with automobile accidents, with rear-end collisions presenting the highest risk. Data from the US National Spinal Cord Injury Data Research Centre indicates that automobile accidents are responsible for 36.7% of spinal cord injuries (McElhaney & Myers, 1993). In the UK, Grundy & Swain (1993) report that up to half the admissions to spinal trauma units are from road traffic accidents (RTAs). Even when the vertebrae themselves are not crushed or broken, damage to the soft tissues between and around the vertebrae can occur. Such soft-tissue cervical sprain injuries, though generally only classified as 1 on the Abbreviated Injury Scale, have been highlighted as having long-lasting and disabling effects. Hopkin et al (1993) have shown that over half of all car occupants involved in RTAs had some degree of neck injury, mostly AIS 1 neck strains, and Ono & Kanno (1993) report a similar proportion in Japan. Nygren (1984) showed that, despite their low AIS rating, neck strain injuries lead to permanent disability (disability degree \( \geq \) 10%) in some 10% of cases. By comparison, the risk of permanent disability associated with other AIS 1 injuries is only about 0.1% (Nygren et al, 1985). Hamer et al (1993) report that neck strain injuries can predispose patients to premature degenerative disc disease some years after the initial injury. These neck strain injuries are frequently referred to as ‘whiplash’ injuries, due to the rapid flexion-extension motion of the head and neck which is usually taken to be the cause of them. It is evident that these injuries are common and they also appear to be increasing (Galasko et al, 1993, 1996, Maag et al, 1993, Morris & Thomas, 1996, Otte et al, 1997).

There is evidence from accident studies of a degree of bi-modality in neck injuries. Several authors report that, among people who die from neck injury, those injuries tend to be concentrated at C2 and above. People who survive (serious) neck injury, on the other hand, tend to have been injured at C3 or below, and particularly around C5/C6 (McElhaney & Myers, 1993, Yoganandan et al, 1995, Huelke et al, 1993).

4.2 BIOMECHANICS OF CERVICAL SPINE INJURY

The mechanisms of cervical spine injury have been extensively researched. However, due to the complexity of this section of the spine, particularly with respect to the range and omnidirectionality of natural movement, the exact cause of any particular injury is difficult to pin-point. The mechanical properties of the spine are very difficult to define, having, as it does, a segmented structure composed of non-linear viscoelastic elements and kinematic elements. The motions of the elements are coupled, and small loads (in physiologically allowed directions) produce large strains. In addition to a significant component of viscoelasticity, the spine demonstrates a second component of time-dependence - pre-conditioning. Over times of the order of hours, the soft tissues absorb fluid which results in increased stiffness. During activity, this fluid is squeezed out of the structure, and the stiffness decreases. This phenomenon should be taken into account in cadaver testing (McElhaney et al, 1988). Another problem in relation to cadaver testing is that, in response to an externally-applied force, the neck will frequently buckle, sometimes in unexpected ways, so that a gross extension motion (bending backwards) can produce vertebral fractures which are more typical of flexion. The explanation for this is that the neck has buckled so that two or three vertebrae are actually in flexion while the rest are in extension. Further problems are due to the fact that small changes in the initial
position of the head relative to the torso (<1cm) prior to impact loading can completely change the injury mode (eg from compression-flexion to compression-extension). The degree to which the head is allowed to move during loading also has a large influence on the resultant injury. If the head is allowed to dig into soft ground or soft padding and thus becomes constrained, increased injury to the neck usually results (Nightingale et al., 1991). The following description of the chief injury modes is taken largely from a review by McElhaney & Myers (1993).

4.2.1 Modes of Injury

4.2.1.1 Compression
Due to the buckling discussed above, pure compression probably occurs relatively rarely. Typical compression injuries include multipart fractures of the Atlas (frequently fatal), burst fractures of the remaining cervical vertebrae (C2-C7) and compression injuries to the intervertebral discs. Ligament damage is rare. The risk of spinal cord damage varies from 40% to 75%, depending on the degree of comminution of the vertebral fracture. The injury is often associated with rollover accidents, particularly for unrestrained occupants.

4.2.1.2 Compression-flexion
Typical injuries from this form of loading include wedge and burst fractures of the anterior portions of the vertebrae, and also anterior dislocation. These usually occur in C5-C7, due to the lower neck being constrained by the less flexible thorax. Anterior dislocation can be bilateral (ie involving both facet joints, and frequently associated with cord damage) or unilateral. The latter has been reported to be associated with combined flexion and head rotation, although head rotation alone is unlikely to produce sufficient torsion in the lower cervical vertebrae, due to the relative weakness of the atlanto-axial joint in this mode. Cord damage is uncommon in unilateral dislocation.

4.2.1.3 Compression-extension
This type of loading produces fractures of the posterior elements of the cervical spine, such as the laminae, the pedicles and the spinous processes, and can occur in the upper or lower neck. The fractures are frequently multiple and can result in cord damage due to bone fragments being pushed into the spinal canal.

4.2.1.4 Tension
Pure tension injuries are almost exclusively restricted to atlanto-occipital distraction (pulling apart of the uppermost cervical vertebra from the base of the skull), with unilateral or bilateral dislocation of the occipital condyles (bones which join the skull to the neck). They are usually lethal due to the associated distraction and transection of the spinal cord. This type of injury is associated with rapid decelerations in motor vehicle accidents, though its prevalence is debated. McElhaney & Myers claim that the method of dissection at autopsy can easily fail to detect atlanto-occipital dislocations, so their occurrence may well be under-reported. Bucholz & Burkhead (1979) found this injury in 9 out of 112 automotive fatalities and they claim, on this basis, that it is a fairly common cause of death. Yoganandan et al (1989) report that this injury is associated with passenger ejection during motor vehicle crashes. The injury also occurs in military pilots involved in high-speed crashes into water, where calculations indicate that the deceleration required to produce the injury exceeds 100g.

4.2.1.5 Tension-extension
This type of loading is said to be common and to occur in one of three ways: when the head is arrested while the torso continues moving forward (eg unbelted vehicle occupants hitting the windscreen, also falls and dives), when the torso is suddenly accelerated forward, resulting in inertial loading of the neck by the head (whiplash mechanism), and when the chin is forced upwards and backwards, as in judicial hanging (hangman’s fracture). Whiplash injuries are typically associated with rear impact vehicle crashes of fairly low severity, and will be considered in more detail below. Higher severity rear impacts can cause damage to the anterior longitudinal ligament and intervertebral discs as well as
horizontal fractures through the vertebrae. The relative risk of ligamentous damage as opposed to bony fracture probably depends to a large extent on bone mineral content. The hangman’s fracture is a fracture of the posterior arch of C2. It can also be caused by blows to the face and chin. Spine fracture or dislocation during extension frequently results in cord damage, although there are documented cases of cord injury in the absence of structural injury (Taylor & Blackwood, 1948). These are thought to be due to bulging of the compressed flaval ligaments (which run along the rear wall of the spinal canal) or the discs into the spinal canal, impinging on the spinal cord.

4.2.1.6 Tension-flexion
This produces injuries similar to those described for compression-flexion, indicating that the flexion component is probably more important than direction of the associated axial loading.

4.2.1.7 Torsion
Injuries due to torsion tend to be confined to rotary atlanto-axial dislocation, with or without tearing of the associated ligaments. As mentioned above, the weakness of the atlanto-axial joint in torsion precludes pure torsion as a mechanism of injury to the lower cervical spine. The contribution of torsion in predisposing the lower cervical spine to injury from other types of loading, although a common belief, is also questioned by McElhaney & Myers, due to the observed absence of significant torques at axial rotations up to "67° from the neutral position. They claim that the head rotations observed in cadaver and primate experiments occur after injury has taken place, and are therefore the result, rather than a contributory cause, of injury.

4.2.1.8 Horizontal Shear
When subjected to shear forces, the cervical spine tends to fail at the atlanto-axial joint, with anterior and posterior subluxations, rupture of the transverse ligament (which confines the odontoid process to the anterior part of the hole in the middle of the atlas) and, in extreme cases, fracture of the odontoid process. These injuries are potentially lethal due to spinal cord involvement, and are very difficult to stabilise surgically.

4.2.1.9 Lateral Bending and Lateral Shear
Injuries due to these types of loading are much less common than those due to loading in the sagittal plane. This is probably due to the relatively lower incidence of lateral loading and to the flexibility of the neck in lateral bending.

4.2.2 Whiplash Injury
4.2.2.1 Characteristics of Whiplash Injury
The term "whiplash injury" is generally felt to be unsuitable by workers specialising in this area, because it implies a well-defined mechanism. In fact, the mechanism of this type of injury is extremely ill-defined, and the use of the term has led some to maintain that the injury cannot exist in the absence of a whiplash-type motion of the head and neck, contrary to observational experience. A term which relates to the injury itself, rather than its presumed causal agent, would be preferable, but here a second problem arises, in that the actual injury is also very poorly defined. Patients typically show normal X-rays (apart from an occasional slight straightening of the normal lordotic curve of the cervical spine) and neither computed tomography nor magnetic resonance imaging provide reliable diagnoses. Panjabi et al (1998) describe whiplash injuries as ‘sub-failure’ injuries - ie there is no complete failure of any particular anatomical structure.

The injury is, in effect, defined by its symptoms, which can include pain in the neck or shoulders, headaches, blurred vision, tinnitus, dizziness and numbness in the upper limbs (Bogduk, 1986), and in some cases these can persist for years (Larder et al, 1985, Gargan and Bannister, 1990, Murray et al,
Spitzer et al (1995) have proposed the term ‘Whiplash-Associated Disorder’ (WAD) for the set of symptoms associated with what they refer to as ‘Whiplash Injury’ - a term which remains undefined.

As well as physical symptoms, whiplash injuries are frequently accompanied by psychological problems similar to those experienced by much more severely injured accident victims (Mayou & Bryant, 1996). Travel anxiety and phobia is particularly associated with whiplash injury, and can result in very significant social impairment.

Whiplash injuries are characteristic of low speed impacts (Kahane, 1982, Larder et al, 1985, Otte and Rether, 1985, Morris and Thomas, 1996) - cases have even been reported below 10km/hr (Olsson et al, 1990, Ryan et al, 1994) - where, due to the application of modern restraint technology and improvements in vehicle design, there is an expectation that no serious injury should occur. Indeed, as measured by the AIS scale, serious injuries do not occur in many of these accidents. Furthermore, a common feature of Whiplash Associated Disorder is the delayed onset of symptoms (Larder et al, 1985, Parmar and Raymakers, 1993, Robinson and Cassar-Pullicino, 1993), so many of these accidents may initially be classified as non-injury (Maag et al, 1993). However, the degree of potential impairment associated with this type of neck injury has been shown to be far greater than would be expected from an initial assessment of injury severity or vehicle damage.

Women are generally acknowledged to be more at risk than men, possibly due to the weaker neck musculature of the average female (Larder et al, 1985, Otremski et al, 1989, Lövsund et al, 1988, Morris and Thomas, 1996) although, since the exact mechanism of injury is unknown, it is perhaps rash to postulate a mechanism to explain differences in risk. Minton et al (1998) showed that, amongst a whiplash-injured population, women suffered significantly greater disability as a result of their injuries than men.

Lövsund et al (1988) have shown that the risk of incurring whiplash injury is higher in rear impacts, despite the fact that rear impacts are, on average, less severe than impacts in other directions. However, Morris and Thomas (1996), while agreeing that whiplash injury risk is undoubtedly greater in rear impacts, have shown that frontal impacts nevertheless produce greater absolute numbers of WAD victims because the number of frontal impacts which occur is much greater.

Seat belt use has been found to be associated with increased whiplash injury risk (Larder et al, 1985, Otte and Rether, 1985, Bunketorp et al, 1985, Otremski et al, 1989), although Maag et al (1990) point out that the introduction of belts has reduced the incidence of severe neck injuries associated with head contacts in frontal impacts. Galasko et al (1993) found an increase in the incidence of whiplash injury from 8% to 21% associated with the sudden rise in UK belt wearing rates following introduction of compulsory belt wearing legislation.

4.2.2.2 Mechanisms of Whiplash Injury

The whiplash mechanism in a rear-end car collision appears to have been first described by Crowe (1928). Experimental attempts to find a mechanism for whiplash injury seem to have begun with McNab (1964), who carried out tests on anaesthetised monkeys. He found a predominance of anterior element injuries, which he ascribed to hyperextension of the entire cervical spine. Bogduk (1986) describes this classical view of whiplash kinematics in a rear impact, without the presence of a head restraint. The seat, which is more or less rigidly attached to the vehicle, moves forward relative to the occupant, who initially sinks into the seat padding. By the time the padding ceases to yield, the occupant’s torso, up to shoulder level, must have acquired the same forward velocity and acceleration as the seat back. The head, meanwhile, being only relatively loosely coupled to the shoulders by the seven separate cervical vertebrae and their associated soft tissues, and with no external forces acting upon it,
The neck continues to distort into a backwards curve, or extension until enough force is exerted through the head’s centre of gravity to accelerate it up to the velocity of the shoulders. At this point, however, the energy stored in the elastic components of the neck structure continues to accelerate the head and catapults it forward, past the shoulders, into a forward curve, or flexion. It is this very rapid backward-forward motion, rather reminiscent of the cracking of a whip, which has given rise to the term ‘whiplash’ injuries to describe loosely the set of symptoms which are frequently associated with such an event.

States et al (1969) postulated that the hyperextension described by McNab could be exacerbated by elastic rebound from the seat back. As discussed above (Secs. 2.1.3, 2.1.5) severe hyperextension, accompanied by either compression or tension, can produce serious injuries to the vertebrae. It is assumed that what is bad for the vertebrae is, in less severe circumstances, also bad for the surrounding soft tissues. Mertz and Patrick (1967) showed that eliminating head motion relative to the torso completely, by having a volunteer’s head permanently in contact with a high, rigid seat back, allowed very severe rear impacts to be survived without ill effect. Hence, it has been postulated, and widely accepted, that the provision of a head restraint which will prevent rearward hyperextension of the neck will prevent the occurrence, not only of severe neck injuries caused by hyperextension, but also of the less severe soft tissue injuries. In 1969, the US government made the provision of head restraints mandatory in all new vehicles.

However, it has been widely reported (Larder et al, 1985, Otte and Rether, 1985, Otremski et al, 1989, Maag et al, 1990, Foret-Bruno et al, 1991, Ryan et al, 1994, Von Koch et al, 1995, Morris and Thomas 1996) that neck injuries which result in symptoms indistinguishable from those observed in rear impact victims can also occur in frontal and side impacts Although rearward hyperextension may occur on rebound from the seat belt in impacts with a frontal component, the risk is likely to be a smaller, due to energy loss during forward excursion into the seat belt, and because neck muscles are more likely to have time to react by the time the rebounding torso strikes the backrest.

Aldman (1986), proposed that the most harmful event occurs early in the motion sequence, when the occupant’s head is translating non-rotationally backward relative to the shoulders, and in the very early stages of subsequent head rotation. The shear forces produced in this early stage cause the neck to distort into an S-shape, and this can also happen in frontal impacts (Walz and Muser, 1995). The transition from the S-shape to the extension mode, as the Atlas begins to exert a horizontal component of force on the base of the head, causing the head to rotate, involves a sudden change in the volume of the spinal canal, and it has been proposed that the pressure gradients induced by the sudden and rapid flow of blood and spinal fluid along the canal and through the associated transverse vessels can result in damage to the nerve roots emerging from the spinal column. An essential feature of this theory is that, because the mechanism is related to fluid flow, the important factors in determining injury risk are the velocity and acceleration of C1 relative to T1, rather than simple displacements.
Penning (1992) also considered the pre-rotational shearing of the neck to be important. However, rather than the pressure-flow characteristics of the cerebro-spinal fluid, he concentrated on osteoligamentous injuries, observing that the displacements of the C0-C2 joints in shear were larger than those of the lower cervical vertebrae, and also larger than the movements of the C0-C2 joints in flexion or extension. These measurements were conducted using X-rays on volunteers exhibiting normal, voluntary motion. He concluded that whiplash is caused by shear damage to the intervertebral joint tissues in the upper neck.

Since, as mentioned above, whiplash injuries have been observed in frontal as well as rear impacts, a mechanism for whiplash must take account of occupant motions in frontal crashes. Von Koch et al (1995) proposed that the prime injurious event is the forward flexion of the neck caused, in frontal impacts, by the sudden deceleration of the torso by the seat belt. In rear impacts, the rebound observed by States et al (1969) has probably become even more pronounced as seat backs have been strengthened to provide better protection in high severity impacts, again resulting in the torso being catapulted into the seat belt. Although no detailed mechanism was proposed by von Koch et al, an injury mediated by forward deceleration would tie in with the observed higher incidence of whiplash injuries in belted occupants (Larder et al, 1985, Otte and Rether, 1985, Bunketorp et al, 1985, Otremski et al, 1989, Maag et al, 1990,).

McConnell et al (1995), on the basis of a very detailed set of volunteer impact tests, at speeds up to 10.9kph, all but dismiss rearward hyperextension as a mechanism for whiplash injury. They point out that, once the head has started to rotate, due to the horizontal force vector applied by the neck not passing through the head’s centre of gravity, the head’s rotational inertia will eventually take it past the point at which the neck’s force vector passes through the C of G. From this point on, the neck is exerting a corrective moment on the head and, in low-velocity impacts, this occurs while the neck is still well within its physiological range of movement, thus precluding hyperextension as a cause of whiplash injury. All their volunteers experienced some slight whiplash symptoms as a result of the tests, though none of them at any time came within 10° of their normal voluntary maximum range of movement. They propose a simple tripod model for the head/neck structure, with a ball (the head), supported on a segmented spinal column (compressed by the weight of the head), and braced from falling over backwards by the two sternocleidomastoid muscle groups, forming the front two legs of the tripod, and normally under tension. The whole structure is supported on a solid block (the torso). When this block is accelerated, the muscle groups are suddenly strained, and must react, while the spinal column is further compressed. High speed film of the volunteers (who did not know exactly when the impact would take place, and who were asked to relax) shows these muscle groups bulging in an attempt to resist the suddenly imposed load, which occurred 110-120ms after impact. The ultimate mechanism of injury is postulated to be muscular strain, possibly accompanied by some compression damage in the intervertebral tissues.

Author’s Comment: It is possible that, even in a completely unaware subject, the sudden noise of impact could be sufficient to cause the autonomous nervous system (which by-passes the brain’s awareness centres) to send all the neck muscles into tension within the time-scale discussed by McConnell et al. However, as impact speed is increased, there must come a point at which the muscles would be unable to react in time. This may limit the applicability of this model.

Ono & Kaneoka (1997) also claim that cervical spine extension in rear impact does not exceed the normal range of motion when a head restraint is used, but point out that whiplash injuries are still happening. In sled tests on volunteers, at speeds up to 6kph they used X-rays at 90 frames per second to measure vertebral motions. An electromyogram was used to check muscle tension. The volunteers were also X-rayed while performing voluntary neck movements to act as a baseline for the dynamic tests. Under impact conditions, they observed initial compression of the spine due to the torso ramping up the seat back, and this produced slight flexion in the neck, followed by compression/extension motion between C3 and C6 as the torso was accelerated forward, leaving the head behind. They confirmed the development of the “Aldman” S-shape, but note that detailed cervical vertebral motion is heavily dependent on initial posture - ie whether the neck is flexed or
extended initially (regardless of the proximity of the head restraint) and by the extent of torso ramping. In normal voluntary motion, the order in which the individual vertebrae rotate into extension is from top to bottom (C1 to C7), with the C4/C5 joint rotating more than any of the others. In impact motion on the other hand, the order of rotation is reversed, because the neck is responding to an input from T1, and the impact pulse will travel up the cervical column. Thus, C6 undergoes a large rotation initially, then C5 rotates until the C5/C6 angle is much larger than normal. They conclude that the prime injury site in whiplash injury may be the C5/C6 joint, and that the initial curvature of the neck and the extent of ramping may be as important as head restraint positioning in determining injury outcome.

Panjabi et al (1998) conducted tests on cadaveric cervical spine specimens, which were mounted on a mini-sled, with a lump of steel attached to C1 to represent the head. Each vertebra was flagged and high speed film was used to follow the vertebral motions when the sled underwent simulated rear impacts, with accelerations rising from 2g-10g in 2g increments. Intervertebral joint damage was assessed by carrying out quasi-static flexibility tests between each progressively more severe impact test. They confirmed the development of the ‘Aldman’ S-shape in the early stages of motion. The C5-C6 joint suffered more severe damage than other joints, in terms of increased range of motion in flexibility tests. In addition, the maximum extension angles between the joints of the lower cervical spine (significantly outside the normal physiological range) occurred while the neck was still in the S-shape, before the phase shift to full extension, and the C0-C2 extension did not go beyond physiological limits. Overall head extension did not exceed physiological limits in any test. In the very high severity impacts, the C0-C1 joint also began to show signs of damage. They conclude that whiplash injury is probably related to ligament and muscle damage in the lower cervical spine.

Barnsley et al (1995) report work on whiplash victims involving diagnostic blocks of the intervertebral joints, by injection of local anaesthetic. Pain symptoms were relieved in 54% of the patients, indicating that their whiplash symptoms originated in these intervertebral joints.

### 4.3 EXPERIMENTAL WORK

#### 4.3.1 Swedish Studies:

A series of papers from researchers at Chalmers University, Sweden, and other Swedish research organisations (Svensson et al, 1989, Svensson et al, 1993a) describe work carried out to investigate the Aldman hypothesis concerning transient pressure/flow surges in the cerebro-spinal fluid (see above). They have subjected anaesthetised pigs to a whiplash-like motion of the head, and discovered associated damage to the spinal nerve ganglia. The observed over-pressures in the (instrumented) spinal canal were highest in the lower half of the cervical spine, and this correlated with the location of the injuries observed. They claim that this nerve root damage may account for the neurological symptoms suffered by humans (eg radiated pain, pins and needles) if this type of damage also occurs in human whiplash victims. However, it is impossible to detect this type of damage in live humans - the pigs had to be killed and dissected. A problem with the experimental method is that the pigs’ torsos were stationary, and their heads were pulled mechanically. There must be some doubt as to how well the resultant head motion duplicates the inertial behaviour of an unconstrained head when the torso is accelerated. However, the existence of neurological symptoms in whiplash victims indicates that the injury mechanism is unlikely to be simple muscle or ligament damage, and this work has shown that neurological damage can be sustained in the absence of physical damage to load-bearing neck structures.

Further developments of this work have involved sled tests on volunteers and the development of a new dummy neck (the Rear Impact Dummy - RID - neck) which is capable of adopting the S-shape observed in the necks of the volunteers. Svensson et al (1993b) tested a number of front and rear production car seats using a Hybrid III dummy fitted with a RID neck in rear impact sled tests. They conclude that rearward displacement of the head relative to the torso does not correlate with whiplash
injury risk since the maximum displacement occurred in the rear rather than the front seats, and rear seat occupants are known to be at lower risk of whiplash injury from accident studies (Carlsson et al., 1985, Otremski et al., 1989). A problem encountered with the dummy was that the lower neck force/moment transducer protruded to such an extent that, in the higher speed tests, it hit the upper seat back frame and brought the torso to an abrupt stop (relative to the seat), influencing the subsequent head/neck motion. However, in tests where the transducer did not hit the seat back structure, the head acceleration relative to the torso was lowest in the rear seats, suggesting that head/neck acceleration, rather than displacement, is related to whiplash injury risk, and this fits in with the Aldman hypothesis.

Boström et al (1996) describe further experiments on pigs and, using the standard equations of fluid mechanics, derive a Neck Injury Criterion:

\[ NIC = 0.2a_{rel} + v_{rel}^2 \]

where \( a_{rel} \) is the relative acceleration between C1 & T1, and \( v_{rel} \) is the time-integral of \( a_{rel} \), set to zero at the moment of impact. The value of NIC is to be calculated at the moment of maximum retraction of the head, just before the transition from the S-shape to extension. In practice, a retraction of 50mm has been proposed as a reasonable standard reference point, and the resulting value is referred to as NIC50. The tolerance level, based on the pig experiments, is 15m/s^2. Attempts to validate the Neck Injury Criterion using volunteers in sled tests have shown that stiffer seat backs produce higher NIC values (Boström et al., 1997). Further tests (Boström et al., 1998), using a Hybrid III dummy (strangely, fitted with a standard neck) in three different car seats (one with a good track record from accident statistics, one with a bad record, and the third a newly developed seat - the Volvo Anti-Whiplash Seat) showed that NIC successfully identified the poor seat, whereas upper neck forces and moments were within suggested limits (57Nm/1100N - Backaitis & Mertz, 1994) for all seats. Muscle strength and reaction time were found to have a large influence on calculated NIC values in volunteer tests, and they conclude that this may explain the differences in risk between males and females.

4.3.2 Interaction Between Neck Motion and Head Restraint:

Other experimental studies include those of Weißner and Enßlen (1985), who measured head rotation on Hybrid II and III dummies in simulated rear impacts and found that a head restraint reduced the angular displacement of the head, hence concluding that head restraints were beneficial. Foret-Bruno et al (1991), having determined from accident studies that a modern, strong seat without a head restraint was associated with the highest risk of neck strain injury, placed a Hybrid III dummy in such a seat and measured head rotation along with a number of neck forces and accelerations, comparing these with similar measurements in other seat configurations. They found that collapse of the seat back reduced the head rotation and neck loads, as did addition of a head restraint. Eliminating elastic rebound from the seat without a head restraint gave results similar to the unmodified seat with a restraint. Von Koch et al (1995) also report work on seat rebound using dummies, and suggest that seats should be designed to undergo controlled plastic deformation in rear impacts, though rear seat passengers must also be taken into account here.

Viano and Gargan (1995) used observational data on the postures naturally adopted by car occupants relative to the seat and head restraint to define a number of test positions for dummies in sled tests. They measured the head rotation of a Hybrid III dummy in a test deemed to represent a “most favourable” condition - ie head restraint positioned high, with a small horizontal gap between head and restraint, and low impact speed (8 mile/hr). This test gave a head rotation of 14°, so angles measured in subsequent tests with different seat/restraint configurations were divided by 14 to give a “relative injury risk” for each configuration. However, this assumes that injury risk is proportional to head rotation alone, and also that the Hybrid III neck has good biofidelity.

Svensson & Lövsund (1992) on the other hand, point out that the neck of the Hybrid III dummy is known to show poor biofidelity (it is too rigid), and report tests on a new neck (the Rear Impact
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Dummy - RID - neck) developed in Sweden. Geigl et al (1995) used the RID neck in tests using dummies, volunteers and cadavers, and concluded that, although the RID neck is better, there is still room for improvement. A continuing problem is inability to model correctly the natural curvature of the entire human spine. Further development of the Chalmers RID neck has been carried out by TNO in the Netherlands (Thunissen et al, 1996) (see Section 4.5).

Ono & Kanno (1993) used volunteers to measure head motion in simulated rear impact (up to 4km/hr). Neck muscle tension was measured via an Electromyogram. Resultant neck forces were calculated from dynamic equations of motion. Impact speed, presence of a head restraint and its height adjustment influenced the results, as did seat inclination and pre-tensing of neck muscles. Eichberger et al (1996) also report volunteer tests, but conducted at between 8 and 11 km/hr, and using various production car seats to obtain a ranking of the seats in terms of a number of factors relating to geometry and padding stiffness etc, as well as the sled test results themselves (head rotations and accelerations). They found a wide variation between seats, to the extent that some seats were considered unsafe for testing by volunteers.

Finally, Guccione & Weiss (1995) have conducted volunteer rear impact tests, using three anthropometrically dissimilar volunteers, each subjected to the same impact acceleration. Measured head/neck responses were compared with the predictions of a simple, single fixed link model of the neck. Results indicate that the variations in head geometry and mechanical parameters (such as head centre of gravity, principle axes and moment of inertia) between the subjects are such as to lead to large numerical and even qualitative differences in head dynamics compared to those predicted by the model. The geometrical and mechanical parameters used as inputs to the model were taken from published datasets, based on cadaver studies, along with some very approximate figures based on live subjects (the technology for accurately measuring head mass distribution in live subjects - laser head scans and MRIs - is said to be still in the developmental stage). The problem is that these datasets were produced by different people at different times, they give different average values of the required parameters and they cannot be combined because error statistics are not available. Guccione & Weiss conclude that all current head/neck dynamic modelling is based on poor experimental data and is therefore unreliable.

4.4 ACCIDENT STUDIES

4.4.1 Serious Neck Injuries

Serious neck injuries include vertebral fractures and dislocations, with or without associated cord damage. They are generally agreed to be rare in road traffic accidents, especially since the introduction of seat belts has reduced the incidence of head contacts. Head contact during loading (when the head/neck complex is called upon to decelerate the torso) constrains the movement of the head and can prevent the neck from moving out of the load path. Nightingale et al (1991) have shown that this increases the risk of injury at all levels of the cervical spine in cadavers, and accident studies have shown that serious neck injury is associated with head/face injuries (McElhaney & Myers, 1993, Hassan et al, 1996). However, serious neck injuries can still occur in the absence of head contact, ie via a pure inertial loading mechanism.

Differences in fatality risk associated with injuries at different levels of the cervical spine have already been mentioned. Among accident victims who suffer a fatal neck injury, that injury is most likely to be at C2 and above, whereas surviving neck injured victims are more likely to be injured at C3 or below, and particularly at C5/C6 (McElhaney & Myers, 1993, Huelke et al, 1993, Yoganandan et al, 1995).

Ommaya & Digges (1985) conducted a prospective study on serious head/neck injury cases in the US. An accident investigation team was notified when a head/neck victim was admitted to hospital, and 84 complete cases (including vehicle-based details) were assembled from over 200 initial notifications. Accident and injury data were fed into a computer model, and forces, displacements, velocities and
accelerations of body components calculated. Biomechanical parameters could then be generated via the model, and compared with actual injury outcomes. However, the computer occupant model was based on the Hybrid III dummy, so biofidelity (especially with regard to the neck response) was a problem. A few preliminary case studies only are reported in this paper, and no papers have been found reporting any continuation of this work.

In a review of cases of non-head impact cervical spine injury to restrained car occupants in frontal crashes, Huelke et al. (1993) present data based on the US National Accident Sampling Survey (NASS), case studies found “+buried” (sic) in the literature, and case studies from their own investigations (both UK and US). They conclude that, due to the complexity of the various mechanisms of injury, and the wide variation in susceptibility to injury among the population (due to age, sex, arthritic state etc), it is most unlikely that any single parameter, such as shear, bending moment or torsional force can be specified as a tolerance limit. They also point out that current dummies do not adequately address the problem of non-head contact neck injuries in restrained occupants.

Hassan et al. (1996) report on a study of moderate to severe spinal injuries (AIS 2), based on the UK CCIS database. One of the criteria for inclusion in this study is that the vehicle must be towed away from the scene of the accident, so these are relatively severe accidents. Moderate to severe spinal injuries are said to be relatively rare, and to be associated with high impact severity. Over half the spinal injuries surveyed were cervical, and a significant number of those involved cord damage. The cervical spine was found to be particularly prone to deceleration injury. No correlation was found with impact type, seating position or belt use, though belt use was associated with less severe injuries. The incidence of cervical and lumbar spine injuries was higher in females, and spine injury was associated with head/face impact, especially roof contact, causing spine compression.

4.4.2 Whiplash Injuries

4.4.2.1 Using Pre-existing Data Sources:

Analysis of a large Swedish database is reported by Nygren et al. (1984). They found that head restraint effectiveness, in rear impacts only, was 25% (fixed restraint) and 15% (adjustable restraint), compared to no restraint. There was no significant difference for other impact directions. There was also no significant effect of either presence or type of head restraint on long-term disability (10% or more). The overall incidence of neck pain among non-fatally injured occupants of cars was 20%, and WAD inflicted in rear impact was more likely to lead to at least 10% impairment than WAD sustained in other impact types. In a further development of this work, Nygren et al. (1985) attempted to investigate the effects of, inter alia, car mass and head restraint adjustment on whiplash injury risk. Restraint adjustment information was not available in the database so they observed the adjustment of 187 cars at random from a traffic stream and found that 83% were in the lowest or second lowest position. They then sat a 50th percentile dummy in various models of cars with the restraint in this position and measured the vertical and horizontal distances from head to restraint. These were taken as the definitive restraint position parameters for each model of car. Analysis of the database showed that car mass was important, as was the vertical adjustment of the restraint, but horizontal adjustment was found to have no effect. However, after allowing for car mass, this method of analysis assumes that variations in whiplash incidence between car models was entirely due to restraint adjustment, and takes no account of structural properties of the car, which affect the crash pulse, (Nilsson et al., 1994, have shown this to be important), or the possibility that whiplash victims may represent a non-standard population in terms of stature, seat/restraint adjustment or posture.

Carlsson et al. (1985) analysing Volvo’s accident database, found a lower incidence of whiplash injury in rear seat occupants, and also an increased risk for taller occupants. In a long-term follow-up by mail questionnaire, they failed to find any link between long-term effects and any vehicle-related factors. Otremski et al. (1989), in a hospital-based study, also found that rear seat occupants are at
significantly lower risk of whiplash injury. This question was examined in detail by Lövsund et al (1988), who found that moving to the rear seat was nearly twice as effective as fitting a head restraint to the front seat in reducing whiplash injury risk in adults (ie approx 30% improvement by fitting a head restraint, compared with approx 50% reduction in risk in the rear seat).

Jakobsson et al (1994), also analysing the Volvo database, in conjunction with a questionnaire survey (questionnaires sent out 1-2 years post-accident) found that crash severity correlated to whiplash injury risk, as did greater horizontal distance to head restraint. However, it is unclear how reliable the restraint distance measurements would be from the memories of victims one to two years later.

Maag et al (1993) in a study based on insurance and police data for 1987 in Quebec (23,800 injured people, of whom 4,328 were car occupants with at least a neck injury), report that initial whiplash incidence (immediately post-accident) is 15-30%, but that these injuries are reported by 60% of RTA victims in the long term. Most neck strain injuries occur in 2-car crashes, without head contact, and the neck injury occurs most often in the lighter car, with women being more at risk than men. Neck injury accidents were initially classified by the police as damage only in 26% of cases, probably due to the delayed onset of symptoms in the victims - a recognised characteristic of WAD. They conclude that whiplash injury is a serious problem in terms of societal costs.

Morris and Thomas (1996), analysing the UK CCIS database, found an overall whiplash injury incidence of 16%, with females being at 50% greater risk than males. Seat belts increased the risk of injury for both sexes in frontal impacts and for males only in rear impacts. Neither seat back height nor occupant age, weight or height had any significant effect, though there was a non-significant trend for female neck injury victims in rear impacts to be associated with higher seat-back/head restraint measurements. There was no significant difference between fixed or adjustable restraints, and neither had any effect compared with “no restraint” cases, though non-significant trends suggested that restraints may be detrimental in frontal impacts. No significant difference was found between two- and four-door vehicles. There was a non-significant trend towards lower incidence of WAD in rear impacts where the seat back yielded.

Other analyses of the CCIS database are reported by Larder et al (1985) and Parkin et al (1995). The former also found no effect of head restraints, and no significant gender differences either, though the proportion of females was higher. Parkin et al found that collapse of the seat back in a rear impact generally had a beneficial effect on neck injury outcome and, using theoretical considerations, they point out that, depending on the crash pulse and the deformation characteristics of the seat back, the torso can be accelerated forward by the seat rebound to a velocity up to twice the ∆v imparted to the vehicle as a whole. In addition, this forward acceleration can be approaching its maximum before the head has made contact with the head restraint, thus exacerbating the extension of the neck.

In a study based on the activities of the accident research unit at Hannover University, Otte et al (1997) carried out a detailed investigation of whiplash cases in their database at low ∆v (<10kph). Head restraint position relative to the head of the victim was not directly available in the database, but the vertical positioning was deduced from the measured height of the restraint above the seat base plus an assumption that the seated height of the occupant is 53% of overall body height. On this basis, they found that WAD victims had better vertical head restraint adjustment than those with no whiplash injury. This leads to the conclusion that hyperextension is not a mechanism for whiplash injury, and they suggest that the most probable mechanism is shear in the first few milliseconds of motion.

The effect of vehicle mass has been investigated by von Koch et al (1995), using police and insurance data and by Eichberger et al (1996), using published German accident data. Both generated injury risk ratings for a variety of different car models, and found that there was a strong link between car mass and whiplash injury risk, with a five-fold difference in risk factor between the best (large mass) and the worst (low mass) cars. However, there were also large differences between cars of similar mass, which must be explained by differences in car structure and the seats fitted. Looking at the striking
car, von Koch et al found that there was again a strong link between whiplash risk and the mass of the striking car (as one would expect), but some cars were far more aggressive than their mass would suggest. Interestingly, some cars had poor ratings for both injury risk and aggressivity, eg Fiat Uno, Nissan Micra, while others were safe to ride in and non-aggressive too, eg Opel Omega, Mercedes 124, indicating that construction is also important.

Boström et al (1997), carrying this approach one step further, analysed 4,432 police reported accidents involving rear impact, where at least one driver had a minor injury. They claim that previous research has shown that nearly all these minor injuries will be AIS1 neck injury - ie whiplash. They generated injury risk ratings similar to those of von Koch et al and Eichberger et al. Because females are known to be at higher risk of neck injury, and also drive a different spectrum of car models, the relative injury risk was adjusted to allow for this, giving gender-compensated risk ratings for various car models. Having gone on to derive a relationship between relative injury risks and car mass ratios (striking car mass over total mass) they use this to compensate for car mass in the injury risk ratings, finally producing intrinsic risk factors for car models, independent of the spectrum of car masses they are likely to be struck by. The intrinsic relative risk was found to vary from 0.8 to 1.7, with Opel Vectra (89-94 models) at 2.6. Older models (80-89) gave lower average risk ratings than 1990s cars, and they speculate that the rising trend in whiplash incidence may be due to fitting stiffer seats in more modern cars. They state that the observed dependence of injury risk on mass ratio implies a strong dependence on $\Delta v$, but this seems to ignore the effect of the initial speed of the striking car, which also enters the momentum equation, and can have a large effect on the $\Delta v$ of the struck car.

Comparing their results with a report from the US Insurance Institute for Highway Safety (IIHS, 1995) which rated cars for whiplash injury risk according to seat/head restraint geometry, Boström et al point out that the observed relative risks from real accident data do not correlate with the IIHS’s assessment. They conclude that head restraint geometry is probably not important below a $\Delta v$ of about 20kph.

4.4.2.2 Field Studies

Only four prospective studies involving long-term medical follow-up coupled with detailed examination of the damaged vehicle are known. Two of these involved demonstration by the victims in their own (or similar) cars as to how they were sitting and how their restraints were adjusted at the time of the accident.

Olsson et al (1990) studied 33 occupants of Volvo cars who had been involved in rear-end collisions, and followed their progress for twelve months. Head position relative to the restraint was measured with the victims sitting in the vehicles, along with seat back angle before and after collision. Vehicle damage was also measured to give an estimate of impact severity. Medical assessment was by reference to a specified set of symptoms, and longevity of symptoms was used as a surrogate for injury severity. No correlation was found between impact speed and either the initial spectrum of symptoms or the duration of symptoms, though there was a non-significant correlation between duration of symptoms and the square of the maximum permanent deformation of the vehicle, in those cases where the stiff side members of the vehicle had been involved. No correlation was found with backrest deflection although, of the four uninjured people in the study, three had experienced some degree of backrest yielding in the impact. However, a statistically significant difference was found between duration of symptoms and whether the occupant’s head had been more or less than 10cm from the restraint - a distance greater than 10cm correlated with symptoms lasting at least a year, as opposed to less than a year. This result has been widely quoted in the literature as representing a higher risk of sustaining whiplash injury if the head is more than 10cm from the restraint. Strictly, however, this study did not examine incidence of whiplash injury, but only the severity of the injury. Eichberger et al (1996), in volunteer sled tests found that, of four tests conducted with the head 10cm or more from the restraint, three volunteers suffered slight neck injury, compared to one who suffered slight injury among a large number at closer than 10cm.
Minton et al (1998a) studied 174 whiplash-injured subjects selected from attendees at a hospital Accident & Emergency department. Injury severity was assessed by reference to a disability scoring system (Murray et al, 1993, 1994), and progress was followed for twelve months. The vehicles which the patients had been travelling in were examined and impact severity assessed. Patients were invited to attend the vehicle examination and to demonstrate the seating position adopted immediately prior to the impact. Measurements of head restraint positioning relative to the head were thus obtained for 103 of the subjects. Women were found to suffer significantly greater disability than men, despite ostensibly more favourable head restraint positioning but there was no correlation with occupant age, height or weight, or with impact severity. Larger horizontal distance from head to restraint was significantly associated with better long-term disability outcome. No clear dependance on vertical restraint position was found. A significant proportion of the sample had suffered lumbar strain injury in addition to whiplash, but segregation of the sample by lumbar injury status failed to give a clearer picture. Seat back angle had a significant effect on the lumbar injury cases, but was not important for non-lumbar cases. They conclude that the failure to find any clear benefit in “correct” adjustment of head restraints may have been due to the wide variety of makes and models of vehicles sampled. The possibility that, given an identical car in an identical accident, a “well-adjusted” head restraint would result in less severe injury than a badly adjusted restraint could not be excluded, but it was thought that any such effect was completely swamped by the effects of variations in restraint, seat and vehicle structural design between different models of vehicles. This is said to imply that changes in vehicle and seat design are likely to be more effective in preventing whiplash injury than simply exhorting occupants to adjust their head restraints “correctly”.

Ryan et al (1994) studied 32 patients who had suffered whiplash injury in traffic accidents around Adelaide, S Australia, and followed their progress for six months. Vehicles were examined to assess crash severity but no attempt was made to determine seat/head restraint parameters for the individual occupants. Rear impacts accounted for 22 of the accidents. Injuries were assessed by reference to a number of objective measurements of the range of head movement coupled with subjective ratings of severity from the victim and the medical examiner. Correlations were sought between each individual measurement/rating and the calculated impact speed, the vehicle deformation and whether or not the victim was aware of the impending impact. Significant correlations were found between some, but not all, of the severity indices on the one hand, and both velocity change and maximum vehicle deformation on the other. These correlations improved when rear impact only was considered. Victims who had been aware of the impending impact had significantly better outcomes for most of the severity indices than those who had been unaware. When the severity indices from the follow-up assessment were considered, no correlation with any vehicle-based factors could be found, but those who had been aware of the impact were found to have recovered much better than their unaware counterparts.

Fildes & Vulcan (1995) report on a study of 120 whiplash victims in Melbourne, Australia. The study involved detailed examination of vehicles and medical follow-up of patients. Head restraint positioning relative to the head was estimated from seat measurements and knowledge of occupant height, weight etc. Longevity of symptoms was used as measure of injury severity. Whiplash injury risk was found to be highest in rear impact, and rear and side impacts were more likely to result in chronic symptoms than frontal. Adjustable head restraints were worse than fixed restraints with respect to chronicity, although they point out that the adjustable restraints tended to be found in the more modern cars, so other factors, such as the trend towards stronger seat backs, could have had an effect. Females, and especially older females were more likely to have chronic symptoms. Greater horizontal distance from head to restraint was associated with higher probability of chronicity, but vertical positioning was not found to have any effect, a finding which they greet with some consternation.
4.5 SEAT DESIGN

4.5.1 Problems with Current Designs

The increasing incidence of whiplash injuries has been well-documented (Galasko et al, 1996, Morris and Thomas, 1996), and has occurred against a background of the increasingly prevalent fitment of head restraints in the front seats and, more recently, the rear seats of cars. These head restraints are fitted to prevent neck injury by limiting rearward hyperextension of the neck in a typical rear impact. Mertz and Patrick (1967) showed that eliminating head motion relative to the torso completely, by having a volunteer's head permanently in contact with a high, rigid seat back, allowed very severe rear impacts to be survived without ill effect. As discussed above (Section 3.3.), studies using dummies in simulated rear impacts have shown that head restraints can reduce head rotation (Weißner and Enßlen, 1985, Foret-Bruno et al, 1991, Viano & Gargan, 1995). Thus, publicity campaigns have been mounted, urging people to adjust their restraints to be as close to the head as possible horizontally, and to be about level with the ears, or the back of the head, vertically. The current WAD “epidemic” and the generally low estimates of head restraint effectiveness - about 20-30% (Nygren, 1984, van Kampen, 1993, Muser et al, 1994) are frequently blamed on the fact that very many people can be observed to ignore these recommendations. However, van Kampen (1993) compared the then current European head restraint height requirements with anthropometric data for the Dutch population, and found that a head restraint which just complied with the regulations would only be suitable for Dutch males below the 25th percentile. The regulations have since been amended, but the tendency of the torso to ramp up the seat back in rear impacts, or at least to straighten, pushing the head upwards relative to the restraint could still present problems (McConnell et al, 1993, Jakobsson et al, 1994).

But having one’s head close to a softly padded head restraint is not the same as being permanently in contact with a rigid structure. Even where the restraint is rigidly attached to the seat and made of fairly stiff material, head movement, particularly for drivers, is essential in modern traffic conditions, and this is incompatible with keeping the head permanently in contact with the restraint. Furthermore, as has been made clear in previous sections, current ideas on the mechanisms of whiplash injury are moving away from the simple hyperextension model. The injury is now thought to occur within the normal physiological range of overall head movement, although the angles between individual vertebrae may be outside their normal range. The injury may occur within the first few milliseconds after the onset of torso acceleration. Both Muser et al (1994) and Geigl et al (1994) claim that the neck itself, as well as the head and torso, must be supported in order to prevent whiplash injury.

Boström et al (1997) state that current ideas as to what constitutes a safe seat and a good head restraint, as propounded by, for example, the Insurance Institute for Highway Safety (IIHS, 1995) must be erroneous, since injury risk ratings based on real world accident data do not correspond with the IIHS ratings, which were based on a theoretical appraisal of seat geometry. They go on to say that, for current designs at least, head restraint geometry is probably irrelevant below about 20kph. However, Eichberger et al (1996) did find agreement between their theoretical seat assessments and real world data. Clearly, what constitutes a good seat design is still very much an open question.

Rebound from the seat after a rear impact is also now fairly well recognised as being harmful (Foret-Bruno et al, 1991, Svensson et al, 19935, von Koch et al, 1995, Krafft et al, 1996), and a consensus seems to be developing that the progressive strengthening of (front) seats over recent years (to provide increased protection in severe accidents) may have contributed to the rise in the incidence of whiplash injury by increasing the available elastic rebound energy. Differences between the seat back and the head restraint as regards their respective force/deflection characteristics have been highlighted by Spitzer et al, (1995) and Minton et al, (19985) as possibly exacerbating this rebound problem.

A paper investigating the apparent lack of whiplash injuries in Lithuania may lend support to the idea that modern seats may be contributing to the whiplash problem (Schrader et al, 1996). The thrust of the paper is to prove that most whiplash injury claims in Western countries are fraudulent, and
motivated by at least an expectation of disability, if not a desire for compensation. They conducted a carefully planned, blinded cohort study of Lithuanians who had been involved in rear impacts, compared to a random sample drawn from the population, and found that the incidence of neck pain was not significantly different between the two groups. However, a throw-away line in this report states that the vast majority of people in Lithuania are very poor, and drive very old cars (41% had no head restraint at all). Another possible confounding factor may be that poor health and safety practices in the former Soviet Union industries may have increased the incidence of neck and back problems in the general population.

Elimination of elastic rebound has been recommended by several authors (Foret-Bruno et al, 1991, Muser et al, 1994, von Koch et al, 1995, Parkin et al, 1995), by introducing plastic deformation elements into the seat back; Minton et al (1998b) suggest that the same effect could be achieved by firing the seat belt pretensioners in rear impacts. Geigl et al, (1994), however, point out that any backward rotation of the seat back about its pivot effectively moves the head restraint away from the head in the early moments of impact, even if rebound is eliminated.

The challenge for seat designers thus appears to be to eliminate all relative motion between head and torso, while still allowing drivers freedom of head movement during normal driving, and also to eliminate elastic rebound from the seat. Svensson et al (1993c), in tests on a modified production car seat using a Hybrid III dummy with a RID neck showed that it is possible to eliminate virtually all neck extension motion. This was done by adding diagonal struts from the seat base to the seat back (eliminating elastic excursion of the seat back), stiffening the lower seat back (to prevent rearward excursion of the pelvis, which increases the relative angle between the torso and the head), adding extra padding to the upper seat back (giving the upper torso more space to move back before being stopped by the rigid seat back frame), reducing the initial horizontal gap between head and restraint, and arranging a flat vertical front face for the head restraint.

4.5.2 Novel Seat Designs

4.5.2.1 The Saab Seat

This is a fairly simple idea. The head restraint is attached to a mechanical lever system embedded in the seat back. In a rear impact, inertial loading of the seat back by the torso activates the lever and pushes the head restraint closer to the head, automatically reducing the head to restraint gap before the displacement of the head relative to the torso becomes too great.

4.5.2.2 The ETH Seat

As part of a programme aimed at the development of a very small vehicle, the Swiss Federal Institute of Technology (ETH) has been studying the special requirements placed upon restraint systems designed for use in these low mass vehicles. Muser et al (1994) presented a prototype automatically positioned head restraint, which is capable of supporting the head and neck, and which uses capacitive proximity sensors to adjust continuously (in the fore-and-aft direction) to the posture adopted by the occupant. The head restraint itself takes the form of a hollow box, suitably padded on the outside and having hinged joints, such that, when one diagonal is shortened by motorised cables, its front face moves forwards, by up to 100mm, to follow the head of the occupant (see Figure 4.2).
Dippel et al (1997) present the latest development of this work, which includes a complete prototype seat incorporating the “roving” head restraint. There is no facility for vertical adjustment of the head restraint, but the entire seat geometry, including seat back height and seat base length and tilt angle can be adjusted to suit occupant sizes ranging from 5th percentile female to 95th percentile male. The seat back padding thicknesses and stiffnesses have been carefully chosen to allow the torso to sink relatively unimpeded into the seat back in the early stages of a rear impact, until the (already small) gap between head and restraint has been taken up, before stiffer padding begins to accelerate the torso and head in unison. Energy absorbing rotational yielding elements in the seat back further aid in reducing the violence of the impact and in eliminating seat back rebound. A series of sled tests using a Hybrid III dummy equipped with a TRID neck have been conducted, and show that the system virtually eliminates relative movement between the head and neck during impact. A possible further improvement to the seat may be to absorb energy through translation of the seat back, rather than rotation, but they state that this may be more difficult to implement in practice, and could cause problems for rear seat passengers.

4.5.2.3 The Head Restraint Airbag

Bigi et al (1998) describe collaborative work carried out by TRW in Germany and Graz University, Austria. The aim of the work was to develop a head restraint which senses an impact and “deploys”, by becoming larger, to take up the gap between the restraint and the occupant’s head. Two mechanical prototypes, which deploy by means of a system of internal springs and levers, are cursorily described. The favoured design, however, incorporates a small airbag inside the head restraint, beneath the foam padding, which enlarges the restraint considerably when it deploys (see Figure 4.3).

Sled tests using a Hybrid III dummy with a TRID neck indicated that the new design reduced NIC values compared to a standard head restraint. Even when the head was initially in contact with the restraint, volunteer tests indicated no discomfort from the deploying restraint, partly because deflection of the seat back moved the restraint away from the head in the first few moments of impact (the volunteers were not instrumented, but merely asked for their subjective opinions in impacts up to 10kph). The airbag restraint was found to be much less aggressive than the mechanically deploying types. Noise measurements during deployment indicated sound pressure levels of about 148dB, which are claimed not to be dangerous.

However, the UK Noise at Work Regulations state that no workers should be exposed to noise levels above 140dB, no matter how short the period. It then becomes a value judgement as to whether a one-off exposure to an impulsive level of 148dB is acceptable, bearing in mind the possible safety benefits, although test volunteers may be unhappy with such a situation.

4.5.2.4 The Toyota Seat

Sekizuka (1998) reports on the development of a new seat by Toyota. The new seat is designed so that the upper torso moves back unimpeded into the seat cushion until the head is in contact with the head restraint, thus reducing relative horizontal motion between head and torso. The lower seat back is stiffened to reduce rearward pelvic excursion, and to induce gentle extension of the whole spine. In the rebound phase, the lower torso rebounds first, again reducing the relative extension of the neck. However, it is not clear how this differential motion between upper and lower torso will affect the risk
of lower back strains, even if it successfully reduces the risk of whiplash injuries. Sled tests were carried out using a Hybrid III dummy with a modified neck (segments cut out to make it less stiff) and also with volunteers using the same X-ray cinematographic technique as Ono and Kaneoka (1997) (see Section 4.2.2.2.). Results are presented for the modified dummy in the new seat at two different impact speeds, but there are no comparison tests using a standard seat. Instead, the dummy neck extension angles are compared to relative extension angle limits proposed by Eichberger et al (1996), based on volunteer tests. The dummy extension angles were within Eichberger’s limits, but it is felt by the present author that some control tests with a standard seat might have been more convincing as regards the relative benefits of the new seat. In the volunteer tests, the new seat was found to produce individual vertebral motions within their normal ranges at a ∆v of 8kph. The new seat is fitted to Toyota’s new Prius model.

4.5.2.5 The Volvo WHIPS Seat

Lundell et al (1998a and 1998b) describe the development of the Volvo WHIPS (WHIplash Protection Study) seat. It was recognised at an early stage that it was important to consider the whole spine, not just the head/neck/upper thorax. The guidelines laid down for the design of the seat were influenced by the fact that the exact mechanism of injury is unknown, so all realistic theories about the mechanism had to be taken into account. Three primary guidelines were established:

Reduce occupant acceleration
Minimise relative movements between adjacent vertebrae and in the occipital joint, ie the curvature of the spine shall change as little as possible during the impact.
Minimise the forward rebound into the seat belt.

The first guideline is based on the common-sense notion that zero acceleration would produce zero injuries. Also, the proposed Neck Injury Criterion is based on acceleration. Compliance with this guideline is claimed to be capable of assessment using standard dummies, since overall acceleration is independent of a biofidelic neck response.

The second guideline addresses most of the currently proposed mechanisms of whiplash injury, and would also address the old hyperextension mechanism. There are no dummies in existence today which could be used to verify compliance with this requirement. The problem was therefore approached using mathematical modelling, geometrical considerations and engineering judgement. In particular, the backrest and head restraint were designed to follow the shape of the spine as closely as possible, and local hard or soft structures which could force the spine into localised bending were avoided. The force/deflection characteristics of the backrest and head restraint were to be as uniform as possible throughout their height.

The third guideline implies good energy absorption in the backrest. Again, compliance is said to be capable of being tested with standard dummies, since it is overall torso rebound which is to be minimised.

No biomechanical thresholds were defined for any of these requirements, since none exist. The goal was to achieve the largest possible reductions for all three kinematic parameters in unison, and never to reduce one at the expense of increasing another.

The result is a seat which, in a rear impact, first allows the occupant’s torso to sink into the padding, reducing the distance between head and restraint without applying too much acceleration force to the torso. When the inertial force on the backrest reaches a certain level, the special WHIPS recliner mechanism is activated, and the backrest translates backwards. In higher severity impacts, the second function of the WHIPS recliner is activated, with a special element in the recliner becoming deformed, as the backrest rotates backwards by up to 15°, to further reduce occupant accelerations. At the end of this reclining phase, the backrest will rebound, but to a much smaller extent than a standard backrest.
The results of sled tests are presented, which show that lower neck horizontal accelerations are significantly reduced in the WHIPS seat compared to a standard production seat. Rebound is also said to be much reduced.

4.6 SUMMARY OF KEY POINTS

1. **Severe neck injuries are rare**, but are very dangerous in terms of the risk of death and permanent paralysis.

2. **Whiplash injuries are widespread** and socially expensive, despite their low rating on a threat-to-life basis.

3. **Mechanisms of whiplash injury:**

<table>
<thead>
<tr>
<th>Model</th>
<th>Mechanism</th>
<th>Important Parameters</th>
<th>Location and Type of Damage</th>
</tr>
</thead>
<tbody>
<tr>
<td>Classical</td>
<td>Hyperextension</td>
<td>Head Rotation, Upper Neck Forces/moments</td>
<td>Upper Neck, Ligaments/muscles</td>
</tr>
<tr>
<td>Aldman</td>
<td>Shear, Transient Pressure Surges in Cerebro-spinal Fluid</td>
<td>Acceleration, Velocity</td>
<td>Lower Neck, Spinal Nerve Roots</td>
</tr>
<tr>
<td>Penning</td>
<td>Shear</td>
<td>Displacement</td>
<td>Upper Neck, Ligaments/muscles</td>
</tr>
<tr>
<td>Von Koch</td>
<td>Frontal Mechanism</td>
<td></td>
<td>Not Further Specified</td>
</tr>
<tr>
<td>McConnell</td>
<td>Shock Loading, Spine Compression</td>
<td>Acceleration, Velocity</td>
<td>Muscles in Anterior Part of Neck, Inter-vertebral Discs</td>
</tr>
<tr>
<td>Ono &amp; Kaneoka</td>
<td>Shear</td>
<td>Rotation of Individual Vertebrae</td>
<td>Lower Neck (C5/C6), Intervertebral Ligaments/muscles</td>
</tr>
<tr>
<td>Panjabi</td>
<td>Shear</td>
<td></td>
<td>As Ono &amp; Kaneoka</td>
</tr>
<tr>
<td>Barnsley</td>
<td>Not Specified</td>
<td></td>
<td>Intervertebral Joints</td>
</tr>
</tbody>
</table>
4. **Tolerance Limits:**
These are currently the subject of international research. They are very difficult to define, because of the wide variations in:
(i) Injury susceptibility in the population (due to age, sex, muscle strength, arthritic state).
(ii) Detailed neck kinematics (due to degree of muscle tensing and strong dependence on initial posture).
(iii) Possible modes of injury (due to the complexity of the neck structure and the omnidirectionality of natural movement)
However, some tolerance limits which have been proposed are detailed below:

Mertz & Patrick:  
- 47.3Nm at Occipital Condyles (Non-injury)
- 56.7Nm at Occipital Condyles (Ligamentous injury)

Backaitis & Mertz:  
- 57Nm/1100N at Occipital Condyles

Yoganandan:  
- 0.35MPa (Pressure in Cerebro-spinal Fluid)

NIC:  
\[ \text{NIC} = 0.2\theta_{rel} + v_{rel}^2 \]

5. **Biofidelity:**
The standard Hybrid III neck is generally agreed to be lacking in biofidelity, particularly with regard to head/neck kinematics in low-speed rear impacts, where accident studies have shown that some very debilitating and socially costly, albeit not life-threatening, injuries are common.

The Chalmers RID neck, and its offspring, the TNO TRID neck represent attempts to produce a more biofidelic neck response, particularly in these low-speed rear impacts. They succeed in more faithfully modelling the shearing motion and the S-shaped distortion of the neck observed in volunteer tests. However, lack of both a clearly defined mechanism for whiplash injury and of tolerance limits for soft tissue neck strain still hampers their applicability.

The RID and TRID necks only operate in two dimensions. A new neck, which has been developed for the NHTSA advanced frontal impact dummy, attempts to give a biofidelic response in the lateral direction also. A new neck model developed in Japan, which is designed to resemble very closely the physical structure of the human neck may also give a good response in the lateral direction.

No matter how biofidelic a dummy neck is, its performance is likely to be compromised unless the biofidelity of the rest of the dummy spine is improved.

6. **Seat Design:**
Current thinking on seat design for reduction of whiplash injuries can be summarised as:
(i) Eliminate all relative displacements and accelerations between head and torso
(ii) Reduce rearward excursion of the pelvis
(iii) Eliminate post-impact rebound from the seat back.

Anti-whiplash seat designs which are either at the prototype or production stage can be categorised as follows:
- Roving Restraint: Saab, ETH
- Moving Backrest: ETH, Volvo
- Airbag Restraint: TRW/Graz University

**Post-script:**
Since this chapter was written, Davidsson *et al* (1998) have reported on the extension of the RID concept to the entire spinal column, producing the BIO-RID I - a dummy for use in low speed rear impact testing and seat design. Validation tests are said to have shown that it has a much more biofidelic response than the Hybrid III for the whole torso in these types of tests.
4.7 REFERENCES


Culver CC, Neathery RF & Mertz HJ (1972): Mechanical Necks with Human-like Responses, 16th STAPP Car Crash Conf. SAE 720959.


Dummy Development: Spinal Injuries


Dummy Development: Spinal Injuries


Dummy Development: Spinal Injuries


5 THORACOLUMBAR SPINE INJURIES

5.1 INTRODUCTION

Injuries to the thoracic and lumbar spine are less common than cervical spine injuries in automotive accidents. A recent study by Hassan et al (1996) on a sample of 290 automobile casualties who suffered spinal injury (MAIS 2) found that over half (57%) had an injury to the cervical spine. The incidence of injury to the thoracic and lumbar spine was 21% and 22% respectively. Perusal of the literature reveals that most of the research effort on the biomechanics of spinal injury has concentrated on the neck region while much less has been published about injuries lower down the spine.

Information on the tolerance of the human lumbar and thoracic spine to acceleration is also sparse. Such tolerance is dependent upon a number of factors including the age of the occupant. The type of restraint system used will affect the nature and location of the damage to the spine in an impact. In the absence of a restraint system, more spinal injuries are likely to occur, especially for occupants of vehicles involved in rollovers. It is extremely difficult to arrive at a limited set of injury criteria for the spine because the failure of the spinal components is not restricted to the bony parts of the spine. Much of the spinal resistance to bending and torsional loads is provided by soft tissues - ligaments, muscles, and cartilage (disc).

The mechanisms of severe injury to the thoracolumbar spine are relatively well understood. Thoracic and lumbar spinal injury is predominantly fracture without associated cord damage often caused by contact loading of the spine (Hassan et al, 1996). This differs from the cervical spine which can incur injury purely by deceleration effects. The thoracic spine is especially susceptible to injury from impacts to the side of the occupant. Lumbar injury occurs frequently when the seat is loaded by passengers or luggage. It also occurs in severe frontal impacts in which the vehicle occupants are restrained by lap belts. All three regions of the spine are vulnerable to injury when compressive loads are transmitted via a head contact.

Failure of the various spinal components can be attributed to a combination of axial and/or bending loads. The articular facets play a vitally important role because in conjunction with the vertebrae, they provide a dual load path for the transmission of axial load.

Several medical researchers have classified the various types of thoracolumbar injury into different categories. Holdsworth (1963, 1970) classified spinal injuries by treating the spine as two columns - the anterior column and posterior column. He identified five types of injury; flexion, flexion/rotation, extension, compression and shear. These classifications were based on observed fractures of the vertebrae. It should be noted that fractures are not always related to neurological impairment.

Denis (1983, 1988) considered the spine as three columns and classified injuries into major and minor categories. The three columns he defined were:-

Anterior column: anterior halves of disc and vertebral body plus anterior longitudinal ligament
Middle column: posterior halves of disc and vertebral body plus posterior longitudinal ligament
Posterior column: everything posterior to the posterior longitudinal ligament

Denis suggested that if 2 out of 3 columns were involved in an injury then the spine becomes unstable. He proposed 4 general categories of injury which again relate only to the bony structures:-

1) Compression fracture
2) Burst fracture
3) Fracture with rotary dislocation (likely to cause spinal cord injury)
4) Flexion/distraction - also known as a ‘Chance’ fracture

More comprehensive classifications of thoracic and lumbar injuries which incorporate 6 or more separate categories have been put forward by White & Panjabi (1978), King (1993) and Magerl et al (1994). The most common types of thoracolumbar injuries are discussed in section 5.2 below.

The biomechanical mechanisms which cause less severe (AIS = 1) soft tissue injury in the thoracolumbar spine are less well understood. Minton et al (1998) discovered that there may be a possible link between whiplash injury in the neck and lumbar spine injury. This has been confirmed anecdotally by Bunketorp (1998).

5.2 INJURY MECHANISMS

5.2.1 Injury Categories
Injuries to the vertebral column can be broadly classified into seven different categories as follows:-

1. Anterior wedge fractures of vertebral bodies
2. Burst fractures of vertebral bodies
3. Dislocations and fracture dislocations
4. Rotational injuries
5. Chance fractures
6. Hyperextension injuries
7. Soft tissue injuries

5.2.2 Anterior Wedge Fractures

These injuries occur at all levels of the spine and are common in both aircraft and automotive accidents. The injury mechanism is combined flexion and axial compression. This type of injury is illustrated in figure 5.1. The loss of anterior vertebral height results in angulation of more than 5 degrees. The posterior wall of the vertebral body remains intact. The loss of height may occur in the upper part of the vertebral body (superior wedge fracture) as shown or in the inferior part of the vertebral body (inferior wedge fracture).

It is a mild form of spinal injury which has been associated with pilot ejection. The region between T10 and L2 is most susceptible to anterior wedge fractures although they can occur in the T4-T6 segment (Kazarian, 1982). Begeman et al (1973) have demonstrated that people restrained by a lap belt and an upper torso belt can develop spinal loads high enough to cause wedge fractures when they are subjected to large forward accelerations.

Figure 5.1 Anterior wedge fracture
5.2.3 Burst Fractures

These injuries occur at higher levels of input acceleration or load when applied more directly over the vertebral body causing it to break up into two or more segments (figure 5.2). The cord may be damaged by the movement of the fracture fragments posteriorly into the spinal canal. The cord can also be injured by the retropulsion of the disc into the canal. In many cases of paralysis, post impact x-rays show a burst fracture with fragments which do not intrude into the spinal canal. This does not mean that the cord was not damaged because often the fragments are pulled back into the vertebral body by the posterior longitudinal ligament leaving no apparent intrusion into the canal. Conversely, there may be proof of gross intrusion into the spinal canal but no neurological damage. Thus, with this type of injury, consideration of the kinetic energy of the impact may give a better indication of the extent of the injury than evidence from x-ray images.

5.2.4 Dislocations and Fracture-Dislocations

These are very important injury mechanisms associated with high forces. They are generally flexion injuries accompanied by rotation and posteroanterior shear. Unilateral dislocations require an axial rotational component, while bilateral dislocations can be caused solely by flexion. The principal difference between a simple wedge fracture and a fracture-dislocation is, according to Nicoll (1949), the rupture of the interspinous ligament. Examples of three types of fracture-dislocation are illustrated in figures 5.3 to 5.5 below.

Figure 5.2. Sketches of a burst fracture taken from X-ray pictures. A - anterior view, B - posterior view and C - lateral view. D and E show the upper and lower part of the vertebral body respectively. (Reprinted with permission from European Spine Journal 1994. A comprehensive classification of thoracic and lumbar injuries by Magerl, v3 1994. © Springer Verlag)
There are varying degrees of dislocation. The inferior facets can be moved upward relative to the superior facets of the vertebra below or the facets can be perched on top of each other. Partial dislocation in which the bone ends become misaligned but remain in contact is known as subluxion. It is an unstable condition which endangers the integrity of the cord.

Forward dislocation is also possible with fracture of the facets or the neural arch as well as forward dislocation with locking of the facets. That is, the inferior facets move up and over the superior facets of the vertebra below and come back down so that they become anterior to the superior facets. Neurological damage is highly likely with this type of injury because the cord is subjected to high shearing and stretching forces. If there is dislocation without wedging, the mechanism of injury is a high shear load in the posteroanterior direction (Kazarian, 1982).
Fracture-dislocations are common in the thoracolumbar region especially around L1/T12 where a change in stiffness and range of motion of the vertebral motion segments occurs (Freeman, 1998).

5.2.5 Rotational Injuries

If the spine is twisted about its longitudinal axis and is subjected to axial and/or shearing loads, lateral wedge fractures may occur (Nicoll, 1949). An example of this type of injury is illustrated in figure 5.6. Other forms of injury include uniform compression of the vertebral body and fracture of the articular facets and lamina. Kazarian (1982) noted that lateral wedge fractures occur mainly in two spinal regions; T2 to T6 and T7 to T10. The damage to the posterior intervertebral joint is on the concave side and is often accompanied by fracture of the transverse process on the convex side. Indeed fracture of one or more transverse processes is often an indication that rotation played a significant part in a spinal injury. Unlike the anterior wedge fracture, this type of injury can result in neurological deficit, including paraplegia.


5.2.6 Chance Fractures

This type of injury was first described by Chance (1948) as being a lap belt related syndrome in which a lumbar vertebra is split in the transverse plane beginning with the spinous process (see figure 5.7). This injury results from a frontal impact in which the lap belt rides over the iliac wings and acts as a fulcrum for the lumbar spine to hyperflex over it thereby causing a marked separation of the posterior elements of the spine. Chance fractures are often accompanied by intra-abdominal injury and ligamentous damage - particularly tearing of the inter-spinous ligament. Denis (1983) reports that the classical Chance fracture (figure 5.7) is just one of a group of seat-belt type injuries. Occasionally, hyperflexion may cause rupture of the posterior ligaments and damage to the annulus fibrosis of the discs without fracture of the bony structure as in the example of figure 5.8. Very often the resulting injury is a combination of those shown in figures 5.7 and 5.8. Shoulder straps reduce the likelihood of this type of injury and since the introduction of three point seat belts it has become quite rare in motor vehicle accidents.
5.2.7 Hyperextension Injuries

Hyperextension injuries of the thoracic spine are very rare in accidents. When the spine is violently hyperextended, the superior outer surface of one or more vertebrae can be avulsed (forcibly separated) along with rupture of the anterior longitudinal ligament (see figure 5.9). This sometimes results in the loss of posterior vertebral body height. When it does, there may be damage to the articular facets, pedicles and/or laminae. Such injuries may be possible in accidents involving vehicles fitted with active restraint systems if large forces are developed on the front of the torso.

Figure 5.7. Lateral diagram of the ‘Chance’ fracture caused by hyperextension. (Reproduced with permission from Denis F (1983). The three column spine and its significance in the classification of acute thoracolumbar spinal injuries. Spine Vol 8 No.8 p817-830.)

Figure 5.8. A variation of the ‘Chance’ fracture in which only the posterior ligaments and disc are ruptured. (Reproduced with permission from Denis F (1983). The three column spine and its significance in the classification of acute thoracolumbar spinal injuries. Spine Vol 8 No.8 p817-830.)

Figure 5.9. Severe hyperextension injury. Note the rupture of the anterior longitudinal ligament. (Reprinted with permission from European Spine Journal 1994. A comprehensive classification of thoracic and lumbar injuries by Magerl, v3 1994. © Springer Verlag)
5.2.8 Soft Tissue Injuries

Low severity soft tissue injuries (AIS = 1) are difficult to define and diagnose because very often the tissue damage cannot be seen on x-ray or MRI images. The soft tissues involved are the intervertebral discs, the various ligaments around the intervertebral joint, the facet joints and their capsules, and the various muscles and tendons attached to the vertebral column. The usual complaint of this type of injury is lower back pain which is frequently accompanied by radiating pain down the buttocks and the lower extremities.

The incident provoking this complaint can be anything from a low speed rear impact, to a vehicle going over a pothole to a severe frontal collision between two cars. If post-impact x-rays reveal no damage to the spine then a diagnosis of lumbar sprain or strain is usually made and the patient is sent home with no more than pain killers. Often, though, the pain persists and eventually a diagnosis of disc rupture, disc bulge or other specific condition is diagnosed and ascribed to the earlier accident. This cause and effect relationship is nearly always based on the history given by the patient and not on the severity of the impact or the biomechanics of the loading of the spine.

Soft tissue injuries to the intervertebral discs occur far more frequently in automobile accidents than injuries to the bony parts of the thoracolumbar spine. However, because lower back pain is a very common complaint, inasmuch as 8 out of 10 people will suffer at least one attack during their lifetime, it is extremely difficult to relate a particular attack to a specific accident. Lower back pain is often termed ‘idiopathic’ meaning that it has no known cause. Although research has revealed a few of the causes of lower back pain, the subject is still not fully understood.

Disc ruptures do not occur as the result of single impact loading events on the spine, unless there are associated massive bony injuries caused by a very severe accident. The failure load for a vertebral body is in the order of 4000N to 7000N depending on the person’s age whilst the failure load for a disc is an order of magnitude greater than these values (Freeman, 1998). Thus, the vertebral bodies will always break before the adjacent discs incur any visible damage (Henzel et al, 1968). Disc rupture is more often than not a result of slow degenerative processes which can take a long time to develop (Galasko, 1998).

5.3 DISCUSSION

The consequences of the various injuries described above depend on the degree of damage to the spine and spinal cord. The outcome for the victim can range from irritating pain, permanent deformity, loss of mechanical performance through to complete paraplegia. Usually, the severity of the injury and its aftermath is directly related to the amount of energy imparted to the spine during the accident which caused it.

All doctors who treat spinal injuries use the term stability or instability of the spine when assessing the seriousness of a particular injury. Whether or not the spine is stable or unstable following an accident determines the subsequent course of treatment and aftercare. There exists some variation in the meaning and definition of spinal stability within the medical profession. However, White and Panjabi (1978) have defined clinical instability as ‘the loss of the ability of the spine under normal loading conditions to maintain relationships between vertebrae in such a way that there is neither damage nor subsequent irritation to the spinal cord or nerve roots. Furthermore, there is no development of incapacitating deformity or pain due to structural changes’. Involvement of the spinal cord in an injury can lead to neurological deficit which ranges from the loss of a single nerve root to complete paraplegia.

Low severity soft tissue injuries (AIS = 1) in the thoracolumbar region were discussed in the preceding section 5.2.8. The main consequence of these types of injury is lower back pain which may radiate along the legs. In the worst cases, in which symptoms persist, there may be some change in
behavioural function of the lower body similar to that produced by ‘whiplash’ injury (chapter 4). The spinal fracture and dislocation type injuries described in sections 5.2.2 to 5.2.7 are more serious. Magerl et al (1994) conducted a comprehensive survey of 1445 such injuries and found that 59% of them occurred at the thoracolumbar junction between T12 and L2. The upper (T1-T4) and lower (L5) end of the thoracolumbar spine and the T10 level were the most infrequently injured. Although the spinal injuries in the Magerl survey resulted from all types of human accidents, it is likely that a similar pattern of injury distribution would have emerged had only motor vehicle accidents been considered.

Wedge fractures do not usually threaten the integrity of the spinal cord. Magerl et al (1994) found that of all the patients who presented with an anterior wedge fracture only 2% suffered any neurological deficit. This type of fracture is often associated with severe pain and slow healing and treatment depends on the amount of damage to the vertebrae involved. Patients with only mild compression loss of less than one third of the original height of one or more vertebrae may be treated with active exercise and mobilization after an appropriate period of bed rest. Patients who suffer more than this degree of compression often develop an unsightly sharply angled curvature to their backbone (gibbus) which has to be corrected by clinical or surgical techniques.

Although lateral wedge fractures are similar in appearance to anterior wedge fractures, Nicoll (1949) reported a 40% complete recovery in patients having the former type of fracture compared with only 21% for the latter. The residual pain associated with lateral wedge fractures was also greater. The reasons for both of these facts are believed to be related to the relative amount of soft tissue damage in the two types of fracture.

As was indicated in section 5.2.3 burst fractures can lead to spinal cord damage and even paralysis if bony fragments are retropulsed into the spinal canal. In a sample of 344 patients who presented with this type of injury, Magerl et al (1994) reported neurological deficit in 32% of them. Unstable burst fractures are accompanied by significant pain and reduction of the patient’s mobility. Loose fragments of bone within the spinal canal pose a permanent threat to the integrity of the cord.
Fracture-dislocation injuries, described in section 5.2.4, can be the most serious and unstable of all injuries to the spine. Riggins et al (1977) found a 61% risk of neurological deficit when dislocation is accompanied by fracture of a vertebral body and its posterior elements and a 56% risk when accompanied by vertebral body fracture alone. These figures agree with the work of Magerl et al (1994) who found neurological deficit in 60% of patients who presented with fracture dislocations caused by rotation. Denis (1983) in a study of 412 patients having thoracolumbar injuries found seven who had fracture-dislocations caused by shear as depicted in figure 6.4. All seven cases were complete paraplegics on admission to hospital. Treatment of these injuries may be long and involved and will vary for each individual case. Unstable fracture-dislocations often require fusion of the vertebral column.

Rotational injuries, discussed in section 5.2.5 are amongst the most severe which can occur within the thoracic and lumbar spine. Magerl et al (1994) report a 53% risk of neurological deficit with this type of injury. Neural injury is caused by fragments displaced into the spinal canal and/or by narrowing of the canal by the translational motion of the vertebral bodies. Treatment is often long term and may involve the fusion of vertebral column and/or the surgical fitting of an orthosis. (An orthosis is a support appliance which exerts corrective forces on the injured spine.)

‘Chance’ fractures look to be more serious than they often actually are (see figures 6.7 and 6.8). In the study by Denis (1983), 19 of the 412 thoracolumbar injury cases had ‘Chance’ or similar hyperextension fractures. However, no neurological injury could be linked with this type of injury in any of the 19 cases. Treatment depends on the clinical stability and for the more severe injuries will involve internal fixation and fusion of the vertebral column in the damaged area of the spine.

Hyperextension injuries to the thoracolumbar spine are very rare in motor vehicle accidents and there is little commentary on them in the literature. Neurological deficit will usually only occur when the hyperextension is accompanied by shear or rotation. Treatment of hyperextension injuries is similar to that for the management of compression injuries such as anterior wedge fractures and burst fractures.

5.4 SUMMARY

A perusal of the literature on injuries to the thoracolumbar spine has revealed the following important points:

a) Spinal anatomy is extremely complex. A thorough knowledge of its construction and function is essential to the understanding of the biomechanics of spinal injury.

b) Substantially more research effort has been concentrated on injuries to the cervical spine than on injuries to the thoracolumbar spine.

c) The articular facets play a central role in the mechanism of spinal support and in the mechanisms of spinal injury. The orientation of the facet surfaces may account for the tendency of subluxion and dislocation of certain vertebrae. (Note: Subluxation is a partial dislocation in which the bone ends become misaligned but remain in contact. It is an unstable condition which endangers the integrity of the cord).

d) The facets are able to resist compressive loads but offer virtually no resistance to tensile loads. It is believed that the ligaments and muscles along the posterior spine perform this role.

e) Impact loads are unlikely to produce ruptures of the lumbar intervertebral disc. Injuries to intervertebral soft tissues, though common, are difficult to diagnose. It has been postulated that damage to the facet capsule is a source of lower back pain.
f) There is a shortage of tolerance data for the human thoracolumbar spine and further research in this area is required to investigate the problem of injury tolerance and injury criteria. In particular, there would appear to be a need to study the tolerance of the facets to dislocation which is a high risk injury.

g) Mathematical models are very useful tools for studying the way in which the spine responds to impact and offer a flexible and low-cost alternative to experimental research. A combined approach of mathematical modelling and complementary experimental testing is arguably the best way of investigating the biomechanics of spinal injury.

5.5 REFERENCES


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Galasko C S B (1998): Private communication in meeting with TRL at Hope Hospital, Manchester.


6  ACCIDENT ANALYSIS

6.1  INTRODUCTION

A brief analysis of the CCIS databases has been undertaken, to obtain a preliminary understanding of the factors influencing spinal injury. The databases relating to Phases 4 and 5 of the CCIS project were examined. These contain details of 5,700 accidents, involving 11,837 occupants. The sampling methodology is biased towards more severe crashes, so any injury rates quoted in the following analysis will be considerably higher than those applicable to the general motoring population in the UK.

A subset of occupants who received an injury (of any severity) to the spine (AIS Body Region 6) was identified, and these 3,238 occupants formed the basis of this analysis. These occupants received a total of 3,726 spinal injuries, of which 419 were at AIS 2 or above, affecting 314 occupants. One hundred and six of the occupants died (though not necessarily from the spinal injury). The rate of AIS 1 spinal injury, at 25% of all occupant injuries, is quite high, particularly bearing in mind that other studies have shown that the incidence of AIS 1 cervical spine injuries, in particular, does not fall off to any great extent as accident severity reduces (see Chapter 4). In contrast, the rate of AIS 2+ spinal injury is fairly low (2.7%), considering the high proportion of relatively severe crashes in the sample. Table 6.1 shows the gender distribution in the whole dataset and in the spinal injury subset:

Table 6.1. Occupant Gender Distribution and Injury Rates:

<table>
<thead>
<tr>
<th></th>
<th>All Occupants</th>
<th>With Spine Injury (rate %)</th>
<th>With AIS 1 Spine Inj. (rate %)</th>
<th>With AIS 2+ Spine Inj. (rate %)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>6,787</td>
<td>1,661 (24.5)</td>
<td>1,495 (22.0)</td>
<td>182 (2.7)</td>
</tr>
<tr>
<td>Female</td>
<td>4,458</td>
<td>1,568 (35.2)</td>
<td>1,450 (32.5)</td>
<td>132 (3.0)</td>
</tr>
<tr>
<td>N/K</td>
<td>592</td>
<td>9</td>
<td>9</td>
<td>0</td>
</tr>
<tr>
<td>Total</td>
<td>11,837</td>
<td>3,238 (27.4)</td>
<td>2,954 (25.0)</td>
<td>314 (2.7)</td>
</tr>
</tbody>
</table>

Note that some occupants received spinal injuries at AIS 1 and AIS 2+ (in different spine regions), so the total numbers will be less that the sum of those in the severity bands. Rates have not been calculated for the ‘gender not known’ cases, since they would not be very meaningful. In addition, the relatively high incidence of this group in the overall sample means that the injury rates calculated in the known male and female groups are slightly higher than would otherwise be the case. This should be borne in mind in subsequent sections.

Although the rate of AIS 2+ spinal injury is slightly higher in females compared to males (in line with the findings of Hassan et al, 1996), the difference in the injury rates for AIS 1 is quite marked. As will become clear later, the vast majority of these AIS 1 injuries were to the neck, so this result also fits in with findings elsewhere that women are more susceptible to whiplash injury than men (see Chapter 4). Further analysis of AIS 1 injuries and of AIS 2+ injuries will be dealt with in separate sections of this chapter.
6.2 AIS 1 SPINE INJURIES

6.2.1 Influence of Gender

For each region of the spine (Cervical, Thoracic and Lumbar) the AIS 90 manual lists only one injury at a severity of AIS 1 - ie ‘strain, acute with no fracture or dislocation’ (injury codes 640278, 640478 and 640678 for the cervical, thoracic and lumbar regions respectively). Table 6.2 shows the distribution of occupants with AIS 1 injuries to the three spine regions by gender, with rates based on the total numbers of occupants, as in Table 6.1:

<table>
<thead>
<tr>
<th></th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Number</td>
<td>Rate %</td>
<td>Number</td>
</tr>
<tr>
<td>Male</td>
<td>1358</td>
<td>20.0</td>
<td>84</td>
</tr>
<tr>
<td>Female</td>
<td>1340</td>
<td>30.1</td>
<td>95</td>
</tr>
<tr>
<td>N/K</td>
<td>9</td>
<td>-</td>
<td>0</td>
</tr>
<tr>
<td>Total</td>
<td>2707</td>
<td>22.9</td>
<td>179</td>
</tr>
</tbody>
</table>

As mentioned above, the vast majority of AIS 1 spine injuries, in both males and females, are to the neck (whiplash injury), with incidence being some 50% higher in women than in men. By comparison, AIS 1 thoracic and lumbar spine injuries are rare, though lumbar injury is slightly more than twice as common as thoracic spine injury. Again, women are more susceptible than men to injury in these other spine regions.

6.2.2 Influence of Impact Direction

Impact direction information was only available for 2,619 occupants whose maximum spine injury severity (in any region) was AIS 1. Table 6.3 shows how these impact directions were distributed, with the distribution of impact directions in the overall dataset included for comparison (10,614 occupants with known impact direction):

<table>
<thead>
<tr>
<th></th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Number</td>
<td>Rate %</td>
<td>Number</td>
</tr>
<tr>
<td>Rear</td>
<td>14,472</td>
<td>55.0</td>
<td>662</td>
</tr>
<tr>
<td>Front</td>
<td>11,715</td>
<td>42.9</td>
<td>616</td>
</tr>
<tr>
<td>Total</td>
<td>26,187</td>
<td>100.0</td>
<td>1,278</td>
</tr>
</tbody>
</table>

For each of the three spine regions, rear impacts constitute a noticeably higher proportion of the impact directions than is the case in the overall dataset, by a factor of nearly two. This seems to imply that the high proportion of AIS 1 spine injuries which occur in frontal impacts is simply due to the predominance of that particular impact direction in accidents generally, whereas rear impacts actually carry a greater risk of spine injury at this severity. As far as neck injury is concerned, this would be in line with findings elsewhere (Morris & Thomas, 1996).
6.2.3 Influence of Impact Speed

Estimates of impact severity are available in the database, in the form of an Equivalent Test Speed, related to barrier impacts, although these are not calculated in every case. They are available for 1572 of the MAIS 1 spine injury cases, and for 6201 occupants in the overall dataset. Table 6.4 shows average ETS values (in km/hr) for males and females in the various spine region categories, with figures for the overall dataset again included for comparison:

<table>
<thead>
<tr>
<th>Sex</th>
<th>All Occupants</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Freq. Ave ETS</td>
<td>Freq. Ave ETS</td>
<td>Freq. Ave ETS</td>
<td>Freq. Ave ETS</td>
</tr>
<tr>
<td>Male</td>
<td>3541 29.8</td>
<td>720 27.2</td>
<td>37 30.0</td>
<td>112 29.0</td>
</tr>
<tr>
<td>Female</td>
<td>2378 27.6</td>
<td>732 25.3</td>
<td>52 24.8</td>
<td>103 25.1</td>
</tr>
<tr>
<td>N/K</td>
<td>282 23.3</td>
<td>6 19.0</td>
<td>0 -</td>
<td>0 -</td>
</tr>
<tr>
<td>Total</td>
<td>6201 28.7</td>
<td>1458 26.2</td>
<td>89 26.9</td>
<td>215 27.1</td>
</tr>
</tbody>
</table>

In the overall dataset, there is a trend for males to have received their injuries in higher severity accidents than females. This is repeated among those with AIS 1 spine injuries, and is even more pronounced in the thoracic and lumbar injury groups. Comparing figures along the "Total" row, there is a slight tendency for AIS 1 spine injuries to have been incurred in lower severity impacts than those for the overall dataset. This is to be expected, since the overall dataset covers all injury severities. There is little difference between the individual spine regions, although in the thoracic and lumbar injury columns, this trend towards lower impact severity is mainly due to the influence of the female group.
6.2.4 Occupant Contacts

Details of the likely cause of each injury suffered by an occupant are recorded in the database, although the level of detail recorded changed substantially half-way through Phase 5. Phase 5b data must therefore be presented separately from that relating to Phases 4 and 5a. Tables 5 and 6 show the numbers of injuries subdivided by likely occupant contacts for the three spine regions for the Phase 4 and 5a data and the Phase 5b data respectively.

More than half the neck injuries in Table 6.5 are "non-contact" injuries - ie caused purely by deceleration, or inertial forces, a mechanism which is widely recognised in association with whiplash injuries, even though it is unclear exactly how, or at what stage in the impact, these forces produce the injury. Almost one quarter of the neck injuries are associated with a head impact, although it is always possible that many of these were really deceleration injuries, with the head contact being coincidental, so that prevention of the head contact would not have prevented the neck injury.

As one progresses down the spine, it clearly becomes more difficult to assign causes to the injuries, with well over half the lumbar injuries in the "Contact not known" category. Only about 8% of the thoracic and lumbar spine injuries are clearly correlated with a spine impact, while indirect loading (where the investigator was unable to identify a specific impact to the spine to account for the injury, so that the loading was assumed to be via another part of the body) accounts for between 30 and 40% of the injuries in these areas.

<table>
<thead>
<tr>
<th>Type of Contact</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td>No Impact</td>
<td>1174</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Head Impact</td>
<td>456</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Spine Impact</td>
<td></td>
<td>10</td>
<td>26</td>
</tr>
<tr>
<td>Indirect Loading</td>
<td></td>
<td>51</td>
<td>102</td>
</tr>
<tr>
<td>Other</td>
<td>9</td>
<td>4</td>
<td>5</td>
</tr>
<tr>
<td>Not Known</td>
<td>371</td>
<td>63</td>
<td>175</td>
</tr>
<tr>
<td>Total</td>
<td>2042</td>
<td>128</td>
<td>308</td>
</tr>
</tbody>
</table>
TABLE 6.6. Occupant Contacts by Region of Spine Injured (Phase 5b)

<table>
<thead>
<tr>
<th>Type of Contact</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td>Non-contact</td>
<td>441</td>
<td>22</td>
<td>66</td>
</tr>
<tr>
<td>Airbag</td>
<td>2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Head restraint/seat</td>
<td>10</td>
<td>10</td>
<td>13</td>
</tr>
<tr>
<td>Seat belt</td>
<td>11</td>
<td></td>
<td>1</td>
</tr>
<tr>
<td>Fascia/A-pillar/windscreen</td>
<td>17</td>
<td></td>
<td>2</td>
</tr>
<tr>
<td>Steering wheel</td>
<td>12</td>
<td>1</td>
<td>0</td>
</tr>
<tr>
<td>Side structures</td>
<td>26</td>
<td></td>
<td>1</td>
</tr>
<tr>
<td>Roof/sunroof</td>
<td>9</td>
<td></td>
<td>1</td>
</tr>
<tr>
<td>Knee impact</td>
<td></td>
<td></td>
<td>1</td>
</tr>
<tr>
<td>Side loading of pelvis</td>
<td></td>
<td></td>
<td>1</td>
</tr>
<tr>
<td>External object</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Other</td>
<td>54</td>
<td></td>
<td>4</td>
</tr>
<tr>
<td>Not Known</td>
<td>69</td>
<td>10</td>
<td>14</td>
</tr>
<tr>
<td>Total</td>
<td>652</td>
<td>44</td>
<td>103</td>
</tr>
</tbody>
</table>

In Table 6.6, non-contact injuries again form a very large proportion of the total neck injuries, but here they also constitute half of the thoracic and over 60% of the lumbar spine injuries. This contact category was not available for thoracic and lumbar injuries in Phases 4 and 5a, so it is difficult to relate these high figures to the distribution in Table 11, unless "indirect loading" can be taken to be equivalent to "non-contact". Although Side Structures stand out in the neck injury contacts, this is partly because the steering wheel has been separated out from the other frontal structures (Fascia/A-pillar/windscreen). Although the head restraint and seat have been lumped together, it was the seat which was the important contact for the thoracic and lumbar spine injuries.

6.2.5 Conclusions for AIS 1 Injuries

i. Women were confirmed to be more at risk of neck injury than men; similar trends were observed for the thoracic and lumbar regions.

ii. Rear impact appeared to carry a higher risk of AIS 1 spine injury in all three spine regions, although frontal impacts produced greater absolute numbers of injuries.

iii. There was a slight trend towards females being more likely to receive spine injuries in lower severity impacts.

iv. Many of these low-severity spine injuries were found to be non-contact injuries, or to involve indirect loading of the spine. Head impacts were quite common in neck injury cases.
6.3 AIS 2+ SPINE INJURIES

6.3.1 Types of Injuries

The 419 AIS 2+ injuries in the sample can be classified as shown in Table 6.7:

<table>
<thead>
<tr>
<th>Type of Injury</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No.</td>
<td>%</td>
<td>No.</td>
</tr>
<tr>
<td>Cord Damage</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>NFS</td>
<td>9</td>
<td>4.5</td>
<td>2</td>
</tr>
<tr>
<td>No # or disloc.</td>
<td>4</td>
<td>2.0</td>
<td>0</td>
</tr>
<tr>
<td>With #</td>
<td>10</td>
<td>5.0</td>
<td>3</td>
</tr>
<tr>
<td>With disloc.</td>
<td>5</td>
<td>2.5</td>
<td>0</td>
</tr>
<tr>
<td>With # &amp; disloc.</td>
<td>13</td>
<td>6.5</td>
<td>1</td>
</tr>
<tr>
<td>No Cord Damage</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>#</td>
<td>129</td>
<td>64.9</td>
<td>87</td>
</tr>
<tr>
<td>Disloc.</td>
<td>22</td>
<td>11.1</td>
<td>2</td>
</tr>
<tr>
<td>Disc Damage</td>
<td>2</td>
<td>1.0</td>
<td>1</td>
</tr>
<tr>
<td>Nerves</td>
<td>5</td>
<td>2.5</td>
<td>1</td>
</tr>
<tr>
<td>Total</td>
<td>199</td>
<td>100</td>
<td>97</td>
</tr>
</tbody>
</table>

(# = Fracture; NFS = Not Further Specified)

Nearly half of the 419 injuries were to the neck - a much smaller proportion than seen in the AIS 1 injuries, but still high for such a physically relatively small area of the whole spine. This reflects the relatively weak and exposed nature of the neck which, in an impact, must deal with inertial forces imposed by the head, without the benefit of stiffening structures like the rib cage or the massive supporting muscles of the lower back. Fracture without cord damage is the most frequent injury type, particularly for the thoracic and lumbar spine. Just over 20% of cervical injuries involved cord damage, compared with only 6% for the thoracic and 14% for the lumbar regions. These proportions are somewhat different from those reported by Hassan et al (1996), who analysed earlier versions of the CCIS data, and found cord damage in 36% of cervical, 26% of thoracic and 5% of lumbar spine injuries (AIS 2+). Hassan et al also found a higher overall rate of AIS 2+ spine injury (3.4% as opposed to 2.7% in the present analysis). It is not possible to say whether this represents a genuine trend towards reduced incidence of moderate to fatal spine trauma.
6.3.2 Influence of Gender

Table 6.8 shows the numbers of occupants with AIS 2+ spine injuries subdivided by gender and spine region. Again, the injury rates are calculated from the numbers in the overall dataset. There is a slight trend for females to be more susceptible to neck injury, but this is nowhere near as marked as the trend seen for AIS 1 neck injuries. Women appear to be slightly less susceptible to thoracic spine injury at AIS 2+, while the rates of injury in the lumbar region are identical for both sexes.

<table>
<thead>
<tr>
<th>Sex</th>
<th>Cervical</th>
<th></th>
<th>Thoracic</th>
<th></th>
<th>Lumbar</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Number</td>
<td>Rate %</td>
<td>Number</td>
<td>Rate %</td>
<td>Number</td>
<td>Rate %</td>
</tr>
<tr>
<td>Male</td>
<td>92</td>
<td>1.4</td>
<td>65</td>
<td>1.0</td>
<td>44</td>
<td>0.6</td>
</tr>
<tr>
<td>Female</td>
<td>82</td>
<td>1.8</td>
<td>33</td>
<td>0.7</td>
<td>28</td>
<td>0.6</td>
</tr>
<tr>
<td>N/K</td>
<td>0</td>
<td>-</td>
<td>0</td>
<td>-</td>
<td>0</td>
<td>-</td>
</tr>
<tr>
<td>Total</td>
<td>174</td>
<td>1.5</td>
<td>98</td>
<td>0.8</td>
<td>72</td>
<td>0.6</td>
</tr>
</tbody>
</table>

6.3.3 Influence of Impact Direction

Impact direction information was available for 301 of the MAIS 2+ spine injury cases. The impact direction distribution is shown in Table 6.9, along with the distribution in the overall dataset for comparison.

<table>
<thead>
<tr>
<th>Impact Direction</th>
<th>All Occupants</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Freq.</td>
<td>%</td>
<td>Freq.</td>
<td>%</td>
</tr>
<tr>
<td>Right</td>
<td>1109</td>
<td>10.4</td>
<td>12</td>
<td>7.1</td>
</tr>
<tr>
<td>Back</td>
<td>924</td>
<td>8.7</td>
<td>10</td>
<td>5.9</td>
</tr>
<tr>
<td>Left</td>
<td>1009</td>
<td>9.5</td>
<td>36</td>
<td>21.3</td>
</tr>
<tr>
<td>Front</td>
<td>6770</td>
<td>63.8</td>
<td>84</td>
<td>49.7</td>
</tr>
<tr>
<td>Non-horizontal</td>
<td>802</td>
<td>7.6</td>
<td>27</td>
<td>16.0</td>
</tr>
<tr>
<td>Total</td>
<td>10614</td>
<td>100</td>
<td>169</td>
<td>100</td>
</tr>
</tbody>
</table>

In contrast to AIS 1 spine injuries, rear impacts do not appear to pose an increased risk of injury at AIS 2+. However, there is evidence that all three spine regions are more vulnerable in non-horizontal impacts. Many of these are likely to involve rollovers, with quite complicated occupant kinematics, but they may be indicative of the spine’s vulnerability to compressive loads via roof contact in such collisions.

The increased incidence of thoracic spine injuries in right hand side impacts is perhaps unsurprising, given that the vast majority of occupants in the database are drivers, who are likely to suffer direct torso loading from intrusion in such impacts. The preponderance of drivers in the database should also be borne in mind when considering the increased incidence of neck injuries in left hand side impacts. It may be that, in a right hand side impact, head contact with the side structures tends to mitigate
serious neck injury (though one can imagine that the resultant head injuries could be quite serious) while, in a left hand side impact a driver’s head has no nearby support available, leaving the neck alone to cope with the inertial forces imposed by the head.

6.3.4 Influence of Impact Speed

Equivalent Test Speed was available for 151 of the occupants with MAIS 2+ spine injuries. Table 6.10 shows the average ETS values (in km/hr) for males and females in the various spine region categories, along with the values from the overall dataset for comparison:

<table>
<thead>
<tr>
<th>Sex</th>
<th>All Occupants</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Freq.</td>
<td>Ave ETS</td>
<td>Freq.</td>
<td>Ave ETS</td>
</tr>
<tr>
<td>Male</td>
<td>3541</td>
<td>29.8</td>
<td>43</td>
<td>35.8</td>
</tr>
<tr>
<td>Female</td>
<td>2378</td>
<td>27.6</td>
<td>40</td>
<td>43.8</td>
</tr>
<tr>
<td>N/K</td>
<td>282</td>
<td>23.3</td>
<td>0</td>
<td>-</td>
</tr>
<tr>
<td>Total</td>
<td>6201</td>
<td>28.7</td>
<td>83</td>
<td>39.6</td>
</tr>
</tbody>
</table>

As might be expected, AIS 2+ spine injuries tend to occur at higher impact severity than the general run of accidents in the database. It is interesting to note that women in this AIS 2+ subset received their neck injuries in higher severity accidents than their male counterparts. This runs counter to the overall trend and to the AIS 1 neck injury trend. For thoracic spine injuries, however, impact severity among males was considerably higher than that among females, and also much higher than that for males in the neck and lumbar injury categories. This may tie in with the generally much stronger chest and shoulder musculature in men compared to women. Male and female lumbar spine injury victims, on the other hand, were much closer in their impact severity estimates.

6.3.5 Occupant Contacts

The occupant contact data for Phases 4 and 5a are shown in Table 6.11.

For the neck, we again have a marked difference between the AIS 1 sample and the AIS 2+ sample. In the former, more than half the injuries were non-contact, deceleration injuries, whereas this category only constitutes about 11% of the neck injuries in Table 6.11. Head injuries, which were associated with about a quarter of the AIS 1 neck injuries, only account for 17% of the neck injuries in Table 6.11. Direct impacts to the neck, however, which were almost negligible among AIS 1 neck injury victims, are responsible for 44% of AIS 2+ injuries. The thoracic and lumbar spine regions show similar changes, with spine impacts up from 8% to 22% in the thoracic spine category, and from a similar level to 16% for the lumbar region, with corresponding reductions in the proportion of indirect loading cases.
TABLE 6.11. Occupant Contacts by Region of Spine Injured (Phases 4 and 5a)

<table>
<thead>
<tr>
<th>Type of Contact</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td>No Impact</td>
<td>17</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Neck Impact</td>
<td>68</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Head Impact</td>
<td>26</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Spine Impact</td>
<td></td>
<td>21</td>
<td>12</td>
</tr>
<tr>
<td>Indirect Loading</td>
<td></td>
<td>31</td>
<td>19</td>
</tr>
<tr>
<td>Other</td>
<td>1</td>
<td>3</td>
<td>0</td>
</tr>
<tr>
<td>Not Known</td>
<td>41</td>
<td>42</td>
<td>45</td>
</tr>
<tr>
<td>Total</td>
<td>153</td>
<td>97</td>
<td>76</td>
</tr>
</tbody>
</table>

Table 6.12 lists the occupant contacts for the Phase 5b data:

TABLE 6.12. Occupant Contacts by Region of Spine Injured (Phase 5b)

<table>
<thead>
<tr>
<th>Type of Contact</th>
<th>Cervical</th>
<th>Thoracic</th>
<th>Lumbar</th>
</tr>
</thead>
<tbody>
<tr>
<td>Non-contact</td>
<td>11</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Airbag</td>
<td>1</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Head restraint/seat</td>
<td>3</td>
<td>1</td>
<td>9</td>
</tr>
<tr>
<td>Seat belt</td>
<td>0</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>Fascia/A-pillar/windscreen</td>
<td>1</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Steering wheel</td>
<td>0</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Side structures</td>
<td>2</td>
<td>5</td>
<td></td>
</tr>
<tr>
<td>Roof/sunroof</td>
<td>1</td>
<td></td>
<td></td>
</tr>
<tr>
<td>External object</td>
<td>4</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>Other</td>
<td>10</td>
<td>2</td>
<td>1</td>
</tr>
<tr>
<td>Not Known</td>
<td>12</td>
<td>10</td>
<td>4</td>
</tr>
<tr>
<td>Total</td>
<td>45</td>
<td>25</td>
<td>22</td>
</tr>
</tbody>
</table>

Cervical spine non-contact injuries are much more common in this group - about a quarter, while "other" and "not known" categories account for a further half. Of the definite contacts, external objects and the seat/head restraint are the most common. Among the thoracic spine injuries, side structures and the steering wheel are, perhaps unsurprisingly, most commonly associated with AIS 2+ injuries, whereas these two were hardly represented at all among the AIS 1 injury contacts. Nearly half the lumbar injuries were associated with a seat contact.
6.3.6 Conclusions for AIS 2+ Injuries

i. Fracture without cord damage was found to be the most common injury for all three spine regions, although cord damage also featured strongly in the neck injury cases.

ii. There were only fairly small differences in injury susceptibility between males and females, in contrast to the AIS 1 injury sample.

iii. There was evidence of increased risk of spine injury at all levels in non-horizontal impacts. The thoracic spine showed increased risk of injury in right hand impacts, while the neck showed increased risk in left hand impacts.

iv. Impact severity trends between males and females were rather confusing, with females showing higher average impact severity for neck injury, although the trends for thoracic and lumbar injury were more in line with expectations.

v. A much higher proportion of direct impacts to the spine was seen among the AIS 2+ cases compared to the AIS 1 sample, although the Phase 5b data still showed a large number of non-contact neck injury cases.
7 MATHEMATICAL MODELLING OF THE HUMAN SPINE

7.1 INTRODUCTION

Injury analysis requires that the mechanical behaviour of the spine is represented in detail: a model must not only describe the global kinematics and dynamics of the spine, head and pelvis but also the local kinematics and dynamics of individual vertebrae and other relevant spinal components.

The conventional two step method of relating injury to the accident or input conditions that cause injury is, first, a calculation is made of the relative dynamic deflection of the body member in response to the forces or accelerations applied to that body member. Second, injury level is determined by comparing the degree of mechanical response or deflection with a data base of injuries that have been produced at the same response level. The predictive accuracy of this method depends on knowledge of the input parameters, typically determined by the accident investigator, on the adequacy of the input-response calculation methodology, on the accuracy of the characterization of the dynamic mechanical properties of the body members experiencing the accident loading and on the adequacy of the data base of injuries versus response levels. Mathematical modelling similarly depends on the mathematical characterization of body members and systems and similarly requires validation against actual loading responses.

According to Ward and Nagendra (1985), the major pitfalls in mathematical modelling of biological systems are: over sophistication, lack of good physical property data, and lack of validation. Over sophistication will result in a model including too many details whose effect on the behaviour of the model will be difficult to retrieve. During the process of modelling, numerous assumptions and simplifications usually have to be introduced, partly due to the lack of reliable physical property data and partly to reduce the complexity of the model. To check on the assumptions used, a model has to be validated. Validation is achieved by correlating numerical predictions with actual experimental results of the accident mechanisms for which the model will be used.

Qualitative description of the biomechanical behaviour of the cervical spine is available in the literature: Huelke and Nusholtz 1986, Kazarian 1982, McElhaney and Myers 1992, Sances et al 1984, White and Panjabi 1990. However, few describe quantitative biomechanical aspects important for modelling the spine. Further knowledge of the biomechanical response of the spine under traumatic static and dynamic, transient loads and of the stability of the injured spine would lead to a better understanding of the injury mechanisms.

Experimental study of spinal biomechanics both in vivo(living) and in vitro (dead) is difficult and often not reliable enough for modelling purposes. Mathematical modelling is also difficult due to the very complicated nature of the spine. This chapter presents a review of research of biomechanical mathematical modelling of the spine system and the spinal sub systems.
7.2 MODELLING TECHNIQUES AND CONSIDERATIONS

The spine is a mechanically, geometrically and biologically complex structure which does not readily lend itself to mathematical modelling. However, driven by the need to better understand the mechanics of spinal injuries many researchers have developed simplified models using a variety of analytical techniques. An extensive review has been provided by Yoganandan et al (1987) who categorized mathematical models into four distinct types.

The four major classifications are based on geometrical and force considerations, the type of analysis and the particular application of the mathematical model. Each of these classifications is then subdivided. For example, from the force point of view, the model could be used by applying static forces or dynamic loads. The former can be used to predict stresses in the lumbar spine (low back), while the latter can be used to study the wave propagation characteristics along the spinal column.

The first category is based on geometrical considerations. Although the geometry of the spinal column is fixed, analytical modelling requires a certain degree of idealization. System models or macro models treat the system as a single structural entity, whereas component models or micro models investigate the response of a particular component, such as the intervertebral disk.

The second of these categories is based on force considerations. Depending on the nature of the external loading, spinal models can be classified as static or dynamic. In static (quasi static) models, the effects of inertia are neglected. In dynamic analysis, the system response before it reaches a steady state is known as a transient. In this case inertial effects are included. Kinematics is the study of the motion of a structure in response to forces and other loads on the system, but does not necessarily consider internal force distribution i.e. stress.

The third category is determined by type of analysis. Two types of analyses, forward and inverse, are routinely performed to study the mechanics of the structure. In forward analysis, with the input of the geometry, basic laws of material behaviour, loading and boundary condition data, deformations can be predicted. The inverse method of analysis helps to estimate material properties with the geometry, loading and boundary condition data, and deformations as input. Even in the linear elastic domain, iterative techniques are necessary to arrive at the solution, and the solution is not always unique.

The final category is based on type of application. From a mechanics standpoint, the application lies in the prediction of stress-strain distributions, estimation of material constants, and tolerance limits. From a clinical viewpoint, it may be used to evaluate the causes of low back pain, damage to the central nervous system, and trauma.

These classifications are somewhat overlapping. A large number of spinal models including static and dynamic, continuum and discrete parameter, and system and component models with varying degrees of complexity have been studied in the past.

Dynamic models were developed when it was necessary to determine the spinal injury tolerance when using aircraft ejection-escape systems in high-speed aircraft which result in high ejection velocities. For the pilot-ejection application, the spine was idealized as a single-degree-of-freedom lumped parameter system. To account for the various levels of injury, several modifications were made and resulted in multi-degree-of-freedom models. These are limited in the sense that they fail to recognize where, when, and how the spinal injury takes place. Continuum and discrete parameter dynamic models of varying complexity were advanced to solve this issue. Initially, the vertebral column was conceived as a bar or beam column, neglecting the effects of torso and lateral bracing due to the rib cage. These models fell into one of two categories: classical continuum mechanics models or discrete parameter models. In the former, the bar is assumed to be homogeneous and the analysis is carried out using the principles of solid mechanics. However, discrete parameter models idealize the spine as an assemblage of individual rigid vertebrae connected by spring or spring/dashpot elements representing the intervertebral disk and the surrounding tissue complex. If these models have very few degrees of freedom, they are referred to as lumped parameter models. The basic difference between the discrete
parameter and the continuum approach is that the former has the capability to utilize the characteristics of the vertebral column at various levels (cervical, thoracic, and lumbar), whereas the latter requires a continuum representation of the characteristics of the spinal elements as a matrix. However, if the scale of discretization employed in the continuum category is similar to the discrete models, then the governing continuum equations would be similar to that of the other type. Some models following these principles were correlated with experimental data, while some remained purely academic.

Continuum models are probably the simplest model form, treating the structure as a homogeneous material subjected to various loads and boundary conditions at the ends. The first continuum spinal model was developed in the late 1950s by Hess and Lombard (1958) to study the problem of pilot ejection. This model treated the entire spine as a straight, homogeneous, linear elastic rod, free at the top, with a prescribed acceleration loading at the bottom. This model was modified in subsequent years by a number of researchers to include viscoelasticity (Terry and Roberts, 1968), head mass (Liu and Murray, 1966) and spinal curvature (Krause and Shirazi, 1971; Li et al, 1971; Moffatt et al, 1971). Soechting and Paslay (1973) developed a model which included the effect of spinal musculature to study the flexural response of the spine.

Lumped parameter models were also developed to study the pilot ejection problem. Latham (1957) developed an elastic single degree-of-freedom model to study the effect of varying seat cushion resilience on the spinal response. Payne (1969) added damping to this model to study the probability of compression fractures of vertebral segments and developed the dynamic response index. One limitation of this model was its inability to predict the location of spinal injuries. This limitation was overcome by Toth (1966), who developed an 8 degree of freedom, non-linear spring-mass-dashpot model and was able to predict failure in the upper lumbar spine.

Head and neck lumped parameter models have been developed to investigate both athletic and motor vehicle related injuries. McElhaney et al (1983) and Sances et al (1984) developed lumped parameter models of the neck to study the effect of helmets on neck injuries. Reber and Goldsmith (1979) developed a two-dimensional model to study head and neck response during whiplash. This model contained 11 mass elements and 76 springs and dashpots to represent the head and neck down to the T3 spinal level.

Interest in investigating injuries resulting from automotive crashes has led to the development of general purpose kinematic simulation programs. These include the Calspan Three Dimensional Crash Victim Simulator (CVS), now called the Articulated Total Body Model (ATB), and MADYMO. Using these programs, a structure is represented by a series of ellipsoidal bodies connected together by various types of joints. Although designed for general purpose use, these programs have been used primarily to simulate occupant kinematics during various automotive crash scenarios. Anthropometric data sets are provided with these models including joint characteristics and full body geometry, originally developed by Baughman (1983). Recent developments in MADYMO have enabled finite element components to be included in the simulations.

Earlier continuum mathematical models intended to predict the stability limit loads using Euler buckling theory indicated unrealistic values. Inclusion of the lateral bracing due to the rib cage increased the limit by three to four times. However, even this was well below the in vivo physiological spinal loads. In principle, continuum mechanics models provide exact answers. However, discrete parameter models were advanced to include the complex features of the spine, such as ligaments responding only to tensile loads, nucleus of the disk behaving as an incompressible fluid, and interconnections between the rigid vertebrae and deformable disks. These models described the equilibrium of the structure and some of them dealt with the geometrical constraints as seen in abnormal or diseased spines.

The application of these models to completely quantify the mechanics of the spine is limited because of the complex geometry and material of the human vertebral column. Furthermore, being a biological parameter and the continuum approach is that the former has the capability to utilize the characteristics of the vertebral column at various levels (cervical, thoracic, and lumbar), whereas the latter requires a continuum representation of the characteristics of the spinal elements as a matrix. However, if the scale of discretization employed in the continuum category is similar to the discrete models, then the governing continuum equations would be similar to that of the other type. Some models following these principles were correlated with experimental data, while some remained purely academic.

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The application of these models to completely quantify the mechanics of the spine is limited because of the complex geometry and material of the human vertebral column. Furthermore, being a biological
structure, the mechanical response is not linear. Numerical methods such as finite element methods are advantageous in contrast to closed-form solutions in being able to address the particular modelling challenges presented by the analysis of the structures such as spinal biomechanics.

7.3 FINITE ELEMENT MODELS

Finite element (FE) methods of structural analyses were introduced in 1956. The first applications of the method were in the aircraft industry. In biomechanics, the early applications were models of cardiovascular problems in 1969, the thorax in 1970, the skull in 1971, bones in 1972 and the spine in 1973. In studies of the mechanics of the spine, FE principles of stress analysis have been used mainly to predict injury mechanisms and to estimate intervertebral disc material properties. The FE method considers the structure as being divided into a finite number of small divisions called elements to which material properties (modulus of elasticity, etc.) are assigned. A stiffness matrix for each of the constituting elements is computed to describe its flexural response to imposed loads. By suitably assembling the stiffness matrices of the individual elements, a stiffness matrix for the global structure can be obtained. Necessary boundary conditions and external loading on the structure are included in the system equations which are then solved for displacements. Once the displacements are computed, secondary quantities such as strains, stresses, bending moments, shears, etc., can be derived by theory of elasticity principles. When modelling the spine it can be considered as a structure formed by various anatomical components, but for the purposes of FE analysis each component is broken down into a large number of deformable elements, each having the continuum material properties of the anatomical component it belongs to. In principle, the finite element method can accommodate any type of geometry, loading, material behaviour and boundary condition data. However, because it is an approximate method, the results are dependent on the choice of element as well as the number of elements and assumptions made in the analysis. Thus, any conclusions from the FE analysis must be reviewed in light of the assumptions made.

Applications of FE modelling to the spine was reported by Liu and Ray in 1973. Early models of the spine were two-dimensional and focused primarily on determining the material properties of the intervertebral discs (Belytschko et al, 1974; Kulak et al, 1976; Liu and Ray, 1978). This represents an inverse problem known as system identification, whereby the loads and deflections are imposed on the model and the material properties are estimated. This method involves the use of optimization techniques to solve for the unknown material properties, bringing the uniqueness and convergence of the solution into question (Yoganandan, et al, 1987).

Hakim and King (1979) modelled a single vertebra including boundary element representations of the intervertebral disc and articular facets. This model was later extended by Yang and King (1984) to include a complete spinal motion segment (two adjacent vertebrae and interconnecting soft tissue). Shirazi-Adl et al (1984, 1986) developed a model of a lumbar motion segment and subjected it to axial compression, torsion, and a combination of compression and torsion. Ueno and Liu 1987, also modelled the lumbar motion segment subjected to torsion. Goel et al 1988, studied the effect of spinal fixation devices on the stress distribution within a motion segment. Ligaments were included in this model as cable elements active only in tension. Using their model, they studied the response of an intact specimen in all loading modes and of a stabilized specimen in flexion, extension, and axial compression.

In 1983, Williams and Belytschko developed a cervical spine model, including the head, for evaluating the responses to frontal and lateral impacts. The head and vertebrae were treated as rigid bodies, connected by beam and spring elements representing the intervertebral disc, muscles, facet joints, and ligaments. M Kleinberger developed in a paper presented to the 1986 Stapp Crash Conference a complete ligamentous cervical spine model, including a rigid head to provide proper application of non-contact inertial loading. The model includes all cervical vertebrae (C1-T1), intervertebral discs, articular facets, and biomechanically relevant spinal ligaments. Musculature was
intended to be added to the model following the completion of an experimental study being conducted to determine the mechanical properties of skeletal muscle.

Finite element models are normally constructed from a 'typical' geometry, making it difficult to account for geometric variations from one specimen to the next. To investigate this problem, Lavaste et al (1992) have developed a method to create a finite element mesh based on digitized bi-planar X-rays from a tested lumbar specimen. This procedure helps to facilitate model validation by eliminating the geometric differences between model and experimental results. This technique was used to construct a complete lumbar spine model which could then be validated against experimental data obtained from the geometrically similar specimen. Breau et al (1991) have used CT (computerised tomography) X-ray scans to reconstruct the three-dimensional geometry of lumbar vertebrae.

The major drawback of finite element modelling is that it yields complex models (with many parameters) that can be computationally expensive. Discrete parameter models are a subset of simplified FE models with fewer degrees of freedom and, therefore, fewer equations but they don’t predict internal stresses for instance. Thus both discrete parameter and finite element models may be suited to describe the mechanics of the spine in detail.

Dietrich et al (1991) describe a three-dimensional finite element model of the human spinal system. The model includes the spine (vertebrae C3-L5), sacrum, pelvis and ribcage, modelled as rigid bodies. They omitted the atlas and axis because of the different function and shape of these vertebrae. The soft tissue components are modelled with deformable finite elements. Basic ligaments of the spine and important muscles that influence behaviour of the spinal system are included too. External forces (static load or inertial forces) can be applied to the model. The model allows for both static and dynamic analysis of forces occurring in the spinal system. An example of a static analysis is included in the paper. Lavaste et al (1992) completed a study to design a three-dimensional geometrical and mechanical finite element model of the lumbar spine. The model's geometry is constructed using six parameters per vertebra. These parameters are digitized from two X-rays (anterior-posterior and lateral), thus yielding an individualized model which can be arrived at from the radiographs of a tested specimen. This procedure makes the model validation easier, as geometry is generally a factor of dispersion in experimental results. The geometrical reconstruction, in the form of a finite element mesh, was effected for the whole lumbar spine. The global coherence of the model was verified and the design provides a reliable model of the spine which allows the analysis of the influence of geometrical factors on the mechanics of the lumbar spine. A summary of finite element models of the spine is presented in Table 7.1.
<table>
<thead>
<tr>
<th>Author(s) (year)</th>
<th>Study</th>
<th>Model</th>
<th>Geometry</th>
<th>Purpose</th>
<th>Experimental validation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Liu and Ray (1973)</td>
<td>Spine</td>
<td>Two-dimensional linear layered media; elastic large deformation excluded</td>
<td>Initially curved spine; soft tissues as rigid torso</td>
<td>Inertial (dynamic)</td>
<td>Stress distribution at disc-vertebra interface (wave propagation)</td>
</tr>
<tr>
<td>Belytschko (1974)</td>
<td>Lumbar disc*</td>
<td>Two-dimensional axisymmetric linear elastic</td>
<td>Rotational and horizontal symmetric 1/4 model</td>
<td>Axial compression (static)</td>
<td>Stress distribution orthotropic (4) elastic constants</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Initially curved linear layered Soft LiSSUC5</td>
<td>Three-noded triangular (Wilson)</td>
<td></td>
<td>Matching load-deformation curves</td>
</tr>
<tr>
<td>Liu and Ray (1975)</td>
<td>Spine</td>
<td>Two-dimensional primate media elastic</td>
<td>Initially curved linear layered spine</td>
<td>Inertial (dynamic)</td>
<td>Stress distribution at disc-vertebra interface (wave propagation)</td>
</tr>
<tr>
<td>Liu et al (1975)</td>
<td>Lumbar disc</td>
<td>Three-dimensional linear elastic model</td>
<td>As close as possible, 1/2 model</td>
<td>Direct PA shear (static)</td>
<td>Orthotropic (5) constants; shear most sensitive</td>
</tr>
<tr>
<td>Kuliak et al (1976)</td>
<td>Lumbar and thoracic discs</td>
<td>Two-dimensional axisymmetric linear elastic</td>
<td>Rotational and horizontal symmetric 1/4 model</td>
<td>Axial compression tension (static) linear parameter</td>
<td>Orthotropic constants and a nunciation behavior for facets</td>
</tr>
<tr>
<td>Koogle et al (1979)</td>
<td>Lumbar spine</td>
<td>Three-dimensional Large displacement</td>
<td>Vertebra, elastic; ligaments, springs and dashpots</td>
<td>Dynamic loading</td>
<td>Orthotropic (9) constants - minimize the index of performance Dynamic response</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Their own experiments</td>
</tr>
<tr>
<td>Author(s) (year)</td>
<td>Study</td>
<td>Model</td>
<td>Geometry and PROGRAM</td>
<td>Loading input</td>
<td>Purpose output</td>
</tr>
<tr>
<td>-----------------</td>
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<tr>
<td>Radons et al (1979)</td>
<td>Spine</td>
<td>Linear sagittal symmetric body; disc, finite beam; ribs, curved beam; muscle-ligament, spring</td>
<td>Vertebral, rigid symmetric body; disc, finite beam; ribs, curved beam; muscle-ligament, spring</td>
<td>Harmonic excitation at L1 (dynamic)</td>
<td>Spinal fusion and Harrington rod studies; natural frequency</td>
</tr>
<tr>
<td>Hakim and King (1979)</td>
<td>Lumbar vertebra</td>
<td>Three-dimensional linear elastic</td>
<td>Bilaterally symmetric (qualitative)</td>
<td>Body loading (static and dynamic)</td>
<td>Stress distribution in the cortex</td>
</tr>
<tr>
<td>Spilker (1980-1984)</td>
<td>Lumbar disc</td>
<td>Two-dimensional linear elastic stress model</td>
<td>Simplified with straight boundary; 1/4 model</td>
<td>Four-noded axisym ring (Spilker and Pian)</td>
<td>Axial compression, stress, bending and torsion</td>
</tr>
<tr>
<td>Yang and King, and Yang et al (1983)</td>
<td>Lumbar disc</td>
<td>Three-dimensional large displacement, linear</td>
<td>Complete with articular facets; bilateral symmetry</td>
<td>Eight-noded brick bar and shell elements (FEAP)</td>
<td>Axial compression 1100 N (38% facets); (Static)</td>
</tr>
<tr>
<td>Ueno et al (1983)</td>
<td>Lumbar disc</td>
<td>Three-dimensional linear</td>
<td>Sagittal and horizontal symmetry, end plates, nucleus, facets</td>
<td>Eight-noded brick (SAP 4)</td>
<td>Axial compression (static)</td>
</tr>
<tr>
<td>Author(s) (year)</td>
<td>Study</td>
<td>Model</td>
<td>Geometry</td>
<td>Elements &amp; PROGRAM</td>
<td>Loading input</td>
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<tr>
<td>Rencis (1982)</td>
<td>Lumbar vertebra</td>
<td>Two-dimensional axisymmetric linear</td>
<td>Bilaterally symmetric</td>
<td>Four-noded quadrilateral and shell elements</td>
<td>Uniform axial compression</td>
</tr>
<tr>
<td>Hosey and Liu (1983)</td>
<td>Head and cervical spine</td>
<td>Three-dimensional linear</td>
<td>Symmetric about mid-sagittal plane</td>
<td>Eight-noded brick four-noded shell four-noded membrane</td>
<td>Half-sine in time and uniform in space</td>
</tr>
<tr>
<td>Privitzer and Hosey (1983)</td>
<td>L1 vertebra</td>
<td>Three-dimensional linear</td>
<td>Vertebra elliptic cylinder with three planes of symmetry with and without disc</td>
<td>Eight-noded brick four-noded shell (MAGNA)</td>
<td>Uniform compression (static)</td>
</tr>
<tr>
<td>BelyLschko et al. (1976-1983)</td>
<td>Spine</td>
<td>Three-dimensional large displacement large rotation, linear and nonlinear material behavior</td>
<td>Head to L5 including muscles, rib cage, seat back, and restraints</td>
<td>Rigid bodies, springs, hydrodynamic elements; plate and shell elements</td>
<td>Acceleration time-input (Gx, Gy, and Gz) (dynamic)</td>
</tr>
<tr>
<td>Shirazi-Adl et al (1983)</td>
<td>Lumbar disc nonlinear</td>
<td>Three-dimensional</td>
<td>Annulus modeled as composite, posterior elements</td>
<td>Bilaterally symmetric, axial, and brick elements</td>
<td>Uniform axial loading (static)</td>
</tr>
<tr>
<td>Ueno (1984)</td>
<td>Lumbar disc</td>
<td>Three-dimensional geometrically nonlinear</td>
<td>Annulus modeled as composite</td>
<td>Eight-noded solid elements, fluid, cable, and gap elements; complete motion segment</td>
<td>Compression flexion, extension, and torsion (static)</td>
</tr>
<tr>
<td>Yoganandan (1985)</td>
<td>Lumbar disc Two-dimensional Rotationally axisymmetric nonlinear</td>
<td>Three-noded triangular, and horizontally symmetric model</td>
<td>Axial compression four-noded quadrilateral fluid elements, no posterior complex and ligaments</td>
<td>Estimation of nonlinear material properties and strain energy distributions in the annulus and end plate</td>
<td></td>
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<td></td>
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</table>
7.4 MODELLING THE CERVICAL SPINE

7.4.1 Modelling techniques

Four types of models describing head-neck dynamics are found in the literature: two-pivot models, continuum mechanics models, discrete parameter models, and finite element models. Pivot models are capable of describing the global motion of the head and neck relative to the torso but cannot describe the mechanics of the neck in detail. In continuum mechanics models, the cervical spine is represented as a homogeneous beam column, that is both geometry and mechanical behaviour are extensively simplified. Both discrete parameter and finite element models allow for a more detailed representation of the mechanical behaviour of the various anatomical structures of the human neck. In discrete parameter models the spine is idealised as an assemblage of individual rigid vertebrae, connected by massless spring and damper elements representing the intervertebral disc and the surrounding soft tissue complex, and (sometimes) the muscles. Mass and inertial properties of the system are lumped into the rigid vertebrae. Discrete parameter models cannot completely quantify the mechanics of the spine because of the complex geometry and material behaviour and the nonlinear mechanical response of the spine. To overcome (some of) these limitations finite element models were introduced which allow for a more detailed description of the mechanics of the spine.

An elaborate description of continuum models and two-dimensional discrete parameter models is provided by Yoganandan et al (1987) who extensively reviewed mathematical models of the spine and spinal components up to 1986 so these models are not included here. Both relatively simple two-pivot models and the more detailed discrete parameter and finite element models describing head neck dynamics are reviewed. Further, attention is given to data on the physical properties of cervical spine components. It is concluded that a number of sophisticated models are available in the literature. These models do not simulate the relative head-motion better than two-pivot models and it is concluded that data on the material characteristics of cervical components are incomplete as well as data for a detailed validation of local vertebral movements.

In studies on the biomechanics of the spine, motion segments are often used. A motion segment, sometimes referred to as a functional spinal unit (FSU), comprises two adjacent vertebrae together with surrounding soft tissues: intervertebral disc, facet joints and ligaments. It is the smallest unit exhibiting biomechanical features similar to the entire spinal column, which may be considered as a structure composed of motion segments connected in series. The total behaviour of the spine, then, is a composite of the individual motion segment behaviour.

The neck consists of three motion segments plus the atlas (C1) which has no vertebral body. Due to its functional arrangement, the occipito-atlanto axial joint is usually treated as a single biomechanical unit. Since the behaviour of motion segments is dependent upon the behaviour of its components these components should be studied too. Thus, to study and model the biomechanics of the cervical spine quantitatively, physical properties of both motion segments and components are needed. Physical properties include the geometrical, the inertial and the material characteristics.

7.4.2 Geometrical Characteristics

Geometrical characteristics include: the dimensions of vertebrae, articular facets, discs, ligaments, and muscles; the locations of the places where the soft tissues are attached to the vertebrae; and the compound configuration of all elements. Much of this information may be collected from detailed (cross-sectional) anatomy books, X-rays photographs, computer tomography scans (CT-scans), magnetic resonance imaging (MRI) and skeletal material.

Francis (1955) studied variations in the dimensions of cervical vertebrae and articular facets from human skeletal material of young adults. Nissan and Gilad (1984) reported average dimensions for the
mid-sagittal appearance of vertebrae, which is idealized by a square-box approximation for the vertebral body to which a triangular shaped arch is attached. Parameters for vertebrae C2-C7 were measured from lateral X-rays of numerous male cervical spines. Liu et al (1986) determined the geometry of all cervical vertebrae of young males by measuring the coordinates of premarked points (36 per vertebra) relative to the vertebral body centre of mass. Furthermore, the orientation of the articular facets and the articular facet-to-centre area ratios were obtained. The data on the measured coordinates are given by Liu et al (1980). Panjabi et al (1991) determined the linear dimensions, angulations and surface and cross-sectional areas of most vertebral components from three-dimensional coordinates of points on cervical vertebrae (C2-C7). Included are, among others, the dimensions of the vertebral body, spinal canal and pedicles. However, no measurements regarding articular facet dimensions were reported. They noted that their results agreed well with those from Francis, Nissan and Gilad, and Liu et al. Goel et al (1986) obtained, for three cadavers, the origins and insertions of all the major muscles of the head-neck region with respect to both anatomical and global reference planes. They also measured weight, natural length and maximum width of each muscle.

7.4.3 Inertial Characteristics

Inertial characteristics include mass, location of centre of gravity, and principal moments of inertia of head, neck and vertebrae. The characteristics assigned to the vertebrae should represent those of live vertebrae inside a complete human neck. Data on the inertial characteristics of the head and the head and neck have been reported by Beier et al (1980) and Walker et al (1973). Liu et al (1971) determined the inertial properties of horizontal segments of a human cadaver.

7.4.4 Material Characteristics

Material properties are specified by constitutive equations. The unknown parameters of these equations have to be estimated from experimental results to obtain a valid model for specific material behaviour. Experimentally, material behaviour is characterised by means of force-displacement, or stress-strain, curves, stiffnesses, load and deformation at failure, and similar parameters (Jager et al, 1993). Force-displacement curves for biomechanical structures such as motion segments or ligaments, qualitatively have the nonlinear, sigmoidal shape depicted in Figure 7.1. The curve starts with a neutral zone in which little force is needed to deform the structure. At the end of this zone, the stiffness increases substantially. The stiffness usually remains fairly linear up to failure, which is identified as a (sudden) substantial drop in force. The load and deformation at this point of failure is then defined as failure strength of the structure. Although stiffness is easily defined as the ratio of force on and related deformation of the structure, the nonlinearity of material behaviour gives rise to difficulties. The experimentally obtained force-displacement curves are often not reproduced in the literature but represented by a stiffness coefficient. However, the load-displacement curves are nonlinear and thus difficult to describe by just a stiffness coefficient. Moreover, different stiffness calculations have been used by different authors. For example, for the curve given in Figure 8.1 stiffnesses have been calculated as: (1) the ratio of (maximum) applied force and deformation at this force; (2) the ratio of (maximum) applied force and deformation at this force minus the neutral zone deformation; (3) the slope of the most linear part of the curve; (4) stiffness calculated from linear regression analysis of the curve; or (5) slope of the curve at a certain load (or deformation). Stiffness calculated by (2) or (3) is usually named "average stiffness". Calculation (5), the exact definition of stiffness for a point of the curve, may be used to represent a complete curve if stiffnesses are given for a sufficient number of points. For all definitions, the load at which or the loading range for which the stiffnesses were calculated should be given (Jager et al, 1993).
Up to 1983, three-dimensional studies on the biomechanical properties of spine segments had been limited to the thoracic and lumbar regions (Panjabi et al, 1986). Biomechanical properties of cervical spine segments have been examined by various investigators. In most studies, segments of the lower and upper cervical spine have been used to characterize experimentally the biomechanical behaviour of the cervical spine. Motion segment stiffness is calculated from the measured force-displacement curves. Different authors may use different stiffness calculations which complicates a good comparison between reported stiffness values (see Table 7.3).

White and Panjabi (1990) collected average stiffness coefficients of a "representative" motion segment of all regions of the spine for all modes of loading. Lower cervical spine studies have been conducted by Panjabi et al (1986), Moroney et al (1988) and Shea et al (1991), while the biomechanical properties of the upper cervical spine have been studied by Panjabi et al (1988) and Goel and co-workers (Chang et al, 1992; Goel et al, 1988; Goel et al, 1990).
Table 7.3: Average stiffness coefficients for lower cervical spine motion segments (C2-T1) [from Jager et al. (1993)].

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<td>tension</td>
<td>53</td>
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<td>-</td>
<td>433</td>
<td>193</td>
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<td>compression</td>
<td>200</td>
<td>141</td>
<td>1318</td>
<td>718</td>
<td>957</td>
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<tr>
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<td>183</td>
<td>123</td>
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<tr>
<td>posterior shear</td>
<td>53</td>
<td>34</td>
<td>49</td>
<td>162</td>
<td>114</td>
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<tr>
<td>lateral shear</td>
<td>53</td>
<td>53</td>
<td>119</td>
<td>-</td>
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(N/m/degree)

- flexion 0.43 - 0.43 0.43 1.13 -
- extension 0.73 - 0.73 0.73 1.88 1.74
- lateral bending 0.68 - 0.68 - -
- axial rotation 1.16 - 1.16 -

(1) reciprocal of reported average flexibility coefficients; range of loads for which coefficients were calculated is 12-38 N
(2) range of loads is 49-74 N for compression, 10-39 N for shear, and 1-2.2 Nm for moments
(3) stiffnesses calculated at 300 N compression-tension, 150 N shear, and 5 Nm flexion-extension
(4) stiffnesses calculated at 500 N compression, 100N tension or shear, and 3.5 Nm extension translation.

7.4.5 Material Characteristics of Cervical Spine Components

Yamada (1970) reported strength characteristics of numerous biological materials (organs and tissues) obtained from human and animal cadavers. With respect to the locomotor system, mechanical properties of bone and vertebrae, cartilage, intervertebral discs, ligaments, muscles and tendons are given. Yamada's data on vertebral bodies, intervertebral discs and a few other tissues relevant with respect to the spine have been reproduced by Sances et al. (1998). Fung (1981) used the principles of continuum mechanics to describe the mechanical behaviour of bio-solids (hard and soft tissues) and bio-fluids. Emphasis is put on constitutive equations that can be used to describe the behaviour of biological tissues. Constitutive equations for, among others, muscles, bone and cartilage are presented.

Vertebrae, intervertebral discs and facet joints - Yamada reported failure strength and deformation data for cervical vertebrae and discs subjected to compression or tension. Stiffness data for cervical vertebrae were not found in this review. Disc stiffnesses have been reported by Moroney et al (1988) (compression) and Pintar et al (1986) (tension). Since the deformation of vertebrae is small compared to the deformation of discs, vertebrae may be treated as rigid bodies. Depending on the application, vertebral deformation may be taken into account by transferring it to the disc characteristics. Biomechanical properties of cervical facet joints have not been found in the literature.
**Ligaments** - Force-displacement curves for spinal ligaments typically have the nonlinear, sigmoidal shape, represented in Figure 7.1. Average failure strengths of the most important upper and lower cervical spine ligaments have been collected by White and Panjabi (1990). Chazal et al (1985) studied the geometrical and biomechanical properties of various spinal ligaments subjected to quasistatic loading (1 mm/min). With respect to the cervical spine, data were obtained for the anterior and posterior longitudinal ligaments. Included are mean values for stresses and strains at three characteristic points of the sigmoidal force-displacement curve. Dvorak et al (1988) reported the failure strength and the dimensions of the transverse ligaments of the upper cervical spine. The ligaments were loaded quasistatically at a rate of 1.5 mm/s. Myklebust et al (1988) reported average values for the failure strength of spinal ligaments for all spinal levels. Ligaments were tested in situ by sectioning all elements except the one under study. Load was applied quasistatically (1 cm/s) until failure of the ligament occurred. Force displacement curves typically exhibited a sigmoidal shape. With respect to the cervical spine, data for the most important ligaments are reported. Yoganandan et al (1989) investigated the in situ dynamic response of the anterior longitudinal ligament and the ligamentum flavum of the cervical spine. Four different loading rates (9, 25, 250, and 2500 mm/s) were applied to obtain the (nonlinear) displacement-force curves up to failure. Both failure strength and stiffness (slope of the most linear part of the response) of the ligaments have been reported.

### 7.4.6 Two-Pivot Models

The global head motion or motion of the head relative to the torso, can already be described by relatively simple three-segment, two-joint models. In these two-pivot models, the neck is modelled as a rigid or extensible link that connects the movement of the torso at T1 to the head. Head and torso motion is determined from experimental sled tests with volunteers or cadavers. The experimentally obtained torso motion is used in the model to predict the head’s angular and linear displacement, velocity and acceleration. This head motion is then compared with the experimental head motion to validate the model. Pivot models have been developed by various authors, among others, by Bosio and Bowman (1986), Tien and Huston (1987) and Wismans et al (1987,1986). From their results it can be concluded that two-pivot models are indeed capable of simulating global head behaviour quite accurately. However, a two-pivot model suited for all impact directions and impact levels has not been developed yet.

The most important finding of the analysis of the human volunteer and cadaver tests is that the observed human head-neck response can be represented adequately by a linkage system with 2 pivots. One link represents the head, one link the neck and one link the torso. This analog system should not be considered as a representation of the very complex anatomical structure of the cervical articulations and the associated structures, however, it is rather a model which describes the head motions realistically in the case of non head impact conditions, assuming that T1 rotations can be neglected. Similar mechanisms could be developed if T1 rotations are not neglected. It should be noted that the use of these two pivot models is limited to the prediction of the kinematics of the head and do not usually enable the prediction of neck injury.

Torques calculated in this study near the occipital condyles and near T1 represent the resultant torque in the head-neck and neck-torso interface, respectively. These torques are developed by tension forces in the musculature, internal neck compression forces, etc. More detailed neck models are needed to study the load contribution of the various neck structures. From the cadaver tests conducted it has been learnt that muscle tone has an effect on the head-neck response: head rotations appear to be larger in the cadaver than in the volunteer tests. Furthermore significant differences in T1 rotations can be observed which are not completely understood yet. The observed injuries in the cadaver tests including two AIS 2 injuries indicate that the impact severity levels in the volunteer tests are quite high.

### 7.4.7 Three-Dimensional Discrete Parameter Models

The first three-dimensional discrete parameter models of the spine were developed by Panjabi (1973) and Belytschko et al, (1973). Both took the human spine to illustrate their general methods for the
construction of three-dimensional discrete parameter models and the determination of the governing equations of motion. The elements used are rigid bodies connected by deformable elements (springs and dampers). Chen (1973) developed a three-dimensional model of the human ligamentous spine suitable for use in both static and dynamic loading situations. Included are rigid vertebral bodies, deformable discs and posterior spinal elements (facet joints and ligaments), and the initial curvature of the spine. Chen reduced the model to two dimensions to analyse the pilot ejection problem.

Huston et al (1978) developed a head-neck model to predict head motion in impact situations. The model comprises nine rigid bodies, representing torso, cervical vertebrae and skull, connected by intervertebral discs, ligaments and muscles. Discs and muscles are modelled as (visco)elastic solids, ligaments as nonlinear elastic bands. Both muscles and ligaments exert force only in tension. One-way dampers are used to model joint constraints which limit the relative motion of the bodies. Tien and Huston (1985) simplified this model by taking an overall representation of the force-displacement behaviour of the soft tissue complex (disc, muscles and ligaments). They used empirical expressions for the forces and moments exerted by the soft tissues on the vertebrae. This yielded a computationally more efficient model with less parameters. Values for the stiffness and damping parameters were obtained from curve fitting of model prediction with experimental results of volunteer sled tests. The resulting model was further simplified by Tien and Huston (1987), who fused the cervical vertebrae into a single rigid body, which resulted in the two-pivot model mentioned earlier. Schneider et al (1989) added a moveable rigid jaw to the model of Tien and Huston (1985), to investigate the dynamic response of the jaw during whiplash (extension). The jaw-head joint allowed for both rotation and translation during jaw opening.

Sub (1977) describes a method to construct a dynamic model of the cervical spine. No quantitative data of the model are given, because not enough data on material properties were available at that time. Skull and vertebrae are modelled as rigid bodies. Ligaments and muscles in passive mode are modelled as nonlinear spring-dampers; and muscles in active mode as force generating elements. Facet joints are modelled with nonlinear spring-dampers that are compliant in tension and stiff in compression. To simulate disc behaviour, it is assumed that the overall effect of a complex (combined) displacement is the sum of the independent displacements for which force-displacement characteristics were measured.

Merrill et al (1984) extended the two-dimensional discrete parameter model of Reber and Goldsmith (1979) into three dimensions. The resulting model was further improved by Deng and Goldsmith (1985, 1987). This lumped parameter model of head, neck and upper torso comprises ten rigid bodies representing torso with T2, the vertebrae T1 through Cl, and the head. The overall mechanical response of intervertebral discs, ligaments and articular facets is lumped into a linear stiffness matrix, relating force and moment to translation and rotation. The off-diagonal elements of this matrix represent coupling of motion in one direction with load in another direction. Intervertebral damping is represented by linear damper elements. The model incorporates fifteen pairs of neck muscles, but only for the passive state. Muscles are represented by three-point spring elements with non linear constitutive relationships. To validate the model, numerical predicted head kinematics were compared to those obtained from frontal and lateral volunteer sled acceleration tests. Qualitatively, the response patterns were in reasonable agreement. Quantitatively however, correspondence is less good: especially the head accelerations which remain well below those from the experiments during the initial impact phase. Finally, Luo and Goldsmith (1991) extended the model of Deng and Goldsmith to include the lower torso. The model comprises ten rigid bodies representing the head; the vertebral pairs Cl-C2, C3-C4, C5-C6, C7-T1; the entire thorax; the lumbar vertebral combinations L1-L2, L3, L4-L5; and the pelvis.
7.4.8 Finite Element Models

Belytschko et al (1976, 1979) developed a three-dimensional finite element model of the head-spine-torso structure to study the pilot ejection problem. The model includes the complete spine, pelvis and skull and may also include the rib cage. Rigid vertebrae are connected by discs, ligaments and articular facets, which are represented by several deformable elements which collectively provide resistance against axial, torsional, bending and shear loads. Although a more detailed representation of the neck is available within the model, only simulations with a simplified (beam element) representation of the cervical spine are reported. The model has been advance recently by Privitzer and Kaleps (1990) to study the effect of head-mounted systems on the dynamic response of the head and spine.

Williams and Belytschko (1983) used the approach of Belytschko et al to develop a detailed head-neck model. The model comprises rigid vertebrae T1 through Cl and the skull connected to each other by deformable elements representing discs, facet joints, ligaments and muscles. Beam elements with linear torsional and bending stiffnesses and bilinear axial stiffnesses are used to represent intervertebral discs (C2-T1) and the connections between C2-Cl and C1-skull. Beams between C2-Cl and Cl-skull were arranged differently to account for the unique properties of this region. Articular facets are represented by a special shaped continuum element, with axial and shear stiffnesses. Ligaments are represented by nonlinear spring elements. Twenty-two different neck muscle groups are included. Muscles are represented by spring elements the axial force of which may be activated independently of the elongation to mimic muscle contraction. These elements include intermediate sliding nodes so that the muscles can curve around bones. The model was validated for frontal and lateral impact accelerations. Simulations in which the muscles are passive throughout the simulation and in which the muscles start contracting after some time are compared to the experimental results. For frontal impacts, predicted head kinematics agree with the experimental results, whereby the model with muscle contraction gives slightly better results than the passive muscle model. For lateral impacts, correspondence is less good, showing substantial deviations between numerical and experimental (maximum) head acceleration and displacement.

Hosey and Liu (1980, 1979) developed a three-dimensional finite element model of the head and neck. The model was developed primarily to study the mechanics of head (skull and brain) injury. It incorporates skull, dura, cerebrospinal fluid space, brain, jaw, cervical vertebrae and discs, and spinal cord. Each vertebra and each disc is modelled as a single element. Since the formulation is linear, the model is restricted to small displacements and rotations.

7.4.9 Validation

A model is validated through comparison of numerically predicted results of head-neck responses to impacts with similar results obtained from experiments. A complete and thorough validation of a detailed model should include a comparison of results on both the global and the local dynamics and kinematics of the head-neck structure in various impact situations. Global refers to the forces on the head, neck and torso and to the motion of the head relative to the torso, T1. Local refers to the forces acting on each cervical component at each vertebral level and to the motion of each vertebra.

For global validation, sufficient data are available on head-neck responses for various impact directions in which no head impact is involved, except for rear-end impacts (extension). Localized kinematics may be validated from the results of axial compression tests on cadaver specimens or from the quasistatic volunteer tests.

For global validation, the results of sled acceleration tests can be used. These tests have been performed with human volunteers and cadavers primarily to obtain the head-neck response to impact accelerations when direct head impact is not involved. Well known are the sled tests performed with volunteers at the Naval Biodynamics Laboratory (NBDL) in New Orleans, Louisiana, and with cadavers at the University of Heidelberg in Germany. Results of these and other sled tests have been

The overall dynamics (angular and linear accelerations) of the head and neck are obtained from accelerometers mounted to the subject's head and thorax (at T1), while the overall kinematics are obtained from high speed film-recordings taken during the impact. Volunteers have been subjected to moderate (non-injurious) impact levels, whereas cadavers have been subjected to moderate - for comparison with volunteer-tests - and severe (injurious) impact levels. Accelerations may be applied in several directions: frontal, lateral, oblique, rear-end and vertical (pilot ejection). In the NBDL tests, human subjects were exposed to short duration accelerations which simulated frontal, oblique or lateral impacts. There are fewer experimental results from rear impact sled tests involving volunteers which give information about the global dynamics and kinematics of the head and neck. This is because of the vulnerability of the neck for rear-end impacts and hence the greater risk of injury in such tests. However, recent work in this field has been reported by Davidsson et al (1998), Ono et al (1998), Eichberger et al (1996 & 1998) and Wheeler et al (1998).

For local validation, a detailed knowledge of the dynamic and kinematic response of the human cervical spine is needed. Most of this knowledge cannot be determined experimentally from volunteers and is even hard to obtain from experiments with cadaveric material. As a consequence only a few experiments that may give such results have been reported in the literature. For quasistatic and dynamic axial compression of the straightened cervical spine, Yoganandan, Pintar and co-workers (1990,1989,1991) obtained the sagittal plane movements of vertebrae and occiput from film-recordings of markers placed in these bony parts. Detailed results on these movements are given. Studies that may give the localized movement of vertebrae in quasistatic, voluntary (muscle induced) motion of the head include the following: Moffat and Schultz (1979) and Van Mameren (1988) conducted X-ray studies on the sagittal plane motion (patterns) of cervical vertebrae for voluntary flexion and extension movements. Margubes et al (1992) used magnetic resonance imaging to measure the in vivo motion of the cervical spinal cord in human volunteers for stepwise flexion and extension of the neck. From these images vertebral movements can be obtained.

7.5 A MODEL FOR LOW-SEVERITY REAR-IMPACTS

Several mathematical models of the human spine exist: Ome and Liu (1971) developed a 2-D model of the human spine which incorporated the axial, shear and bending resistance of the discs, and modelled each vertebra as a rigid body with three degrees of freedom. This model was later adapted by Prasad and King (1974) and validated by comparison with cadaver drop tower tests (loaded in a vertical direction). Belytschko and Privitzer (1979) developed a 3-D model in which the human body is represented by a collection of rigid bodies interconnected by deformable elements. This model was validated by comparing its vertical impedance with experimental measurements by Vogt et al (1968). None of these models were validated for a (horizontal) loading situation similar to a rear-end or frontal impact.

The objective of van den Kroonenberg’s et al 1997 study was to develop a mathematical model of a seated occupant and to better understand the biomechanical response of the spine and the occupant's interaction with the seat during rear-end collisions. For this purpose, a 3D mathematical model of a 50th percentile sitting adult male was developed for use in simulations of rear impacts. Special attention was paid to the modelling of the spine, including the neck, and the occupant's interaction with the seat. To obtain an insight into its biofidelity, the model's response was compared with rear-end sled tests using volunteers and human cadavers at a deltaV of up to 30 km/h. The model was then used to study and quantify the motion of the spine in low and medium severity rear-end collisions. This study revealed that, during the "torso loading phase", the pelvis was lifted from the seat while the vertical motion of the T1 vertebral body relative to the vehicle was slight. Spinal compression occurred during this phase, but it remained slight. Although a thorough validation of the model developed was not possible due to lack of experimental data, van den Kroonenberg's mathematical
model of the lumbar and thoracic spine was developed and added to an existing rigid body model of the neck and head.

The model that probably comes closest to the one presented by van den Kroonenberg et al is the spine model developed by Jakobsson et al at Chalmers University (1994), in which the human spine is represented by 24 rigid bodies interconnected by hinges. The geometry of the spine elements was obtained from Robbins (1985). The spine model was implemented in the 50th percentile sitting male Hybrid III dummy database developed by TNO (1996). Non linear rotational stiffnesses and damping coefficients were defined for each joint. Van den Kroonenberg et al admit that although their study represents a major step forward in the study of head/neck responses during low severity rear-end collisions it still has some limitations. For example, the arms and shoulders are not modelled as separate bodies, i.e. the mass of the arms and shoulder was added to the mass of the spine. Also, the validation of this model was mainly qualitative and was done by applying a rear-end collision pulse to the model, similar to the pulse used by McConnell et al (1993), (deltaV = 8 km/h), and comparing the resulting head angles. However, the authors indicate the curvature of the upper part of the spine appears to differ between the model and the volunteer in McConnell's study. This may affect the validity of the interaction between the seat-back and the back of the occupant model.

The model consists of 3 components: a thoracic+lumbar model representing the human spine up to T2; a head-neck model (starting with T1); and finally, the remaining body parts (such as arms and legs). The starting point was a validated MADYMO model of a 50th percentile sitting adult male Hybrid III dummy. In this model, the bodies and joints representing the entire spine and the head were replaced by a more detailed representation as reported by van den Kroonenberg et al (1997).

The complete model's performance in a rear impact situation was compared with several rear-end sled experiments at low and high velocities with volunteers and cadavers sitting in a seat with a rigid back support - Hoofman et al(1997). After considering the overall kinematics of the model, the head angle and the vertical displacement of T1 are considered. The static performance of the spine was also investigated. The experiments were simulated by placing the human model on a rigid seat, and applying the appropriate pulses. Three different pulses were used with values for sled-deltaV ranging from 6 to 30 km/hr.

Van den Kroonenberg et al (1997) reported on their results as:

- **Spine deformation** - For the low severity runs, no contact with the head restraint occurred. The deformation of the spine during a simulation is divided into two phases: during the first phase which was called the "torso loading phase", T1 is moving backward with respect to the vehicle; the second phase, the so-called "torso rebound phase", starts as soon T1 begins to move forward. A general observation is that during the torso loading phase the spine is pushed back further as the level of severity of the pulses increases. Also, the head-neck motion is more pronounced in the cases without a head restraint compared to those with a head restraint.

- **Joint rotations** - A comparison of the initial spine configuration with the final configuration during the "torso loading phase" reveals that extension occurs in the lumbar and thoracic spine, and that flexion takes place only at the midthoracic level. Also, the maximum extension occurs in the lumbar joints.

- **Axial displacements** - The elongation of the spine from the first sacrum level, S1 to T1 during the loading phase is negative (compression) and ranges between -3 mm for the low severity cases up to about -12 mm for the highest pulse. For the medium and highest pulses, the elongation became positive during the torso rebound phase. No clear effect of the presence of a head-restraint on spine elongation could be detected.

- **Displacement of T1** - The vertical displacement of T1 during the torso loading phase ranges between 1 and 18 mm upward and was larger for the cases without a head restraint compared to
those with a head restraint. Finally, during the torso rebound phase the vertical position of T1 increased markedly, in particular in the higher severity cases.

- Displacement of pelvis - The vertical displacement of the pelvis increases from about 2 mm in the low severity case to more than 5 cm for the higher pulse representing elevation from the seat. An effect of the head restraint on the vertical pelvic displacement was not found.

- Head rotation - The maximum head angle was larger for the cases without a head restraint compared with those with a head restraint. Also, for the cases without a head restraint, the maximum head angle increased with the level of severity of the pulses.

The predictive ability of the model developed can be assessed by considering the head restraint contact forces and the shear forces at the occipital joint. The predicted values for these parameters can be compared with values obtained from experiments with volunteers conducted by Mertz and Patrick (1967). As expected, the maximum head restraint contact forces increase with the severity of the pulse. When the maximum head restraint contact force is plotted against the horizontal distance between the head and the head restraint, good correspondence with experimental data is obtained. Regarding the occipital shear forces, the maximum shear forces increased with the level of severity. Also, the shear loads were lower when a head restraint was used, compared with the cases when no head restraint was present, suggesting that head restraints reduce this type of load. Finally, the predicted values for the occipital shear forces correspond well with the experimental data.

Although the current study focused on the response in the mid-sagittal plane, and therefore only considered 2 degrees of freedom, in principle the complete model is three-dimensional and could be used for oblique impacts as well.

A limitation of this study is that due to lack of experimental data from rear-end sled tests, the human model developed could not be thoroughly validated. However, by considering the overall kinematics of the model and comparing the head angle and the T1 displacements with experimental data, encouraging results were produced. Despite the limitation mentioned above, this study represents a major improvement in the field of human body response modelling during rear-end collisions. However, to enable more complete validation and to make this model more effective, more experimental data on rear-end sled test volunteers and cadavers is needed.
7.6 DISCUSSION AND CONCLUSIONS

The goal of mathematical modellers in this field of study is the development of a model representing the mechanical behaviour of the human spine to describe the biomechanical response of the human head and neck to various impact situations. Such a model has to incorporate injury mechanisms and describe the local kinematics and dynamics of individual vertebrae. From this review the conclusions are:

Material characteristics - There is adequate detail of the geometrical and inertial characteristics in the literature. However, quantitative information on the mechanical behaviour of cervical components and motion segments remains incomplete though much progress has been made in recent years. For some cervical spine components material characteristics are not available at all, while for other components data on the material characteristics are incomplete with respect to types (quasistatic, dynamic) and directions of loading. Experiments have been conducted to obtain material characteristics for both upper and lower cervical spine segments. With respect to lower cervical spine segments, experiments were conducted under either static or quasistatic loading. Only in a few of these studies were loads applied up to failure of the specimens. With respect to the upper cervical spine, data from static experiments (with low maximum load) for various types and combinations of loading have been reported. Results for quasistatic and dynamic applied loads (up to failure) have been reported for axial rotation.

Mathematical models - A number of sophisticated models describing head-neck dynamics have been reported in the literature (Jager et al 1993). These include the discrete parameter models of Deng and Goldsmith, and Tien and Huston and the finite element model of Williams and Belytschko, which is the most detailed model. These models have been validated, but only for a small number of impact situations and only for global (head) kinematics. Model predictions were compared to the results of volunteer sled tests. In general, the numerical and experimental response patterns showed good correspondence. Quantitatively, correspondence was usually less good and substantial deviations were reported. It should be noted that only a very limited amount of physical property data on cervical (motion) segments was available at the time the models were developed. Properties were either tuned until the model showed reasonable behaviour or were estimated from data obtained from non-cervical components. More accurate data may improve the model predictions.

When the performance of these detailed models is compared to the performance of two-pivot models, it appears that the detailed models do not simulate the global movements more accurately than the two-pivot models. Thus, detailed models and pivot models seem to be equally capable of simulating the global kinematic behaviour of the head-neck system. However, two-pivot models are of limited use in the investigation of injury mechanisms and cannot simulate the local behaviour of individual vertebrae.

In conclusion - two important problems in mathematical modelling of the mechanical behaviour of the spine are:–

- the incompleteness of experimental data on the physical properties needed to develop a detailed model, and
- the incompleteness of experimental data needed to validate a model thoroughly.
7.7 FUTURE SPINAL MODELLING WORK AT TRL

Mathematical modelling of the human spine will play a very important role in the future of the current project. A computer model of the human thoracic and lumbar spine has already been developed at TRL as part of a long term ABlue Skies research programme sponsored by the Science and Technology Policy Division of DETR. This model is capable of replicating the kinematic behaviour of the spine and predicting some of its more common injury mechanisms. The model will be used to investigate what happens to the spine in a vehicle impact and how spinal injuries are caused. One possible future course of research might be to obtain a mathematical model of the human neck to link to the existing thoracolumbar spine model. This would enable further investigations of the mechanisms which cause whiplash injury and other more severe neck injuries to be made.

TRL also has use of a computer model of the Hybrid III dummy which has been produced jointly by First Technology Safety Systems Plc (manufacturer of the dummy) and Ove-Arup (the UK distributor of DYNA3D finite element software). It is planned that the human spine model will be incorporated into this Hybrid III model so that a comparative study of the performance of the human and existing Hybrid III spine under impact conditions can be made. It is expected that this exercise will indicate possible designs for an improved dummy spine.

7.8 REFERENCES

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Dummy Development: Spinal Injuries


Dummy Development: Spinal Injuries


Dummy Development: Spinal Injuries


8 THE HYBRID III DUMMY SPINE AND ITS LIMITATIONS

8.1 SUMMARY

The occurrence of spinal injuries, particularly whiplash type injuries, in road traffic accidents are significant. The Hybrid III dummy is the standard dummy which is used in determining injury, however, it may not be sufficiently biofidelic to predict the type of injuries seen in low speed rear impacts. Certainly, there appears to be no dummy which can provide the measurements for prediction of whiplash injuries.

This section of the literature review begins with a short history of the Hybrid III dummy and then goes on to describe each part of the Hybrid III dummy spine before exploring it’s limitations.

The literature reviewed indicates that the vehicle occupant is usually relaxed at the time of a rear impact and that the Hybrid III neck has the behaviour of a tensed occupant. The neck is too stiff during low severity rear impacts and the maximum head angles were too low to represent a relaxed person. Neck flexion occurs earlier in the Hybrid III than it would in a cadaver. Also the neck injury tolerance limits may need updating.

The review suggests that the neck and spinal structure of the Hybrid III is unlikely to interact with the seat-back in the same compliant way as the human spine would. The Hybrid III is less prone to ‘ramp up’ along the seat back than a vehicle occupant would be. It is suggested that this could be because the structure of the Hybrid III thorax is too stiff.

The literature documenting Hybrid III lumbar spine suggests that it has significantly greater initial stiffness than cadaver specimens. The Hybrid III dummy lumbar spine has a high shear stiffness at lower load levels and does not have a comparable performance until it experiences loads greater than 500N. Shear loads greater than 200N result in complete failure of the posterior ligaments. Anterior and posterior longitudinal ligaments become stretched. Disc injuries would include splitting and detachment from bony end plates. The Hybrid III lumbar spine does not appear to be suitable for assessing soft tissue damage.

8.2 INTRODUCTION

The Hybrid III is a fiftieth percentile, adult male crash test dummy that was developed by General Motors Corporation in the 1970s to improve both the biofidelity and injury prediction, in motor vehicle impact tests, of existing dummies. This started in 1972, when General Motors developed an advanced anthropomorphic test dummy under a contract with the National Highway Traffic Safety Administration. This dummy, called the GM-ATD 502, was completed in 1973 and is described in the paper by Tennant (1974). The dummy featured a head design that mimicked the geometric, inertial and forehead impact response characteristics of an average sized adult male. The shoulder structure was designed for improved fidelity of shoulder belt interaction. A curved lumbar spine was used to resemble human-like automotive seating posture. The limb joints were designed for ease of setting at a constant 1 G frictional resistance. The dummy was a significant improvement in crash dummy technology, but it was not truly human-like since contract deadlines precluded incorporation of the biofidelic neck and chest structures that were still being developed. General Motors initiated an internal project to incorporate these biofidelic components in 1974 (Foster et al) and hence the Hybrid III was developed.
The Hybrid III became commercially available in 1977 and has undergone further developments to incorporate load transducers to measure the internal reaction’s between the lumbar spine and the pelvis and between the neck and the thoracic spine.

In 1983 General Motors Corporation petitioned the National Highway Traffic Safety Administration to allow the use of the Hybrid III as an alternative test device for FMVSS 208 compliance testing. As part of the petition, GM provided a set of Injury Assessment Reference Values (Mertz, 1984). NHTSA responded favourably to the GM petition and the Hybrid III was incorporated into Part 572 of Federal Motor Vehicle Safety Standards in 1986. On November 13th 1990 GM petitioned NHTSA to make the Hybrid III the only dummy allowed for FVMSS 208 compliance testing. There is now a world-wide use of the Hybrid III dummy in new vehicle development programmes.

This section reviews the limitations of the Hybrid III dummy spine. The occurrence of spinal injuries in road traffic accidents are significant, and the cost of whiplash injuries alone has been estimated at about two and a half thousand million pounds, i.e. eighteen percent of the total cost of UK road traffic accidents in 1991. The Hybrid III dummy spine might not be sufficiently biofidelic to predict the type of injuries seen in rear impacts. Certainly, there is no dummy which can provide the measurements for prediction of whiplash injuries. The dummy must have good biofidelity in order for its responses (trajectories, velocities, accelerations, deformations and forces) to be representative of responses of a human exposed to the same test environment (Mertz 1984). The dummy must also be instrumented to measure responses that can be associated with various types of occupant injuries so that the likelihood of such injuries may be predicted in standard crash tests. The neck measurements with the Hybrid III will be misleading if the thoracic and lumbar spines are not sufficiently biofidelic.

### 8.3 CERVICAL SPINE

#### 8.3.1 Hybrid III Dummy Neck

The Hybrid III dummy neck (figure 8.1) is a one-piece, flexible component with biomechanical bending and damping responses in both flexion and extension. Three rigid aluminium vertebral elements are moulded in a 75-durometer (shore hardness) butyl elastomer to form the neck structure (Mertz 1972).

The butyl elastomer was chosen for its high damping characteristic in order to approximate the biomechanical hysteresis requirements. Aluminium end plates were moulded into the neck to provide attachment surfaces. A single steel cable runs through the centre of the neck to provide a high level of axial strength. The neck is fabricated by the injection of the elastomer into a neck mould, which provides a cavity to define the neck geometry, and locates properly the three vertebral elements and two end plates. An asymmetrical cross sectional geometry was selected to provide higher bending resistance to flexion than extension. Saw cuts on the anterior half provide an additional reduction in extension bending resistance without affecting flexion response. The design has an adjustable neck bracket to enable levelling of the head when the dummy is seated in a vehicle. A transducer is attached to the neck and head through a nodding joint. The transducer is used to measure axial and shear loads, and moment about the occipital condyle.

Figure 8.1 Hybrid III dummy neck. (Reprinted with permission, SAE 770938)
8.3.2 Hybrid III Dummy Neck Performance

8.3.2.1 Rear Impact
During an evaluation of the Hybrid III dummy neck, Thunnissen et al (1996) stated that in rear impacts 74% of whiplash patients did not anticipate the accident and were therefore relaxed at the time of impact. Thunnissen suggests that the Hybrid III dummy neck is too stiff for low severity rear impacts since the maximum head angles recorded in tests, were too low to represent a relaxed person.

Thunnissen quoted literature showing the Hybrid III has a lack of rear impact biofidelity. The points made were that it had too high resistance to bending of the neck and the torso at low severity. He refers to the work done by Foret-Bruno et al (1991) and Scott et al (1993) when suggesting the need for a new low severity rear impact testing dummy. He reports on work done by Mertz and Patrick (1971), which was the basis for the Hybrid III dummy neck, where human volunteers were subjected to static and dynamic environments which produced noninjurious neck responses for neck extension and flexion. Cadavers were used to extend the data into the injury region. Mertz and Patrick stated that the angle between the head and the torso is not a good physical measurement of trauma for injuries resulting from neck extension or neck flexion. They created head and angle-moment of force corridors for neck flexion and extension (figure 8.2) and suggested tolerance levels for hyperextension. The non-injurious level was defined at 47Nm with a maximum voluntary head rotation of 75 degrees. The ligamentous damage level was defined at 57Nm. It was noted that size, weight, age and sex would affect the levels.
Mertz and Patrick believed that hyperextension was the cause of whiplash injury. More recent work (e.g. Svensson et al) suggests that injury occurs earlier in impact, so the early tolerance limits may well be wrong.

Thunnissen concludes that the response of the Hybrid III dummy has been evaluated with the maximum head and angle-moment of force corridor and it was shown that the Hybrid III neck is too stiff for low severity rear impacts if the occupant is unaware of the impact. The Hybrid III tends towards the response of a vehicle occupant, whose neck muscles are tense in the expectation of an impact.

The injury assessment values that are used to evaluate the Hybrid III neck response measurements were discussed by Mertz (1984). Mertz defines injury assessment value as the human response level below which a specified significant injury is considered unlikely to occur for a given individual (e.g. Head acceleration, neck loads, chest acceleration, etc.).

In neck bending the neck flexion moment ($M_f$) has an injury assessment value of 190Nm. The implications of the value being that if the neck flexion moment is less then 190Nm, then significant neck injury due to flexion bending is unlikely if the neck is hyperflexed. The neck extension moment ($M_e$) has an injury assessment value of 57Nm. . The implications of the value being that if the neck extension moment is less than 57Nm, then significant neck injury due to extension bending is unlikely.
if the neck is hyper extended. (Mertz states that there was very limited data for this injury assessment value). For neck tensile, compressive and shear forces, injury assessment is defined by time dependent curves. The curves in figures 8.3, 8.4, and 8.5 define axial tensile neck loading, axial compressive neck loading and fore/aft shear force respectively. The implications of the curve values are such that if the force levels for all duration’s lie below their respective curves, then significant neck injury due to those types of neck loadings are considered unlikely.

![Graph of Injury Assessment Criteria for Axial Neck Tension](image1)

**Figure 8.3** Graph to show the Injury Assessment Criteria for axial neck tension

![Graph of Injury Assessment Criterion for Axial Neck Compressive Loading](image2)

**Figure 8.4** Graph to show Injury Assessment Criterion for axial neck compressive loading
Mertz states that there are limitations to the injury assessment values when interpreting the responses of the dummy. Many of the injury assessment values are based on limited biomechanical data, and therefore, are not constant and in many instances are unknown. Not all significant injuries sustained by car occupants are comprehended by the injury assessment values.

Foret-Bruno et al (1991) carried out a comparative test on a cadaver and the Hybrid III dummy under very severe crash conditions (speed 65km/h, deceleration distance 930mm, acceleration (average) = 18g). The seat had a stiffened back and a stiff head restraint in contact with the head. The head movement with relation to the thorax was almost non-existent in both tests. The autopsy of the cadaver showed no injury to neck tissue or ligament. However, in the Hybrid III test high values of shear force and torque were recorded. The shearing force reached 590N and the torque measured 18Nm. Foret-Bruno states that this level of shearing force experienced by the Hybrid III is suggestive of a high risk of cervical injury. This statement contradicts Mertz’ injury assessment values.

Foret-Bruno goes on to suggest that without any relative movement between the head, the neck and the thorax there is no risk of injury and thus the shearing force measured at the neck load measurement device level does not signify a risk of injury when there is violent contact between the head and the head restraint. In this case the large shear force arises from Newton’s 3rd law (every action has an equal and opposite reaction). The strong force exerted by the head restraint on the head leads to a proportional reaction force measured between the head and the neck. He states that under these conditions, only the torque and angle of head rotation in relation to the thorax should be used as injury criteria when there is head contact. This is because the biofidelity of the Hybrid III in extension is only based on tests with cadavers without head support. Furthermore, dummy neck design is such that any relative movement of the head in relation to the neck, however small, leads directly to shearing forces. The dummy does not have, as the human spinal column has, the possibility of natural movements which do not lead to stress.

Lövsund and Svensson (1996) reported that the neck and spinal structure of the Hybrid III dummy is stiff and unlikely to interact with the seat-back in the same compliant way as the human spine would. They reported that Seemann et al (1986) found the Hybrid III neck far too stiff to respond in a human like manner in the sagittal plane (the sagittal plane is parallel to the join between the two parietal bones, which form the top and sides of the skull). Deng (1989) found that results from a mathematical model of the Hybrid III neck indicated that the neck has a torque response similar to that of a human.
neck but has a higher shear response. They recounted that in volunteer tests McConnell et al (1993) found that during the acceleration phase of a rear-impact, when the occupant’s body was pressed against the seat-back, the spinal curvature straightened. This in turn caused an upward motion of the head and thus an elevated head contact point on the head restraint. They also reported that in a comparative study using volunteers and a Hybrid III, Scott et al (1993) found that the dummy was less prone to ‘ramp up’ along the seat back than were the volunteers.

8.3.2.2 Frontal impacts

Kallieris et al (1992) used human cadavers and Hybrid III dummies in crash simulations of frontal collisions, in order to investigate the biofidelity of the Hybrid III dummy. A total of eleven frontal collision tests were undertaken using a 50th Percentile Hybrid III dummy and six cadavers. The simulations were performed at an impact velocity of 50 km/h. The cadavers and Hybrid III dummies were protected with either a three-point seat belt or an air bag-knee bolster system.

The overall kinematics of the Hybrid III and cadaver were comparable. But there were temporal and spatial differences in head/neck motion. Translational movement of the head occurred for the first 40ms after the start of the impact in both the cadaver and Hybrid III tests. Thereafter, head rotation was observed which was caused by the restraining effect of the belt. The magnitude of the head/neck flexion was greater with the cadaver and maximum flexion occurred later than was observed with the Hybrid III. Except for one laceration of the ligamentum flavum at T1/T2 (C3, AIS 2), no vertebral column injuries were observed in the investigated cadavers. Ligamenta flava contain elastic tissue and connect the laminae of adjacent vertebrae from C1 to S1. A possible reason for the greater neck flexibility observed in cadavers is that the upper thoracic vertebral column is included in the bending process. This may also account for injuries observed at the transitional zone between the thoracic and vertebral spines.

The authors also commented on previous work where the Hybrid III dummy neck stiffness was found to be greater than that of a cadaver’s neck. The results of padded and rigid Hybrid III and cadaver head impacts, at an impact speed of 20 and 22 km/hr, had shown that the maximum cadaver head/neck extension was about 30% greater than Hybrid III extension. The maximum bending angle occurred earlier with the Hybrid III than the cadaver. This was also observed in 25km/h rear-end collisions.

8.4 THORACIC SPINE

8.4.1 Hybrid III Dummy Thoracic Spine

The Hybrid III dummy thoracic assembly consists of a spine and rib cage covered by a removable chest jacket (figure 8.6). The whole assembly is balanced for correct weight and centre of gravity location. The thoracic spine is a welded steel box construction and provides for attachment of the neck, clavicles, ribs and lumbar spine.

The steel box houses a triaxial accelerometer mount, located at the assembly centre of gravity. A rotary potentiometer is attached to the top of a bracket which extends over the lumbar spine. A rod extends from the transducer to the sternum, to provide input to the potentiometer.

Figure 8.6 Hybrid III dummy thorax

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8.4.2 Hybrid III Thoracic Spine Performance

The injury assessment values that are used to evaluate the Hybrid III chest measurements were discussed by Mertz (1983). The injury assessment value for the resultant chest exceedance acceleration is a 3ms exceedance value of 60g. This implies that if the resultant chest acceleration does not exceed 60g for more than 3ms then significant thoracic organ injury due to gross chest acceleration is unlikely. This criterion does not appear to address spinal injury.

Kallieris (1992) conducted frontal crash simulations with human cadavers and Hybrid III dummies in order to make an assessment of the biofidelity of the Hybrid III based on the kinematic behaviour of the thoracic responses. The measured thoracic accelerations at T12 were in agreement but differences were found for accelerations measured at the level of T1 and sternum. The accelerations of the cadaver thorax were less than those of the dummy thorax, and the measured thoracic deformations were greater with the cadaver’s thorax. In the anthropometrically matched pairs the thoracic deformations of the cadaver were twice that of the Hybrid III, which suggests that the Hybrid III trunk articulations may be stiffer than the equivalent parts of the cadaver.

Shoulder belt forces depend on body mass and tissue deformability in frontal collisions. Clear differences between the dummy and the cadavers were observed even for anthropometrically matched pairs. Higher belt forces were observed for the dummy, and as the masses were similar, this also indicates that the dummy thorax is stiffer than the cadaver.

Kallieris went on to suggest that in frontal collision simulations, the biofidelity of the dummy neck and thorax should be improved to allow for more valid assessment of occupant protection criteria.

No references discussing the performance of the Hybrid III dummy thorax, during rear impact, were found.

8.5 LUMBAR SPINE
8.5.1 Hybrid III Dummy Lumbar Spine

The Hybrid III dummy lumbar spine is a moulded, curved polyacrylate elastomer member with moulded-in end plates for attachment to the thoracic spine and pelvis (figure 8.7). Two cables pass through the lumbar spine and attach to the end plates. The cables provide lateral seating stability while permitting fore and aft flexibility. The curved lumbar spine allows the dummy to assume a slouched position when placed on a vehicle seat. Thus providing a more human-like seated posture and a more realistic eye location.
8.5.2 Hybrid III Dummy Lumbar Spine Performance

Begeman et al (1994) carried out a study of the viscoelastic shear responses of the cadaver and Hybrid III lumbar spine. Functional lumbar spinal units and a complete lumbar section were tested in both anterior and posterior directions. Similar tests were performed on the Hybrid III lumbar spine for comparison. All of the musculature and extra soft tissue were removed from the spines. Various bony parts of each vertebra were reinforced with sheet metal screws before embedding in quick setting epoxy blocks.

Comparative testing was performed on the complete lumbar spine specimen (T12-L5) and the Hybrid III spine. There were significant differences in lumbar stiffness and it was concluded that a muscle-free in vitro model could not be used for the design and/or validation of the lumbar spine construction of an anthropomorphic test device. For this reason there were no more multilevel tests.

Sixteen lumbar motion segments were tested quasi-statically for their viscoelastic properties in a multidirectional (5-axis) spine machine. The specimens were loaded to sufficient displacement to cause hard or soft tissue failures. Before loading, the initial centre of shear was determined experimentally by an iterative technique (The point of load application was manually altered until no rotation of the vertebral body could be measured less than 0.3 degrees). The centre of shear was usually located within the intervertebral disc. This was done in order to isolate the forward and rearward shear stiffness from associated rotational and translational stiffness.
For the single-level testing the total length of the Hybrid III dummy spine was subdivided into six levels, so that one-sixth of the lumbar spine would represent a functional lumbar spinal unit (functional spinal unit is defined as two adjacent vertebrae and all interconnecting ligamentous soft tissue). This length corresponded to 40mm of the rubber spine. The shear loading was applied parallel to the “mid disc plane” defined at the lower one sixth of the arc-length of the composite column, orthogonal to the metal cables.

In the quasi-static single level tests Sagittal shear force was recorded against translation. A comparison of the measured average in vitro shear stiffness with corresponding data from the single level test of the Hybrid III dummy showed that the dummy spine had a significantly higher stiffness at lower load levels.

Dynamic cyclic loading consisted of imposing a displacement of 1.5mm over a set time period and then holding the sample at the displacement for 300s whilst recording the force applied to the specimens. The peak loads for the Hybrid III in this single level test were 90% -200% larger than the corresponding loads for the cadaver functional spinal units. This implies that it takes 90% - 200% more force to displace the Hybrid III spine and hence it is a lot stiffer than the human lumbar spine.

The cadaver specimens were also tested to destruction. After an initial period of translation during which forces up to 200N were applied, the facet of the superior vertebra slipped over the inferior facet. Both facets remained intact but started a sagittal rotation resulting in complete failure of the posterior ligaments i.e. interspinous (which is the ligament which connects the individual spinous processes), supraspinous ligaments (which is a continuous ligament directly behind the interspinous ligaments) and ligamentum flavum (which are the ligaments which connect the individual lamina). Anterior and posterior longitudinal ligaments were stretched. Disc injuries included splitting and detachment from bony end plates. It was seen that soft tissue failure loads were dependent on failure rate.

Hybrid III spine showed significantly greater initial stiffness than cadaver specimens, but had a comparable performance at shear for loads greater than 500N.

Cadaver anterior lumbar failure in shear started at 1200N and depended on loading rate as well as test set-up.

These results suggest that the Hybrid III lumbar spine is not suitable for assessing soft tissue damage.

8.6 CONCLUSIONS

NECK

Rear Impact

- When compared with volunteers, the Hybrid III dummy is less prone to ‘ramp up’ along the seat back (Scott et al 1993).
- In a rear impact accident the vehicle occupant is unlikely to anticipate the crash, and will therefore be relaxed at the time of impact. The maximum head angles achieved by a Hybrid III dummy are markedly lower than those of a relaxed person in a rear impact (Thunnissen et al 1996).
- Injury occurs early in impact, so the tolerance limits, from earlier work by Mertz and Patrick (1971), may not be applicable, for low severity injuries (Thunnissen et al 1996).
- The Hybrid III neck is too stiff during low severity rear impacts ((Thunnissen et al 1996) (Lövsund and Svensson 1996)).
- The neck and spinal structure of the Hybrid III is unlikely to interact with the seat-back in the same compliant way as the human spine would (Lövsund and Svensson 1996).
Frontal Impact
- When compared with cadavers, in frontal impacts, the magnitude of the head/neck flexion of the cadaver is greater than that of the Hybrid III (Kallieris 1992).
- Maximum flexion in cadavers occurs later than that observed with the Hybrid III (Kallieris 1992).

THORACIC
- The Hybrid III thorax is stiffer than that of an anthropometrically matched cadaver (Kallieris 1992).

LUMBAR
- The Hybrid III dummy lumbar spine has a significantly higher shear stiffness at lower load levels than an in vivo spine (Begeman et al 1994).
- Hybrid III spine shows significantly greater initial stiffness than that of cadaver specimens, and has a comparable performance at shear, only for loads greater than 500N. However, during destructive testing of cadaver spines, shear loads greater than 200N result in complete failure of most of the posterior ligaments, with the anterior and posterior longitudinal ligaments being stretched. Disc injuries include splitting and detachment from bony end plates (Begeman et al 1994).
- The Hybrid III lumbar spine appears to be unsuitable for assessing soft tissue and hard tissue damage (Begeman et al 1994).

8.7 REFERENCES


9 RECENT DUMMY DEVELOPMENTS

9.1 THE CHALMERS REAR IMPACT DUMMY (RID) NECK

Svensson & Lövsund (1992) describe the early development of the RID neck, which is specifically designed to give a better simulation of the shearing behaviour of the human neck, where, under inertial loading, the head can translate backwards (or forwards) relative to the torso without undergoing rotation. The new neck consists of seven cervical elements, each of which can rotate by up to 10° in extension and 5.6° in flexion relative to the element below it. Two thoracic vertebrae are also included, the lower of which is fixed to the dummy thorax, while the upper can rotate by up to 3° relative to the lower in either flexion or extension. Figure 9.1 shows the basic layout.

![Figure 9.1: The Chalmers RID Neck](Reproduced from the International IRCOBI Conference Proceedings, Svensson & Lövsund (1992))

In the seated posture, the neck is flexed 14°. The total angular range of motion of the head is 83° in extension and 48° in flexion. Data on the voluntary range of motion of the head/neck complex vary widely, depending on the source of the estimate. The chosen range of motion was thought to represent a reasonable average, and some extra movement was built in so that the neck could go into hyperextension without the elements bottoming out against each other. Validation testing confirmed that the RID neck response was still within the response corridor suggested by Mertz & Patrick (1971). Comparison with volunteer tests showed that the kinematic response of the RID neck simulated the human response much better than did the Hybrid III neck.

Subsequent use of the RID neck in experimental work confirmed that it was a big improvement over the standard Hybrid III neck, but that there was still room for improvement (Geigl et al, 1994). Svensson et al (1993) state that the RID neck may still be too stiff.

Subsequent development work on the RID neck has been undertaken by TNO in the Netherlands, and is described by Thunnissen et al (1996). Two problems with the RID neck were that the dummy chin could contact the neck during rebound, and the articulating elements could grab the head restraint material. Either of these could compromise the kinematic response of the head/neck complex. The result of the development work is the TRID II neck, in which the number of articulating elements has...
been reduced to seven. The problems mentioned above have been addressed, and rubber buffers have been added between the elements to give a ‘soft landing’ as they approach the bottoming out condition. This also prevents possible permanent deformation of the compressible elements between the ‘vertebrae’. The new design is said to give improved reproducibility compared to the original and to agree well with published head/neck rotation and other parameters. However, they point out that head/neck kinematics in rear impacts are heavily influenced by bending of the thoracic spine, and that what is needed is a biofidelic model of the complete spine, so as to facilitate the testing of complete seats, rather than just the head restraint. In particular, the straightening of the human spine under rear impact loading (McConnell et al, 1993) and, to a lesser extent, the ramping of the torso up the seat back (Ono & Kaneoka, 1997, van den Kroonenberg et al, 1997), both of which result in a higher point of contact between head and restraint, are not well reproduced by the Hybrid III dummy spine.

9.2 THE SWEDISH REAR IMPACT DUMMY - ‘BioRID’

The most recent anthropomorphic test device for assessing the performance of car seats and head restraints was launched at the IRCOBI conference in Gothenburg, September, 1998. It is a new biofidelic rear impact dummy - BioRID (Davidsson et al, 1998). The development of this new dummy was part of the Swedish Vehicle Research Program and was conducted and supported by Autoliv AB Research, Volvo Car Corporation, Saab Automobile AB and Chalmers University where most of the technical work was done.

BioRID is essentially an extension of the original segmented Rear Impact Dummy-neck (RID-neck) developed and validated by Svensson and Lovsund (1992). This neck was designed to be used on a Hybrid III dummy in rear end collisions at low impact velocities. It consists of seven cervical and two thoracic vertebrae. Thunnissen et al (1996) modified the RID neck by reducing the number of pin joint segments from nine to seven and hence produced the TNO Rear Impact Dummy-neck or TRID-neck. The TRID-neck was developed to give improved reproducibility and repeatability which was a weakness of the RID neck. Both necks have a very similar dynamic response to rear impacts. The two neck types have been described previously in chapter 3 and they both have a segmented hinged structure similar to that of the neck in BioRID.

Figure 9.2 shows a schematic diagram of BioRID. The dummy has a new torso, arm attachments, articulated spine, neck muscle substitutes and pelvis, and has been fitted with Hybrid III legs, arms and head. The BioRID spine consists of 7 cervical, 12 thoracic and 5 lumbar vertebrae and thus closely resembles the human spine. The thoracic spine has a kyphosis (concave curvature) and the lumbar spine is straight in the seated position while the neck has a lordosis (convex curvature).
Figure 9.2: Schematic drawing of the BioRID dummy torso, arm attachments, spine, neck and modified pelvis with Hybrid III head in seated position.
(Reproduced from the International IRCOBI Conference Proceedings, Davidsson et al, 1998)

An occipital interface piece is rigidly mounted to a modified version of a Denton Hybrid III upper neck load cell. The top cervical vertebra and the occipital interface were designed to allow the head to be horizontal whilst maintaining the same joint characteristics as the rest of the neck joints. The bottom lumbar vertebra is connected to a pelvis interface which is mounted to the pelvis.

The vertebral bodies are made of durable plastic and are connected by pin joints which only allow for angular motion in the sagittal plane. This 2-dimensional performance of the spine is a serious limitation of the new dummy. The chosen angular ranges of motion of the lumbar, thoracic and cervical spine were based on equivalent data from the literature for the human spine.

In the spaces between the vertebrae are blocks of polyurethane rubber glued to the nearest inferior vertebra. The size, hardness and position of these rubber blocks determine their contribution to the joint stiffness characteristics. In the thoracic and lumbar spine, the steel pin joints constitute linear torsion springs which may be adjusted to change the overall curvature of the spine. A M ADYMO model was used to determine the most appropriate joint stiffness characteristics and hence BioRID was constructed accordingly (Linder et al, 1998).

The BioRID neck also incorporates muscle substitutes which consist of wires originating from the head, at the front and rear of the occipital joint, guided through the cervical vertebrae and terminating at T1. At this point the wire load is transferred, via nylon coated steel wires and wire housing to a spring in parallel with a damper. This mechanism permits a more realistic replication of the head and neck retraction motion in which the head movement lags behind that of the neck (Linder et al, 1998).

The torso of BioRID comprises chest and abdomen moulded in a soft silicon rubber. The static bending stiffness of the rubber torso accounts for about 40% of the overall upper-body stiffness. The torso surface contour resembles that of a seated 50th percentile male. The spine is enclosed in a curved rectangular container inside the torso at the back of which is a Teflon foil to reduce friction between the vertebrae and the torso. A total of 15 steel tubes, of 10mm diameter, connect the spine to the rubber torso and simulate the clavicles and rib cage. The abdominal region contains a water filled bladder (volume = 2 litres) which reduces the bending resistance of the torso. A modified Hybrid III shoulder joint is attached to a scapula-clavicle structure which is moulded into the silicon rubber.
In the BioRID pelvis, the original Hybrid III anterior-superior iliac spine height has been decreased to conform with the modifications of the Advanced Anthropomorphic Test Dummy (Schneider et al, 1992). The original pelvis front flesh has been removed to allow the abdomen to bulge forward. Furthermore, the pelvis flesh has been modified to reduce femur joint flexion/extension resistance. The mass distribution within BioRID has been designed to closely resemble that of a 50th percentile male although the mass of the spine would appear to be considerably greater than that of a human spine.

At the time of writing, development of BioRID is still continuing. Early sled tests with the new dummy have shown it to give repeatable and reproducible results. It has been validated against volunteer tests (Davidsson et al, 1998b) and compared with Hybrid III performance. In these tests the BioRID’s T1 and head rearward and angular displacements were close to that of the average volunteer while the Hybrid III’s displacements were much smaller than those of the average volunteer. Neither the BioRID nor the Hybrid III were able to mimic the volunteer T1 and head upward motion. Hence further development work is required to improve the biofidelity of BioRID.

9.3 DEVELOPMENT OF A REAR IMPACT DUMMY AT TNO, HOLLAND

Researchers at TNO are currently engaged in a project to develop a more biofidelic dummy to assess the performance of seats and head restraints in rear impacts. The project is a collaborative effort involving a number of car manufacturers across Europe plus a seat manufacturer, a dummy manufacturer and a university. The project is due to be completed sometime in the year 2000.

The approach at TNO has been to modify the standard Hybrid III dummy to give a more realistic performance in rear impacts. The main modifications have been:-

a) The incorporation of the TRID neck, developed at TNO, which has an adjustable base plate which allows a more realistic positioning of the dummy head and neck.

b) The inclusion of a black stiff foam back section which gives a more realistic contact between the dummy and the seat.

c) The pelvis has been cut out and a pin joint inserted so that during a rear impact the dummy legs will tend to stay in their original positions while the torso lifts up (i.e. hyperextension of the hip). This replicates the "ramping up" behaviour which has been witnessed in cadaver and volunteer tests.

d) A load cell has been mounted on the rear interior of the dummy head to monitor any head restraint contact.

e) The dummy wears a rubber wet-suit to improve the friction between the dummy and the seat.

In making these modifications TNO’s main concern has been to produce an acceleration and velocity pulse at T1 which agrees, within limits, with those recorded in volunteer and cadaver tests. The manner in which the input pulse to the neck is generated at T1 is of secondary importance. Thus, TNO are not concerned with the design of the spine below T1.

It is planned that the final model of the new rear impact dummy will be tested at 10, 16km/h plus a third (higher) speed yet to be defined. The results from these tests will be validated against equivalent volunteer and cadaver tests complemented by MADYMO modelling (van den Kroonenberg, 1997)

9.4 SPINAL DEVELOPMENTS IN CANADA AND THE USA
Schneider et al (1992) have described an advanced anthropomorphic test device based on the Hybrid III dummy. It includes a flexible thoracic spine connected to a more humanlike rib-cage (see figure 9.3). The improved spine system consists of upper and lower rigid thoracic segments separated by a flexible rubber segment. Bilateral steel cables have been incorporated into the assembly to provide durability and lateral stability. The location of the rubber spinal articulation corresponds approximately to the T7 location in the human. The upper thoracic spine segment comprises two side plates with two steel shelves welded inside for mounting the modified neck bracket or load cell at the top and the thoracic spine articulation at the bottom. The front of the upper thoracic spine is open to allow access to wiring and associated connectors while the back is enclosed with a stairstep plate designed to orientate and fix the top four ribs at their specified angles.

The lower thoracic spine has been similarly designed but is longer and thinner to allow for the mounting and inward rotation of four chest deflection transducers. The bottom three ribs are attached to the back of the lower spine which is also fitted with some lead ballast to increase the torso mass and to lower the thorax centre of gravity to a humanlike position. Both the upper and lower thoracic spine segments are able to accommodate angular velocity transducers to monitor the relative motion at the thoracic spine articulation.

The new geometry and new thoracic spine elements have enforced a new design of lumbar spine to interface between the thoracic spine and the Hybrid III pelvis. This lumbar spine consists of steel plates moulded into the ends of a block of natural rubber with holes provided for two bilaterally positioned steel cables. A six-axis load cell may be installed between the lumbar spine and the pelvic ballast block.

At the other end, a new neck mounting bracket and lower neck load cell have been incorporated so that the Hybrid III neck is mounted directly in line with the upper thoracic spine in such a way that the head and top of the neck retain their original (Hybrid III) orientations and positions relative to the pelvis in the pre-test posture. The overall result is a more humanlike spinal contour with a continuous, uninterrupted curvature from the lumbar spine to the top of the neck. The results from sled tests using the improved dummy thorax described above have shown that the whole dummy “responds in a more compliant and decoupled manner than the Hybrid III”.
Figure 9.3. Schematic diagram of the spine and thorax of an advanced anthropomorphic test device from Schneider et al 1992.

(Reprinted with permission, SAE 922520 © 1992, Society of Automobile Engineers, Inc.)
The National Highway and Traffic Safety Administration (NHTSA) in the USA have been investigating the mechanical simulation of the human neck for many years. As a result of this research, White et al (1996) have designed an improved neck which can be incorporated in the advanced anthropomorphic test device. This neck consists of a column of five rubber pucks separated by aluminium discs. At the top of the neck is located a six component load cell that incorporates the condyle pin which is attached to the Hybrid III head. At the front and rear of the neck column there are cables which connect the head to compression springs at the base of the neck. These compression springs develop a restraining force on the head as the neck moves in either flexion or extension thereby generating a lag between the movement of the head and movement of the neck. This lag between the two motions is typically observed in volunteer and cadaver tests but is not seen in tests using the standard Hybrid III dummy.

Sled tests with the new neck have shown it to have greater biofidelity than the standard Hybrid III neck as well as replicating realistic head and neck motion. However, it appears to be mechanically too stiff when its performance is compared with the response of cadaver and living human necks to impact. Hence, further development work is required for this particular neck design.

Gibson et al (1994) at Biokinetics and Associates Ltd in Canada have modified the standard Hybrid III neck for use in motorcycle crash tests. Their modified hybrid III neck incorporates a torsional element mounted at the top of the neck which connects with the Hybrid III head. This element extends the range of flexion and extension motion possible. Effectively, it also introduces a low stiffness torsional spring in series with the neck thereby reducing the overall initial stiffness for the first 40° of rotation. The addition of this element which has a mass of 0.84kg and increases the length of the neck by 12mm has markedly enhanced the biofidelity of the Hybrid III neck.

The most recent crash test dummy developed in the USA is THOR (Test device for Human Occupant Restraint). It is the successor to the advanced anthropomorphic test device described above and has a similar construction. This new dummy has been designed for use in frontal impacts and is still being evaluated. Figure 9.4 shows the head and neck of the THOR dummy in comparison with those of a human volunteer and the Hybrid III. The diagram also shows the relative position of T1 for all three cases.

The THOR head and neck are connected via the condyles pin and by means of two cables; one at the front and one at the back of the neck. These cables are attached to springs assembled inside the THOR head. Before every impact test, these springs should be tightened so that the head is kept upright before a test. However, there should be no large pretension in the cables so the springs should not be over-tightened. The upper load cell records forces and torques at the top of the neck. The tension forces in the two cables are recorded by a front spring load cell and an aft spring load cell respectively. A lower neck load cell records loads at the neck base.
Figure 9.4: Schematic diagram of the heads and necks of a human volunteer, the THOR dummy and the Hybrid III dummy.
(Reproduced from the International IRCOBI Conference Proceedings, (Hoofman et al, 1998))

The construction of the THOR neck encourages the typical lag between head motion and neck motion which is observed in human volunteer and cadaver tests. Initially, during a test the relative rotation between head and neck takes place without significant friction at the condyles pin. When the aft stop hits the upper load cell surface the head position is ‘locked’ with respect to the neck and a moment starts to develop at the condyles pin.

Recently, Hoofman et al (1998) at TNO have evaluated the performance of the THOR neck in frontal impacts and have found that, although better than a Hybrid III neck, it elongates more than a human neck in forward flexion. There is also some doubt over it’s durability in regular frontal impact tests.

9.5 OTHER WORK ON PHYSICAL NECK MODELS

Myers et al (1989) compared the torsional response of the Hybrid III neck to that of humans, prompted by concerns about torsional effects in side impacts and (to a certain extent) frontal impacts. The Hybrid III neck was found to be much stiffer than the human in torsion, but modification of the dummy neck by addition of an ‘atlanto-axial’ joint improved it.

Sekizuka (1998), during development work on a new anti-whiplash seat for Toyota (see below), modified a standard Hybrid III neck by cutting segments out to make it less stiff. Sekizuka claims that this gives a response similar to that of the TRID II neck, but the present author feels that if producing a biofidelic neck were really this easy, then Chalmers, TNO and other research organisations would surely have thought of it a long time ago.

The US Naval Biodynamics Laboratory has conducted research on neck injuries and head/neck kinematics in impacts for many years, using both volunteers and cadavers (Wismans et al, 1986, 1987, Thunnissen, 1995). They have developed a model of the neck based on a rigid link, with omnidirectional ball and socket joints either end, corresponding to the atlanto-occipital and T1 joints. Rubber bumpers are used to control the range of motion, and springs at the rear control movement (de Santis, 1991). However, the main thrust of all this work was primarily to mitigate injuries to military personnel involved in severe frontal and side impacts. Its relevance to soft tissue injuries in rear impacts is probably limited.
Bilston et al (1993) report on the development of an isolated cervical spine model, including a spinal cord and brainstem, to study cord injury due to neck trauma. However, this was designed to focus on severe injury only.

Shimamoto et al (1998) describe work on a neck model (the Biomechanical Cervical Model - BCM) for use in low-speed collision testing. The model consists of realistically-shaped vertebrae (cast in polyurethane from medical models), intervertebral discs made from silicone rubber and ligaments made from three-dimensional textiles. The whole assembly is encased in clear silicone rubber so that the motions of the individual components can be observed. Rear impact sled test comparisons between the BCM (fitted to a Hybrid III), the standard Hybrid III neck and the TRID II neck indicated that the BCM is more flexible than even the TRID II neck, and the authors conclude that it shows promise as a tool for investigating low speed neck injury mechanisms.

A paper which does not deal with the construction of a neck model at all, but which contains basic data which may be useful for would-be neck modellers is Panjabi et al (1991). This contains detailed measurements of the dimensions of the cervical vertebrae (C2-C7) taken from 12 cadavers, including means and standard deviations of all dimensions.

9.6 OPPORTUNITIES FOR COLLABORATION WITH TRL

During the course of this literature review, one of the authors (PRD) visited research workers at TNO, Holland and in Gothenburg, Sweden. These visits yielded a considerable amount of valuable information on the current state of dummy spine development in Europe which has been reported in this chapter. The possibility of TRL collaborating with one of the research groups concerned with the development of rear impact dummies was also explored.

Volvo did not seem particularly interested in setting up a collaboration with TRL because of the commercial nature of the BioRID project and because their dummy has not yet been fully validated. Once all development work on BioRID has been completed, sometime in 1999, it may be possible for TRL to test and assess this particular dummy.

TNO viewed the prospect of future collaboration with TRL more favourably and indicated that TRL could become involved in testing and assessing TNO’s modified Hybrid III dummy (for rear impacts) with permission from the rest of the consortium members. However, as the assessment of the modified dummy will take place whether or not TRL are involved, this collaborative exercise may not be worthwhile.
9.7 REFERENCES


10 CONCLUDING SUMMARY

This report has presented a comprehensive review of the published literature on the biomechanics and injury mechanisms of the spine in relation to motor vehicle accidents and dummy development. The large number of books and articles written indicate that it is a vast and complicated subject. A brief survey of the anatomy of the spine (chapter 2) reveals that it has an intricate structure and that the interaction between the various ligaments, muscles, discs and bony parts is complex. However, a thorough understanding of the construction and biomechanics (chapter 3) of the spine is required before it can be accurately modelled either physically or mathematically. Such knowledge is also essential to the understanding of injury mechanisms.

In the past, engineers have tended to look at parts of the human skeleton, such as the spine, mainly from a mechanical perspective with little reference to any medical information. This approach is limited because the experience of the doctors and surgeons who see and treat injuries sustained in road traffic accidents everyday should also be taken into account. There is a huge quantity of medical literature concerning injuries to the spine, especially the neck, and a full perusal of it all has been beyond the scope of the current survey. Nonetheless, the review has focussed on some of the more relevant technical papers, which may have a bearing on future dummy spine design. This study could be enhanced with a broader ‘non-automotive’ survey of spinal injury.

Much of the medical literature deals with the diagnosis, treatment, rehabilitation and prognosis of spinal injury and at first sight may appear to be of limited use in the design of an improved dummy spine. However, the content of this literature highlights the fact that some of the long term disabling injuries (e.g. behavioural dysfunction) may not be predicted from instrumentation measurements made using a lifeless dummy. The review has benefitted from the input of spinal surgeons in Manchester and Nottingham and these valuable collaborations should be maintained in any future work.

The perplexing subject of cervical spine injuries has been discussed in chapter 4. It is clear that there is still no overall agreement on the mechanisms and injuries, which cause ‘whiplash associated disorders’. Each medical expert has his or her own favourite theory about how these types of injury are produced. The leading theories were highlighted at the end of chapter 4. It is most likely that ‘whiplash’ injury results from many different effects. Thus, the design of a biofidelic neck must allow the measurement of a number of potentially relevant parameters during the research phase. This must be accompanied by a continual monitoring of current whiplash research.

More severe spinal injury which may involve damage to the cord, ligaments and vertebral bodies is better understood, particularly in the thoracolumbar spine (chapter 5). However, this section of the spine has been rather neglected in favour of research on the neck and neck injuries. Thus there is a shortage of data on the tolerance and biomechanics of this region of the spine to impact. There is also a paucity of information on the physical properties of the vital components of the thoracolumbar spine. What little data there is tends to be ill-defined and disparate with large variations in the results reported by different researchers.

It could be argued that the bias of research effort towards study of the neck is justified because accident statistics show that this is where most spinal injuries occur. However, serious injuries to the rest of the spine are also important. The disabling effects of them can have considerable financial implications for society at large not to mention the cost in terms of human suffering. Chapter 6 detailed a brief survey of spinal injuries in traffic accidents. This survey could be extended to examine the relative importance of thoracolumbar spine injuries compared with cervical spine injuries. The results of the preliminary survey, described in chapter 6, suggest that the occurrence of the former injuries are much rarer than the latter. Other factors affecting spinal injury, such as impact direction, severity of impact, type of vehicle and seat stiffness could also be investigated in more detail.
Chapter 7 showed that mathematical models can be cost effective and flexible research tools for the study of the response of the spine to impact. These models are particularly powerful when supported and validated by a complementary experimental test programme. Thus, it is expected that finite element modelling will have a prominent role to play in any future work on dummy spine development at TRL.

In chapter 8 the limitations of the Hybrid III spine were discussed. These deficiencies are well known and are widely reported in the literature. The original design of the Hybrid III neck is now some twenty five years old. Some of the theories which circulated at the time of its development, such as ‘whiplash injury is caused by hyperextension, have been discredited. It is not surprising, therefore, that the Hybrid III neck is too stiff for low severity impacts. The rest of the Hybrid III spine is also clearly lacking in human biofidelity and yet this dummy is used worldwide in new car assessment programmes.

The Hybrid III dummy was originally designed for frontal impacts and so is of limited use in side and rear impact tests. It is possible that current measurements of head acceleration using the Hybrid III may yield misleading results because the head sits on top of a spine whose dynamic behaviour does not match that of a human spine. This concern has initiated an Australian project to develop a new omnidirectional neck for the Hybrid III dummy. The new THOR dummy, developed in the USA, marks an important stage in anthropomorphic dummy design and has the potential to offer improved biofidelity of the neck in the future.

When it comes to designing a new dummy spine, engineers and scientists in Sweden are very active in the field (chapter 9). Much of the Swedish research is founded on a hypothesis first proposed by Professor Aldman in 1986. He postulated that sudden and violent pressure changes which occur, within the spinal canal of the neck, during a rear impact may be sufficient to cause damage to the spinal ganglia. Subsequent experiments with pigs demonstrated that this was a feasible injury mechanism. It appeared to be all the more plausible because the typical symptoms of whiplash (e.g. dizziness, radiating pain etc) are consistent with the suggested neurological damage.

Next came the development of the rear impact dummy neck (RID-neck) and from that the new BioRID. The Navier-Stokes theorem was applied to the fluid dynamics within the neck and hence the neck injury criteria (NIC) was derived based on the relative acceleration between the top and bottom of the neck. At the recent IRCOBI conference in Gothenburg (September, 1998) there was a proposal for a new Swedish rear impact test procedure which incorporated both BioRID and NIC.

However, they should proceed with caution because the original Aldman hypothesis may be flawed and has not yet been proved to be a real injury mechanism in humans. Furthermore, NIC has not been properly validated for human subjects. It may be a good indicator of neck injury but not for the reasons originally put forward. Lastly, BioRID is a dummy that is designed for pure rear impacts and has no lateral or rotational sensitivity.

The rear impact dummy development at TNO, Holland also appears to be quite limited. Their approach has been to make some slight adjustments to the existing Hybrid III and to incorporate the TRID neck, which is a modified version of the Swedish RID neck. Again this neck only works in two dimensions and can only be used in low velocity rear impact tests up to 20km/h.

In conclusion, this report has illustrated the complexity of spinal injuries, particularly whiplash, in humans and spinal developments in dummies. The causes of whiplash associated disorders are still poorly understood and current test hardware is limited in its biofidelity and performance. There still appears to be a need for a biofidelic omnidirectional dummy spine and neck for the assessment of vehicle seats and the prediction of spinal injury. With appropriate application of resources and research effort, TRL should be able to make a valuable and worthwhile contribution to this field.
11 ACKNOWLEDGEMENTS

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12 APPENDIX A: Glossary of Medical Terms

**Annulus Fibrosus**
Tough fibrous flexible ring which constitutes the outer part of an intervertebral disc.

**Anterior**
Relating to the front part of the body or of any body component.

**Anterior Longitudinal Ligament**
A long ligament which runs continuously down the front of the vertebral column and attaches to each of the intervertebral discs.

**Atlanto-axial**
Relating to the joint between the Atlas and the Axis.

**Atlanto-occipital**
Relating to the joint between the Atlas and the Occiput, or base of the skull.

**Atlas**
The uppermost cervical vertebra (C1), so called because it supports the head.

**Axis**
The second cervical vertebra (C2) which determines the rotational motion of the head.

**Coccyx**
Four fused vertebrae at the very bottom of the spine.

**Comminuted Fracture**
A fracture in which the bone breaks into more than two pieces.

**Dens**
Latin name for the odontoid process which is a ‘peglike’ bone emerging from the axis (C2) and acts as a pivot about which the atlas (C1) and head can rotate laterally.

**Dislocation**
Displacement from their normal positions of bones meeting at a joint (often facets).

**Distraction**
Mechanical pulling apart of the bones and associated soft tissues at a joint.

**Electromyogram (EMG)**
Continuous record of the electrical activity of a muscle.

**Erector Spinae**
A large muscle which keeps the spine in tension and holds the spine erect.

**Extension**
Backwards bending of the spine or part of the spine.

**Facets**
The articulating surfaces of the articular processes at the rear of the spine.

**Flexion**
Forwards bending of the spine. Lateral flexion is a bending of the spine to either side.

**Illo Psoas**
Large group of muscles which connects the transverse processes of the lumbar spine with the pelvis.

**Inferior**
Relating to the lower part of the body or of any body component.

**Inferior Articular Processes**
Pairs of bony structures which emerge either side of the bottom of a vertebral arch. They articulate with the superior articular processes of the neighbouring vertebra and so determine the natural range of spinal motion.

**Intervertebral Foramen**
An opening between two vertebrae through which pass nerves connecting the spinal cord to various parts of the body.

**Interspinous Ligament**
Ligament which connects one spinous process with a neighbouring one.

**Kyphosis**
Outward curvature of the spine (ie concave when viewed from the front). A normal spine has kyphotic curvature in the thoracic and sacral regions.

**Laminae**
Two flat bony structures which are joined together to form the rear portion of the vertebral arch.

**Ligamentum Flavum**
A separate sheet of elastic connective tissue which passes between the neural arches and connects one lamina to the next.

**Lordosis**
Inward curvature of the spine (ie convex when viewed from the front). A normal spine has lordotic curvature in the lumbar and cervical regions.

**Nucleus Pulposus**
Transparent jelly material found inside intervertebral discs.

**Occipital Condyles**
Two rounded surfaces at the base of the skull which articulate with the atlas.

**Odontoid Process**
A ‘peglike’ bone emerging from the axis (C2) which acts as a pivot about which the atlas (C1) and head can rotate laterally.

**Osteoligamentous**
Relating to bone and ligaments

**Pedicles**
Two short bony structures either side of a vertebral body which connects the body to the laminae.

**Posterior**
Relating to the back part of the body or of any body component.
**Posterior longitudinal Ligament**
A long ligament attached to each intervertebral disc which runs continuously down the inside of the spine at the back of the vertebral bodies.

**Processes**
Bony projections which emerge from the vertebrae. A typical vertebra will have seven processes.

**Sacrum**
Five fused vertebrae (in the adult) between the base of the lumbar spine and the coccyx.

**Sagittal**
Describing the fore-and-aft plane extending down the long axis of the body, dividing it into right and left halves.

**Spinous Process**
Bony projection emerging from the rear of a vertebral arch.

**Sternocleidomastoid Muscle**
A long muscle in the neck extending from the mastoid process to the sternum and clavicle. It serves to rotate the head and flex the neck.

**Subluxation**
Partial dislocation, so that the bone ends are misaligned but still in contact.

**Superior**
Relating to the upper part of the body or of any body component.

**Superior Articular Processes**
Pairs of articulating bony structures which emerge either side of the top of a vertebral arch. They articulate with the inferior articular processes of the neighbouring vertebra and thereby determine the natural extent of spinal motion.

**Supra-spinous Ligament**
A ligament which joins all the spinous processes together.

**Thoracolumbar Spine**
A term which refers to a combination of the thoracic spine and lumbar spine. The term is often used when referring to the junction between the thoracic and lumbar spine.

**Transverse Processes**
Bony structures which project laterally from the vertebral arches and act as connection points for the muscles.

**Vertebral Arch**
An arch of bone, comprising two laminae and two pedicles, which extends rearwards from a vertebral body and encloses the spinal canal.

**Vertebral Body**
The ‘horse-shoe’ shaped portion of bone at the front of a vertebra which has a load bearing function as well as protecting the front of the spinal cord.

**Vertebral Foramen**
The space between the vertebral arch and the vertebral body which contains the spinal cord. Collectively, the vertebral foramina of all the vertebrae form the spinal canal.
Annex 2: Volunteer approvals

Any work involving live human subjects generates serious ethical and insurance issues and those issues that are particular to this study are described here. These are worth documenting because this is the first report of experimental rear impact tests being carried out on human volunteers in the UK. The experimental protocol was thoroughly assessed and was, in due course, approved by the North West Surrey Local Research Ethics Committee.

Ethical considerations
Performing low energy rear impact tests on human subjects does carry a small but finite amount of risk of injury to the participants. For this reason, there is a serious ethical question as to whether such impact tests on individuals can be justified. Essentially, the answer to this comes down to whether the gain in knowledge and understanding outweigh the risk involved. The fact that this type of work has been safely conducted in numerous studies around the world and there have been no reports of long term injury to any of the volunteers involved endorsed the possibility of conducting a rear impact trial at TRL. These experiments, if properly controlled and supervised, should not be any more dangerous and should more likely be far less hazardous than, for instance, a typical dodgem car ride at a fairground.

Nevertheless, because of the small risk involved and the controversial nature of this type of work, both TRL and the DETR (now the DfT) agreed that ethical approval should be sought for the proposed volunteer study before proceeding. Thus a proposal document which included an assessment of the likely risks and which outlined the test protocol was submitted to the North West Surrey Local Research Ethics Committee for consideration. This body comprised 13 people and included hospital consultants, local GPs and a number of lay persons.

The final version of the proposal document that was sent to the ethics committee can be found in Appendix A of the interim report on the volunteer testing. It was prepared in consultation with TRL Limited’s medical advisors, Drs Strudley and Whitfield and contains a volunteer consent form, volunteer information sheet, medical screening questionnaire and an adverse event form for use in the unlikely occurrence of injury to one of the volunteers. These forms are typical of those used in any trial involving human volunteers.

The ethics committee met on 1st October, 1999 to discuss the proposal for conducting low speed rear impact tests on human volunteers at TRL. The case for conducting the tests was presented at the meeting by Drs Strudley and Whitfield and TRL. The committee duly granted ethical approval for the study subject to the following stipulations: -

i) The volunteers must not be remunerated for taking part.

ii) Volunteers may only be recruited from outside TRL and must not be relatives of TRL employees.

iii) Written consent must be obtained from prospective participants prior to any medical examination.

iv) Full indemnity insurance must be in place and the necessary documentation must be submitted to the committee for review.

v) The volunteer information sheet should include details of the lead investigator and the medical supervisor. It should also outline the risks involved and address the subject of compensation in the event of an adverse incident.

vi) Consent should be obtained from participants for their GPs to be informed of their participation in the study.
Points (iii) to (vi) were relatively straightforward to deal with. TRL Limited’s insurers were notified of the proposed study and they agreed to provide insurance cover providing that it was to be a ‘one-off’ experiment. An additional premium of £500 was charged and an excess of £5,000 per volunteer was imposed.

The demands of (i) and (ii) were completely unexpected and caused problems in recruiting volunteers which led to a significant revision of the project. The reason for (ii) was that the committee was concerned that if TRL Limited staff were allowed to participate they might be coerced into so doing. The committee introduced stipulation (ii) so that an individual’s decision to become involved should not be influenced by any financial reward.

Volunteer recruitment
As outlined in the Proposal document sent to the ethics committee:

Recruitment issues
- Volunteers for the study will be recruited from the TRL data base of potential volunteers together with other contacts from outside the company. No TRL staff or family members will be used and volunteers will receive no financial reward for participating. Receipted travel and accommodation expenses will be paid.
- All volunteers will be medically screened before and after their participation in the tests.
- No tests will be conducted using a volunteer without first obtaining their informed written consent. Any volunteer may, of course, opt out of the test programme at any time.

Inclusion criteria
- Volunteers should be aged between 18 and 40 years.
- At least 10 male volunteers will be selected with masses and heights close to that of the Hybrid III dummy (i.e. 81kg, 175cm).
- Some female volunteers may be included in the study to investigate the difference between male and female spinal behaviour.

Exclusion criteria
Volunteers will be excluded from the study if they:
- are less than 18 years old
- are more than 40 years old
- suffer from back problems
- have any spinal disorder
- have any cardiovascular disorder
- suffer from epilepsy
- are pregnant.
Kinematics of the Human Spine in Rear Impact and the Biofidelity of Current Dummies

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ABSTRACT
There is widespread concern over the incidence of whiplash injuries in the world of automotive medicine and much research has been conducted into this poorly understood injury. This paper describes research into the dynamic mechanical behaviour of the whole human spine in low speed rear impacts. A series of rear impact sled tests using ten male volunteers has been conducted at 4 km/h and 7 km/h and comparative tests using Hybrid III, THOR and BioRID at 7 km/h. The principal aims of the research were to:

1. Determine human kinematics in a low speed rear impact and to use this as a target for the future design of a rear impact dummy;
2. Study the interaction between a human torso and a semi-rigid seatback and to compare this with the equivalent interaction between dummies used for rear impact;
3. Evaluate the muscle activity in some of the major muscle groups of the neck during rear impact.

The dummies evaluated were the Hybrid III with the TRID neck, BioRID and the THOR dummy with its standard neck and with a EuroSID-1 neck. The most biofidelic dummy appeared to be the BioRID dummy. This was particularly evident when comparing dynamic seatback pressure mapping.

Surface electromyography (EMG) measurements were taken during the volunteer tests. These showed that some of the major neck muscles had a rapid involuntary response to the impact, and appeared to be rapid enough possibly to affect the head and neck kinematics during the impact. The responses may have been stimulated by motion in the lower body, which was the first body part to respond to the impact.

The EMG results suggest that observations based on cadaveric testing, for neck behaviour may be misleading and that any criteria that may be adopted to evaluate whiplash injury may have to incorporate more than simple neck based measurements.
1 INTRODUCTION

Accident studies show that low severity neck injury is relatively common, often occurring in low severity accidents. Many of these soft tissue neck injuries disappear over a period of time with no long-term consequences, but some can have a long-term, >12 months, debilitating effect. The precise cause of these injuries is not well understood. In the automotive situation neck injuries are often associated with minor rear end impacts or shunts and are generally described as Whiplash injuries, based on the hypothesis that they are caused by the head ‘whipping’ backwards as the torso (upper body) is driven forwards, due to the impact. These neck injuries are now generally classified as WAD type injuries (Whiplash Associated Disorders). Accident and injury studies have shown that these injuries are not unique to the rear impact situation are also associated with impacts from other directions. It is suggested that the WAD injuries are also with the rebound phase of an impact, as well as the initial loading phase.

Most accident and injury studies focus on the more severe end of the injury spectrum, often being categorised by means of the Abbreviated Injury Scale (AIS) [1]. The AIS injury scaling code focuses on what might be termed ‘life threatening injury’, although not exclusively. Social and clinical studies have indicated that people who sustain ‘minor’ injuries, as classified by AIS, can have very long term problems, in terms of pain, restricted movement and in some instances behavioural changes. Many WAD type injuries, on the AIS scale, are only classed as AIS 1 (minor).

Pain and discomfort issues are personal ones but some assessment of the extent of these injuries can be gauged from the vehicle insurance industry and the extent of personal injury claims. Insurance data strongly indicates that WAD type injuries and associated personal injury claims are rising rapidly for a number of different reasons. WAD type injury claims in the UK are reported to have a value of between £1.2b and £1.6b based on ABI data (Association of British Insurers), exceeding the value of all other personal injury claims. Murray et al [2, 3] have carried out a number of studies on the costs of injury, slight and serious, leading to long term disability. It might be expected that seriously injured people would have long-term problems but many occupants with slight injuries were also seen in the sample. They noted that the average days off work, over an 18-month period, were 60 days for seriously injured patients and 19 days for the slight group.

2 ACCIDENTS AND INJURY

A number of accident studies have been carried out into whiplash injury and its cause. In the UK the most comprehensive in-depth study is the CCIS [4] (Co-operative Crash Injury Study). Unfortunately, this samples the more severe accident cases whereas most whiplash type injuries are likely to occur below the sampling threshold of the CCIS. Even so Morris et al [5] have reviewed soft tissue injuries within the CCIS, up to 1996. They observed that soft tissue injuries occur in both frontal and side impact accidents, as well as rear and that many of them are ‘self reported’ rather than clinically diagnosed. They noted a significant difference between male and female risk, with women being much more prone to soft tissue injury than males. Of their sample of 1887 occupants 21% of females had a soft tissue neck injury compared to 13% of males. In their study the value of head restraints, at reducing neck injury, was found not to be statistically significant. They did note that seat back yielding had a slight trend towards reducing neck injury but the change was not statistically significant. They also observed that the frequency of soft tissue neck injuries increased over their ten-year study period, the implication being that current vehicle's seats (Year 2000+), may now induce more frequent injury. This is an observation being confirmed within the insurance industry with a rise in personal injury claims, related to whiplash injury.

Possibly the best source for current information of the occurrence of whiplash injury emanates from the insurance industry as the reporting of such injuries are likely to result in a financial claim for...
treatment or compensation. A recent Direct Line\(^3\) review [6] suggests that the cost of whiplash injury, in the UK is £800 million per annum, based on year 2000 data, equating to £30 of every motor insurance premium. This figure is below that proposed by ABI but is still a massive cost.

Head restraints have been proposed as a means of reducing WAD type injury and a number of studies have suggested that a good well-positioned head restraint can have positive benefits. A static head restraint assessment device and evaluation procedure have been developed to encourage the better design of head restraints.

Most accident studies have focused on neck injury. The UK Department for Transport, Local Government and the Regions (DTLR) has also looked at low severity lower back and neck strain injury [7]. Just under half of the investigated cohort of neck sprain subjects reported lower back injury for more than one week after the first assessment, with a similar frequency between males and females. A higher risk of lumbar injury was noted in rear impacts. One of the conclusions drawn from their study is that studying neck injury in isolation may not yield the benefits that an integrated approach may have and that lower back injury should also be evaluated.

One other very interesting fact that is observed within the literature is the gender difference in WAD injury risk. Föret-Bruno et al [8] suggests that females experience up to twice the risk of WAD injury than do males. This finding is also supported elsewhere which raises a very interesting issue of gender based injury criteria and populations at risk.

It has been suggested that muscle reaction time may be too slow to influence whiplash injury risk. Szabo et al [9] studied EMG activity in ten 16km/hr whole vehicle (1970 model year Volvo) rear impacts. They deprived their subjects of external indications of an impending crash, trying to minimise subject pre-tensing. They did observe that all neck muscle groups seemed to fire together being triggered by lumbar spine acceleration, 90-120ms after the onset of this pulse. Peak muscular activity was noted during cervical spine flexion. One might hypothesise that if the subjects had been more aware of an impending impact then this activity might have occurred earlier as muscle activity can be quickened with pre-knowledge. It is likely that muscle activity could influence the risk of soft tissue injury.

3 BACKGROUND RESEARCH

3.1 Hybrid III

The Hybrid III dummy is currently the standard dummy used in determining injury risk in frontal impacts. Until recent years this was the only anthropomorphic crash test dummy that had been used to study rear impact and develop improved seating systems. A number of research groups have questioned its value when used in low speed rear impact studies and have suggested that it may not be sufficiently biofidelic to predict whiplash type injury.

During an evaluation of the Hybrid III dummy neck, Thunnissen [10] stated that in rear impacts 74% of whiplash patients did not anticipate the accident and were therefore relaxed at the time of the impact. From this study they suggested that the Hybrid III dummy neck was too stiff for studying low severity rear impacts since the maximum head angles recorded in tests were too low to represent a relaxed person.

Lovsund and Svensson [11] reported that the neck and spinal structure of the Hybrid III dummy was stiff and unlikely to interact correctly with the seatback in the same compliant way as would the human spine. Seemann et al [12] also found the Hybrid III far too stiff to respond in a human-like manner in the sagittal plane. Deng [13] found that results from a mathematical model of the Hybrid III neck indicated that the neck has a torque response similar to that of the human but has a higher shear

\(^{3}\) A large UK Insurance company
response. In volunteer tests McConnell et al [14] found that during the acceleration phase of a rear-impact, when the occupant's body was pressed against the seatback, the spinal curvature (lordosis) straightened. In a comparative study using volunteers and a Hybrid III, Scott et al [15] found that the dummy was less prone to "ramp up" along the seat back than were the volunteers.

3.2 Dummy developments

3.2.1 Rear impact dummy neck (RID) [16]
The RID neck, a Swedish development, was specifically designed to give a better simulation of the shearing behaviour of the human neck. Early use of the RID neck in experimental work confirmed that it was a significant improvement over the standard Hybrid III neck, but that there was still room for improvement. Subsequent development work on the RID neck has been undertaken by TNO in the Netherlands, and is described by Thunnissen et al [10] The result of that development work is the TRID II neck. The new design is said to give improved reproducibility and to agree well with published head/neck rotation and other parameters. However, they point out that bending of the thoracic spine heavily influences head/neck kinematics in rear impacts and what is needed is a biofidelic model of the complete spine.

3.2.2 BioRID
BioRID is the latest development to come from a Swedish research group. It is essentially an extension of the original segmented RID-neck but to the whole spine. The dummy has an articulated spine and quasi-static neck muscle substitutes, the thoracic spine also has a kyphosis (concave curvature) whilst the neck has a lordosis (convex curvature) closely resembling that of a human spine. Early sled tests with the new dummy have shown it to give repeatable and reproducible results. The BioRID is now at a second stage of development and is now a marketable dummy. Having a long flexible spine, the BioRID has the potential to evaluate all injuries or changes in deformation along the whole length of the spine in a human like manner.

3.2.3 EC supported Whiplash research
Within the EC supported 4th Framework ‘Whiplash’ project further developments have been made with the TRID dummy neck and Hybrid III dummy to make it more biofidelic. Early dummy developments in the project modified the TRID neck, into the TRID II, with some minor modification of the standard Hybrid III dummy to give a more realistic performance in rear impacts. This has now been further developed using parts of the Hybrid III dummy and the new THOR dummy. The new dummy, which at the present time is not available for general evaluation, is called RID-2 α-prototype [17]. The input requirements for neck performance, from the research carried out in the Whiplash-1 programme, are based on the reproduction of acceleration and velocity at the T1 level, the top of the thoracic spine generated from data recorded in volunteer and cadaver tests. Whiplash-1 did not try and include other spinal input criteria.

3.2.4 Spinal developments in Canada and the USA
Schneider et al [18] have described an advanced anthropomorphic test device based on the Hybrid III dummy. It includes a flexible thoracic spine connected to a more humanlike rib cage. The results from sled tests using the improved dummy thorax have shown that the whole dummy "responds in a more compliant and decoupled manner than the Hybrid III."

The National Highway and Traffic Safety Administration (NHTSA) in the USA have designed an improved neck, which can be incorporated in the advanced anthropomorphic test device (White et al [19]). Sled tests with the new neck have shown it to have greater biofidelity than the standard Hybrid III neck. However, it appears to be mechanically too stiff when its performance is compared with the response to impact of cadaver and living human necks, hence further work is required.

Gibson et al [20] at Biokinetics and Associates Ltd in Canada have modified the standard Hybrid III neck for use in motorcycle tests, enhancing biofidelity.
The most recent crash test dummy developed in the USA is THOR [21] (Test device for Human Occupant Restraint). It is the successor to the advanced development anthropomorphic test device TAD50M. This dummy has been designed for frontal impacts but some groups are evaluating its low speed rear impact performance. One such evaluation is presented within this report.

4 Current study

The aim of the current study is to evaluate and develop a better understanding of the biomechanics of whiplash and a suitable crash test dummy that could be used to predict spinal injury in rear impact tests, with a focus on whiplash injury. In order to make realistic improvements with dummy developments, a full understanding of the kinematics, biomechanics and injury mechanisms of the human spine is required. Much research has taken place in this area and much more will need to be carried out in order to understand it more fully.

The first phase of the TRL experimental work was to conduct a series of low speed rear impact tests using the current Hybrid III dummy to establish its dynamic response, as this dummy could be considered to be the base from which further development could be made. TRL had already developed a finite model of the human thoracic and lumbar spine and during the project this model was further developed. The full spine model was integrated into a Hybrid III model in order to try to understand interactions between dummy and human. This part of the research is not included in this paper and will be presented in a later publication. During this development process it became apparent that there was a lack of validation data in conditions that could be easily replicated with test dummies.

A programme of human volunteer testing under very tightly controlled conditions was undertaken that could be easily replicated. These were carried out to generate an understanding of the behaviour of the human spine in low speed rear impacts on which to base and rate future dummy developments. The general methodology was based on that used by EEVC Working Group 9 [22, 23] who examined the cadaver database for side impact data. These data were then processed to derive biofidelity design targets. Once corridors have been created it is then relatively easy to compare and evaluate human surrogate test devices.

The research reported in this paper covers a programme of volunteer testing as well as comparative dummy evaluations. The paper presents a set of biofidelity design targets based on a simple impact sled, which can be easily reproduced at any time in the future. The assessment targets are based on transducers attached to the volunteers in low speed tests at 7km/hr. The research programme was designed to address spinal injury, not just whiplash. Other data are presented comparing volunteer and dummy responses across the whole of the torso.

4.1 Impact Testing

4.1.1 Test conditions

For biofidelity assessment it is important to have a simple easily reproducible test procedure. Many previous volunteer studies have either been based on vehicle seats or with vertical rigid seat backs at 90E to the base, which do not replicate vehicle posture. The former test condition is not easily reproducible and the latter not very vehicle like. In this study a surrogate vehicle seat was based on the bench seat used in ECE Regulation 44 to assess child restraints [24]. The R44 bench seat and test facility is well understood in child restraint evaluation. It was felt to be inappropriate to accelerate and than decelerate the volunteer backwards for safety reasons as well as the fact that it could stimulate unwanted muscle reactions as well as variable initial positions. To overcome these concerns a twin sled system was used. The second sled was used as bullet mass to impact the R44 sled. An aluminium honeycomb block was placed between the two sleds to limit the amplitude of the decelerating pulse experienced by the volunteer and dummy to 2g, in effect replicating a rear-end shunt type impact. The change in velocity, experienced by the subjects, was 7km/hr. For safety reasons the height of the R44 seatback was increased to support the shoulders. The higher seat cushion was made from a polyurethane foam (EV30) manufactured by Zotefoam and had a height of 590mm above the CR
point of the R44 bench\textsuperscript{4}. The front of the seat squab was 580mm forward of the CR point, both measurements being made along the plain of the surface. A head restraint was also fitted to limit gross motion of the volunteer's head, Figure 88.

The volunteers and dummies were fitted with accelerometers on the head and body (T1 and pelvis), Figure 89. The head accelerometers were fitted to the volunteers and dummies via a close fitting webbing harness with sufficient adjustment to ensure that the transducers were always mounted at a common point. The T1 accelerometers were fitted to a small aluminium bracket and surgically taped to the subject at the T1 location as firmly as could be achieved. The pelvic accelerometer was similarly fitted at the base of the spine above the bony protrusion of the sacrum. All tests were filmed with high-speed cine cameras but no kinematic data are presented in this paper.

Muscle activity and neck bracing are factors that could influence neck injury risk. All of the volunteers were equipped with surface mount EMG sensors mounted adjacent to the primary neck muscle groups, sternocleidomastoideus, erector spinae and trapezius, to try to determine if muscle activity could be a factor in the whiplash process.

A number of advanced occupant position sensing systems are being developed to determine to size, shape and position of an occupant, as a means of optimising the deployment of active restraint systems. It is therefore necessary that a dummy correctly interacts with the vehicle seat as pressure distribution can be one of the occupant sensing systems. Although not designed as a crash sensor a large multi-channel TECKSAN pressure mat was used to map dynamically the pressure distribution between the seat back and the volunteers and the dummies in most of the tests. The pressure mat used was 480 x 520mm utilising 1596 pressure nodes. It was mounted 50mm above the seat bite line. The maximum sampling rate of the system was 125hz. At this low sampling rate a general indication of any differences can be observed.

The impact test conditions used for the dummies were the same as those used with the volunteers. External head instrumentation was used for both dummies and volunteers except for the standard THOR dummy, where only internal head instrumentation was used. The data presented in the target corridors is for external instrumentation, except for the standard THOR dummy. The data from the dummy tests, when both internal and external head accelerations were measured, were found to be very similar. Thus the comparisons across the dummy range are believed to be valid, even though the standard THOR dummy results are based on internal measurements.

4.2 Test Results

\textsuperscript{4} A dynamic specification for the seat cushions will be presented in a later publication.
4.2.1 Volunteer Evaluations

Ten male volunteers (average age 26.5 years, height 1.785m and mass 77.5kg) were subjected to rear impacts at 7 km/h, to establish basic target data and for comparison with dummy and computer modelling techniques. These targets included acceleration profiles for the head, T1 and pelvis level as well as a consideration of the distribution of force exerted on the seat back. The transducer data were processed according to the procedures developed by EEVC Working Group 9 to derive appropriate target corridors. In brief the procedure consisted of developing a mean response and a corridor about the mean with a width of $\pm$ one standard deviation of the peak value. The standard deviation corridors were then simplified to create a definable corridor based on up to ten co-ordinate points.

Ethical approval was sought and granted for the volunteer test programme. Prior to and following testing all subjects were medically screened, and no problems were identified. A follow up investigation revealed no problems even after 12 months, with several of the volunteers offering to undergo further testing.

The data shown in Figure 90 and Figure 91 are for one parameter (T1 fore/aft acceleration) and shows the variability in volunteer data within this parameter. It should be noted that this was one of the most variable of the volunteer measurements. Figure 90 presents the mean volunteer response with the $\pm$ one standard deviation corridor, presented in Section 4.2.2. Figure 91 shows the straight-line conversion corridor and all of the volunteer data from which it was derived. Due to the statistically derived nature of the corridor it is not surprising that some of the volunteer data occur outside it.

4.2.2 Biofidelity design targets

Table 24 presents design target corridors for head acceleration and Table 25 for the T1 acceleration and pelvis acceleration for a rear impact dummy, based on the modified R44 test bench, using the analytical approach of EEVC WG9 [22, 23].

Figure 90 Volunteer mean response, $\pm$ one standard deviation and the generated corridor for T1 fore/aft acceleration.

Figure 91 All the volunteer data and the generated corridor for the T1 fore/aft acceleration.
Table 24 Dummy target corridors - Part 1

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Table 25 Dummy target corridors - Part 2

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4.2.3 Dummy Evaluations
Dynamic tests have been conducted on one of the TRL sled facilities with a Hybrid III dummy fitted with the TRID neck, an unmodified THOR dummy, a THOR dummy modified with the EuroSID neck, as it was felt that this neck may exhibit better omni-directional characteristics and the BioRID-I dummy, to establish their comparative performances in low speed rear impacts. The dummy responses are compared to the biofidelity targets developed from the volunteer tests under the same conditions. In this paper five parameters are reviewed.

4.2.3.1 Transducer data
Graphical results are presented for the Head vertical (x) and fore/aft (z) directions, Head resultant which includes the lateral component (y), Neck T1 acceleration in the fore/aft direction (z) and Pelvis acceleration in the fore/aft direction (z), shown in Figure 92 to Figure 96. The straight-line corridors are those derived from the volunteer data, Table 24 & Table 25, for the relevant parameter. The key to the five graphs is shown in Figure 97.

The pelvic BioRID data are not available, (Figure 96).
Figure 92 Fore / aft head acceleration

Figure 93 Vertical head acceleration

Figure 94 Resultant head acceleration
Figure 95 Fore / aft T1 acceleration

Figure 96 Fore / aft pelvis acceleration

Figure 97 Figure key
4.2.4 Pressure data

Figure 98 pictorially presents the pressure mapping data comparing results from the different sled tests. Due to synchronisation differences between tests the pressure mapping snap-shots show general comparative trends for common time intervals. All shades and colours are equivalent for all tests. These clearly show changes in pressure distribution from one time period to another and between dummies. The volunteer pressure maps indicate a relatively flat pressure contact with the seat with higher-pressure areas corresponding to the shoulder/scapula and at the pelvic wings. The Hybrid and THOR dummies show high pressure areas along the spine with some late pressure build up around the pelvis. The BioRID dummy shows a relatively flat pressure mapping much closer to the volunteer. The lateral pressure line with the BioRID is due to the instrumentation umbilical cable, which was placed behind the dummy.

Most dummy tests were repeated more than once (BioRID 4, Hybrid III with TRID neck 2, Standard THOR 1 and THOR with EuroSID neck 2). The volunteers were only impacted once at the reported velocity. Although only one sequence is shown for each configuration good repeatability was noted, even between volunteers where larger variability might have been expected.

4.2.5 Musculature

Surface electromyography (EMG) measurements were taken during the volunteer testing. Even though the volunteers were ‘inactive’ when impacted, pre-impact muscle activity was noted as they were audibly aware that an impact was imminent. In some muscle groups in a few tests background ambient EM noise largely obliterated any detectable changes in EMG signals, reducing the quality of the observations. Following the actual impact, muscle activity above the ambient was noted. The timing of this activity was compared to the transducer time histories. From the recordings that were considered to be valid it was found that involuntary responses in some of the major muscle groups in the neck were rapid enough potentially to affect head and neck motion during the impact. The sternocleidomastoideus muscle group, according to the EMG recordings, was the most active muscle group during the impact. Initial activity was noted at about 60ms and peak activity at about 80ms. There is evidence to suggest that both the trapezius and the erector spinae also become active during the whiplash motion at similar times. In one subject one of the trapezius muscles activated much earlier (within 10ms) and its opposite muscle at about 30ms. Compared to the other tests this was an outlier observation but one might hypothesise that the subject may not have been facing fully forward at the time of impact and these muscles fired very quickly in order to stabilise the head from lateral or torsional movement. These observations support the comments of other researches, that all the neck muscle groups work in unison in a pure perpendicular rear impact, Szabo. [9]

It is not easy to come to firm conclusions on the potential influence of muscle activity and the risk of WAD injury, based on these limited EMG evaluations. The results suggest that muscle activity could be a factor in stimulating or controlling whiplash injury. Muscle activity does occur very early in the impact suggesting that it is not a slow unimportant process. This observation suggests that whiplash studies with cadaveric material may be misleading, due to lack of muscle tone. The current study does not give any understanding of what physiologically stimulates the very rapid EMG reactions.
### Figure 98 Typical pressure map data for volunteers and dummies

<table>
<thead>
<tr>
<th>TIME (ms)*</th>
<th>0</th>
<th>400-432</th>
<th>440-472</th>
<th>480-512</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volunteer A</td>
<td><img src="image1.png" alt="Image" /></td>
<td><img src="image2.png" alt="Image" /></td>
<td><img src="image3.png" alt="Image" /></td>
<td><img src="image4.png" alt="Image" /></td>
</tr>
<tr>
<td>Volunteer B</td>
<td><img src="image5.png" alt="Image" /></td>
<td><img src="image6.png" alt="Image" /></td>
<td><img src="image7.png" alt="Image" /></td>
<td><img src="image8.png" alt="Image" /></td>
</tr>
<tr>
<td>BioRID**</td>
<td><img src="image9.png" alt="Image" /></td>
<td><img src="image10.png" alt="Image" /></td>
<td><img src="image11.png" alt="Image" /></td>
<td><img src="image12.png" alt="Image" /></td>
</tr>
<tr>
<td>Hybrid III with TRID neck</td>
<td><img src="image13.png" alt="Image" /></td>
<td><img src="image14.png" alt="Image" /></td>
<td><img src="image15.png" alt="Image" /></td>
<td><img src="image16.png" alt="Image" /></td>
</tr>
<tr>
<td>THOR with EuroSID neck</td>
<td><img src="image17.png" alt="Image" /></td>
<td><img src="image18.png" alt="Image" /></td>
<td><img src="image19.png" alt="Image" /></td>
<td><img src="image20.png" alt="Image" /></td>
</tr>
<tr>
<td>THOR with EuroSID neck and no head restraint</td>
<td><img src="image21.png" alt="Image" /></td>
<td><img src="image22.png" alt="Image" /></td>
<td><img src="image23.png" alt="Image" /></td>
<td><img src="image24.png" alt="Image" /></td>
</tr>
</tbody>
</table>
Figure 98 notes

* Times quoted are general time windows. They do not have the same start time as the transducer data thus time comparison with the transducer data should not be made.

** It is thought that the pressure line across the BioRID at the top of the lumbar is due to the transducer cable loom and not a dummy attribute.

It is noted, when examining time history acceleration measurements for both dummies and volunteers, that the first impact is felt in the lower body at the pelvis. Acceleration pulses are then sequentially observed in the thorax, neck and head. Based on this time sequence one might hypothesise that the early neck muscle activity might be stimulated from the pelvis, the first body part to be physically excited. One might also hypothesise that controlling pelvis acceleration might be a way of attenuating the speed or magnitude of upper body muscle activity.

5 General discussion

This report does not attempt to review injury mechanisms or injury risk. Kinematic comparisons are possible between volunteers and dummies and appropriate design target corridors and comparisons made, but these are not included in this paper.

The study has developed a set of transducer-based design target corridors for a stylised vehicle impact test, which can be readily reproduced and used to evaluate any new or improved rear impact test dummy.

The study has supported the hypothesis that neck muscle activity is rapid and that it could influence neck motion during a whiplash injury producing rear impact. If muscle activity does contribute to whiplash injury risk then reducing the initial stimulus could be an important factor in reducing injury risk and severity. These findings need to be further investigated as some of the EMG observations related to the erector spinae and trapezius groups were masked by background noise.

Examining the acceleration time histories one can observe, from the volunteer data, that the first sensing of the impact is seen at the pelvis followed by the thorax (T1) and then rapidly by the head.

There is no simple assessment methodology to indicate which is the best dummy when comparing the response of different ones against design target corridors. Factors that need to be taken into account are amplitude, time and rate of application. From the previous discussions and observations on the timing of the various parameters pelvis responses appear to be the first to detect the impact, as might be expected, thus it is possibly the first part of the dummy that should respond correctly. The BioRID can assess pelvic acceleration but the data are not available in this analysis. Peak acceleration levels are similar and of the correct magnitude but the onset rates for the dummies are too rapid. None of them gave the same two level response observed with the human volunteers. In terms of time domain sequence T1 is the next body part that responds to the impact. In terms of fore and aft acceleration (Figure 95), the BioRID appears to be the most human like in terms of rate but the peak value is slightly too low. The Hybrid III and THOR derivatives are too stiff but their peak values are closer to the corridor. All of the other dummies appear to behave in a similar manner (Figure 96).

Assuming that the pelvis and T1 response are correct then the head will be influenced by the characteristics of the neck. Head vertical acceleration (Figure 93), is quite variable across the dummies. BioRID is the only dummy to respond in the same way as the volunteers with the initial 1g negative pulse. It then tracks the performance target corridor quite well. The three other responses are deemed to be poor. Head fore and aft acceleration is again variable across the four dummy derivatives. The standard THOR appears to respond fairly well. The BioRID pulse is similar but delayed by about 20 ms. When examining the head resultant acceleration the Hybrid III with TRID
Dummy Development: Spinal Injuries

neck responds poorly. The standard THOR dummy is fairly good, but about 10 ms too early. The BioRID in shape is fairly good but again delayed by 15-20ms.

From the dummy parameters presented in this paper it seems that the Hybrid III TRID dummy has a poor response compared to the others with the BioRID dummy yielding the best response, although somewhat delayed at the head, suggesting that it might be slightly too weak initially in the neck.

It is important that a rear impact dummy should load the seat back in a human like manner. Thus pressure mapping time history or occupant to seat back interaction must be biofidelically correct. Correct pressure loading could also be important if pressure is being used to assess the size, shape and location of an occupant if it is being used to activate injury reducing counter measures. Comparison between dummies and humans is again subjective. Although only two pressure maps are presented from the volunteer data surprisingly good similarity is observed. Comparing these maps with the dummies clearly shows that the Hybrid III thorax is a poor surrogate for the human where the rigid spine box construction of the dummy is apparent. The THOR thorax is slightly better but again it shows the rigid nature of the spine. The BioRID appears to be a significantly better surrogate not having the very high-pressure areas noted with the rigid spine dummies.

From the volunteer pressure mapping data it is suggested that it is important that a rear impact dummy should have a generally compliant uniform back stiffness.

The observations made in this report compare dummies to volunteers at one non-injury level of rear impact severity, 7km/hr. It is important that protection systems are not sub-optimised to a one severity condition and that they will adequately protect occupants at higher severities. It is not possible to test humans at injury sustaining levels but good validated computer models of a human could be used to study higher velocity events and possible injury mechanisms. Future research should build on the current study making use of advanced modelling techniques with further dummy testing to try and quantify performance at higher levels.

6 Programme Review and Conclusions
1. Ten volunteer tests have been carried out at a sub-injury level to derive human responses in a low severity rear impact.
2. A set of dummy design corridors has been developed for five dummy attributes, according to the procedures developed by EEVC Working Group 9.
3. The test procedure is a simple one based on the ECE Regulation 44 test sled and a second impacting trolley.
4. Pressure mapping has been undertaken with both volunteers and a range of dummies. Good repeatability was observed as well as large differences between the dummies evaluated and humans under identical test conditions.
5. Four dummies have been tested, the Hybrid III with the RID neck, the THOR with its standard neck and with the EuroSID neck and the BioRID. The BioRID dummy appeared to be the best human surrogate.
6. EMG studies have been carried out with the volunteers suggesting that all muscles in the neck complex operate in unison and within the time frame of possible injury development.
7. Both the BioRID and THOR dummies have been further developed from the versions tested within this programme. Further evaluations may be necessary to assess possible improvements or changes.
8. Other kinematic design response corridors should be developed from the volunteer data and compared to the dummies.
7 References
1. AAAM, The Abbreviated Injury Scale (AIS), American Association of Automotive Medicine.
4. CCIS Phases 4, 5.
Acknowledgements
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