Development and validation of a C5/C6 motion segment model

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DEVELOPMENT AND VALIDATION OF A C5/C6 MOTION SEGMENT MODEL

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A thesis submitted in fulfilment of the requirements for the degree of Doctor of Philosophy

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ABSTRACT

There is a large body of work investigating whiplash-associated injury in motor vehicles and its causation. Being unable to detect the actual injury and having to use the symptoms of the sufferer as a surrogate has made progress in understanding the injury causation slow. Still lacking are the causal relationships between the biomechanical load on the vehicle occupant in the crash, the resulting loading on the neck and the actual injuries suffered. The optimisation of the design of vehicle safety systems to minimise whiplash needs a better understanding of human tolerance to these injuries.

This thesis describes the development of a mathematical multi-body C5/C6 motion segment model to investigate the causation of soft-tissue neck injury. This model was validated with available static *in-vitro* experimental data on excised motion-segments and then integrated into the existing, validated multi-body human head and neck model developed by van der Horst, to allow the application of realistic dynamic loads. The responses and injury sensing capability of the C5/C6 model were compared with available data for volunteers and cadavers in rear impacts.

The head and neck model was applied to the investigation of a group of real rear impact crashes (n = 78) of vehicles equipped with a crash-pulse recorder and with known post-crash injury outcomes. The motion of the occupants in these crashes had previously been reconstructed with a MADYMO BioRID II dummy-in-seat model validated by sled testing. The occupant T1 accelerations from these reconstructions were used to drive the head and neck model. The soft-tissue loading at C5/C6 of the head and neck model was analysed during the early stage of the impact, prior to contact with the head restraint. The loading and the pain outcome from the vehicle occupants in the actual crash were compared statistically.

For the longer-term whiplash-associated pain outcomes (of greater than 1 month duration) for these occupants, the C5/C6 model indicated good correlation with the magnitude of the shear loading on the facet capsule. In lower severity impacts, the model result supported a second hypothesis of injury to this motion segment: facet surface impingement.

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Finally, the patience of my family Anne, James and Stuart for putting up with the demands on my time required in bringing this study to completion.

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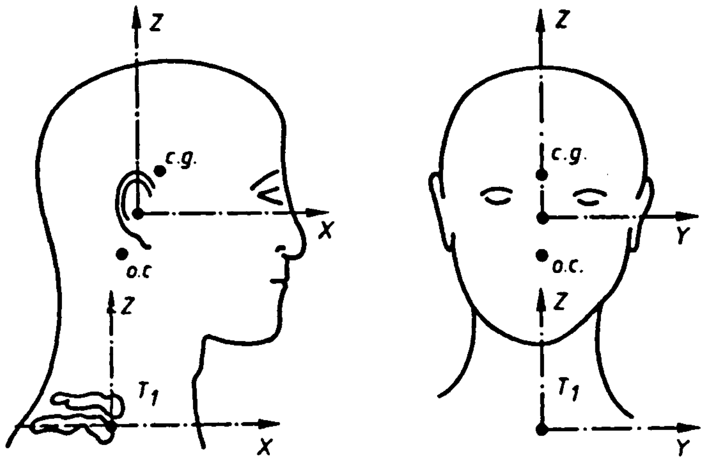
NOTATION

a	Acceleration, m/s^2
AF	Anulus Fibrosus
AIS	Abbreviated Injury Scale
ALL	Anterior Longitudinal Ligament
ATD	Anthropomorphic Test Device/Dummy
Axis System	The standard axes used for the anatomical coordinate system in the study, as used by Wismans et al. (1985).



BioRID	Biofidelic Rear Impact Dummy
C	Cervical vertebrae, eg C5 is the 5th cervical vertebra
CofG, cg	Centre of gravity
CPR	Crash pulse recorder
CSF	Cerebrospinal fluid
CT	Computer Tomography
DRG	Dorsal Root Ganglion
ε	Relative elongation, strain
E	Young's Modulus, Modulus of Elasticity, MPa
EMG	Electromyography
F	Force, N
FC	Facet Capsule, facet joint, facet capsular ligament
FE	Finite element analysis methods
FL	Flaval Ligament
g	Acceleration due to gravity, 9.81 m/s

HIII	Hybrid III anthropomorphic test dummy
HARM	A measure of the actual cost of injury, in Australia this has been developed by Monash University Accident Research Centre
IAR	Instantaneous Axis of Rotation
ICR	Instantaneous Centre of Rotation
ISL	Interspinous Ligament
IV	Intervertebral
LF	Ligamentum Flavum
M	Moment, Nm
MADYMO	Mathematical Dynamic Model, impact biomechanics modelling software developed by TNO, the Netherlands
NCAP	New Car Assessment Program
NIC	Neck Injury Criterion for soft tissue neck injury in rear impacts proposed by Bostrom et al. (1996)
$\mathbf{N_{ij}}$	A frontal Neck Injury Criterion developed by Kleinberger et al. (1998b)
$\mathbf{N_{km}}$	Neck injury criterion based on the N_{ij} proposed by Schmitt et al. (2002)
NP	Nucleus Pulposus
PLL	Posterior Longitudinal Ligament
PMHS	Post-Mortem Human Subjects
OC, C0	Occiput, Occipital Condyles
RID	Rear impact dummy
SAHR	Saab Active Head Restraint
SSL	Supraspinous Ligament
STNI	Soft tissue neck injury
TRID	TNO rear impact dummy
V, *v*	Velocity, m/s
WAD	Whiplash associated disorder, the name for the various whiplash-associated injuries proposed by the Quebec Taskforce.

CHAPTER 1 INTRODUCTION

1.1 Problem Statement

Whiplash associated injury following a motor vehicle impact is a major source of third-party motor vehicle insurance claims and is a considerable cost to society. In Australia, the proportion of claims due to this type of injury varies between states depending on whether a 'no-fault' system operates as in Victoria and Tasmania, or the tort system, which governs the other states. Ryan and Gibson (1998) estimated that 30,000 new cases of whiplash-associated injury occur annually in Australia, with an incidence rate of about 167 per 100,000 of the population. The total cost of these cases was estimated at $540 million – just less than 10% of the estimated annual cost of all road injuries in Australia. Until recently, attempts to control the problem in Australia have included changes to the rules governing the criteria for making a claim, and concerted efforts to detect fraudulent claims. Lowering the cost of injuries to the community requires the best use of current knowledge by means of a multi-disciplinary approach to the problem.

There is a vast body of work investigating whiplash associated injury (or disorder WAD, as it is sometimes called) and its causation at all levels, including crash investigation, clinical studies, post-mortem studies, biomechanical testing, and simulation. The slow progress in understanding the injury causation is due to the fact that we are yet unable to detect the actual injury, relying on the symptoms of the sufferer as a surrogate. The knowledge gained from these areas of study has significantly advanced our understanding of both the causes and treatment of such injuries.

Investigations into crashes with whiplash associated injury outcomes have resulted in evidence regarding the circumstances and occupant types most likely to be affected (States, Balcerak & Williams 1972; Morris & Thomas 1996). Crash researchers have concluded that:

- The severity of whiplash symptoms is related to that of the impact, particularly for rear impacts (Krafft et al. 2000);
- Women are more likely to be affected than men (Morris & Thomas 1996); and,

- Being aware of the impending impact has a protective effect at the time of the injury (Ryan et al. 1993).

Crash investigation has also added insight into the roles of the vehicle and seat structures in the causation of WAD. Researchers have made the following observations:

- Wearing a seat belt appears to increase the likelihood of injury (Morris & Thomas 1996; Viano 1992a);
- Seat back characteristics have a greater effect than the vehicle structure itself (Haland et al. 1996); and,
- Attempts to regulate effective head restraints have been ineffective (Kahane 1982).

Clinically, it has been found that many of the traditional treatments have been ineffective in reducing the symptoms (Bogduk 1998). While the lesions are indiscernible using normal radiographic techniques, the pain symptom patterns are characteristic of persons with chronic spinal pain (Dwyer, Aprill & Bogduk 1990). In a major breakthrough, Lord, Barnsley and Bogduk (1993) rigorously demonstrated that for more than 50% of patients suffering chronic whiplash symptoms (those persisting for more than six months), the pain originates from within the cervical facet joints (zygapophysial joints). This has been achieved through the investigation of facet joint pain by means of controlled, double blind, differential, diagnostic anaesthetic blocks. The work strongly supports that patients with persisting symptoms after whiplash have physical lesions and real pain (Barnsley, Lord & Bogduk 1998).

Post-mortem studies on the necks of crash victims have revealed the existence of a wide range of soft-tissue neck injuries that may be connected with the pain symptoms of whiplash associated injury (Taylor & Taylor 1996). Further insight into the causation of the injury found in autopsy has been gained by means of:

- *In vitro* testing of isolated cervical spine motion segments (Siegmund et al. 2000b; Winkelstein et al. 2000);
- Dynamic testing of intact head and neck complexes (Yoganandan, Pintar & Kleinberger 1998); and
- Dynamic testing of whole cadavers (Deng et al. 2000a).

In complementary studies, the dynamic head and neck motion during a whiplash event, and the effects of the active muscles have been investigated by volunteer testing to near injurious levels (Szabo & Welcher 1997). Recent volunteer testing using high-speed radiography has been useful in defining normal head and neck motion in a rear impact (Ono et al. 1997).

The greater availability of accurate test data on the mechanical properties of the soft-tissue neck components has supported the development of mathematical models of the human head and neck, with acceptable biofidelity in their dynamic responses. Some of these models have the capability of including the effects of active muscles (de Jager et al. 1996). In their present stages of development, these models can simulate the kinematic response of the head and neck to impact loading, but cannot adequately simulate the detailed soft-tissue injury mechanisms (van der Horst et al. 2001).

Still lacking are the causal relationships between the biomechanical load on the vehicle occupant in the crash, the resulting loading on the neck, and the actual injuries suffered. The primary users of such information – vehicle manufacturers worldwide, are presently experiencing a period of rapid change driven by two major influences: increasingly stringent safety requirements and rapid developments in the technology available to meet them. To optimise the design of vehicle safety systems to minimise whiplash injuries, automotive engineers need to improve their understanding of the causes of the injuries, the injury mechanisms, and human tolerance to the injury. The relationship between the crash loading applied to the vehicle occupant and the risk of injury to the neck needs to be better defined.

1.2 Aims of the Study

The aims of this study are:

- To develop a mathematical C5/C6 motion segment model that can be used to investigate the causation of soft-tissue neck injury in rear impacts;
- To validate the predicted motions and injury of the C5/C6 model with available static *in-vitro* experimental data;
- To integrate the C5/C6 model into an existing, validated human neck model for the application of dynamic loads;

- To validate the predicted motions and injury of the C5/C6 model with available dynamic experimental data from volunteers and cadavers in rear impacts;
- To demonstrate the injury-sensing capabilities of the C5/C6 model by applying it to the analysis of a group of rear impact crashes with known pain outcomes; and,
- To investigate the correlation of early soft-tissue injuries to the neck with existing neck injury hypotheses.

1.3 *Research Methodology*

The various stages of this study are summarised in Figure 1.1 below.

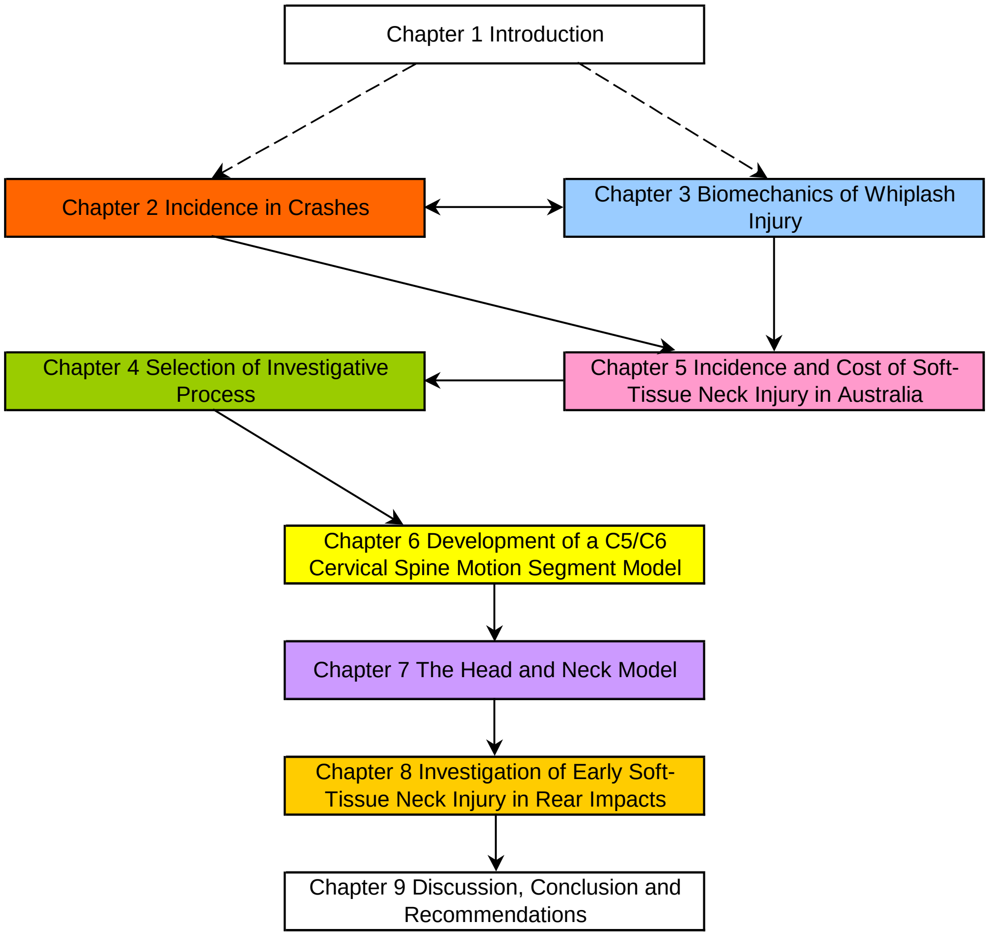


Figure 1.1 Diagram showing the various stages undertaken in the study and the corresponding chapters in this thesis. The chapters 5, 6, 7 and 8 contain the new work. The colour coding is followed through the thesis.

A thorough review of the published literature was conducted in the early stages. Chapter 2 is a review of selected field accident studies and vehicle engineering interventions that have been used to mitigate whiplash injuries to date. This is to place into context the motor vehicle and occupant related factors involved in the incidence and characteristics of soft tissue neck injury. Chapter 3 reviews the anatomy of the neck required to model the C5/C6 motion segment; the available biomechanics of injury from clinical and post-mortem studies; relevant experimental studies based on both cadavers and volunteers; and, various hypotheses of soft-tissue neck injury mechanisms.

A preliminary investigation into the crash-related characteristics of neck injury associated with chronic pain is reported on in Chapter 4. The results of the investigation are used to make estimates of the incidence and cost of whiplash injury in Australia, which are not directly from insurance claim based data.

Chapter 5 reviews the available mathematical models of the head and neck, and presents the background of the modelling approach used for the development of the investigative model. The basis for the development was a MADYMO multi-body human head and neck model, developed by van der Horst et al. (2001). This well developed model has already been validated, has acceptable kinematic biofidelity and includes active muscle capability.

A mathematical C5/C6 motion segment model was developed in Chapter 6, based on the van der Horst human head and neck model (van der Horst et al. 2001). The new motion segment model incorporated recent descriptive work on the anatomy of the neck disc and the results of the most recent biomechanical experiments. It was validated using the results of recent *in-vitro* testing of cervical spine motion segments reproducing whiplash-type loading.

The integration of this new mathematical C5/C6 motion segment model with the van der Horst human head and neck model is described in Chapter 7. The response of the new head and neck model was verified dynamically with the results of published tests of volunteers, whole cadavers and intact cadaver head and neck complexes.

The new head and neck model was then applied to the investigation of a group of real crashes (n = 78) of vehicles equipped with a crash pulse recorder and known post-crash outcomes in the form of pain duration. These crashes had been reconstructed with a MADYMO BioRID II dummy and seat model, which was validated by sled testing (Kullgren et al. 2003). The T1 acceleration from these reconstructions was used to drive the head and neck model. The soft tissue component loading of the C5/C6 motion segment was analysed. The motion segment loading and the whiplash-associated outcomes for the vehicle occupants in the crash were analysed statistically.

1.4 Ethics

Permission was granted by the NSW Privacy Commission to access the Police Accident Records of patients from the Cervical Spine Research Unit, Newcastle, for the investigation in Chapter 4.

CHAPTER 2 INCIDENCE IN CRASHES

2.1 Introduction

Lowering the cost of injury to the community increasingly requires multidisciplinary approaches to the problem. For the automotive engineer to be able to design vehicle safety systems to minimize injuries such as whiplash, an understanding of the causes of the injuries, the injury mechanisms, and human tolerance, is required.

Whiplash injury is examined in this Chapter from the perspective of an automotive engineer. The main areas covered and the context of this chapter with in the thesis is shown in Figure 2.1. The results of several important field accident studies on the incidence of soft-tissue neck injury and the significant crash and victim parameters identified are summarised. This is extended to the vehicle factors known to be involved in injury causation: the structural response, the seat design, the head restraints fitted and the interaction with a restraint system. A brief overview of the design methods currently being used by engineers to minimise whiplash-associated disorder (WAD) are presented. Further, the regulatory and consumer information interventions are also outlined. Finally the need for an accepted dynamic test methodology is discussed, including the identification of the available anthropomorphic test devices (ATD).

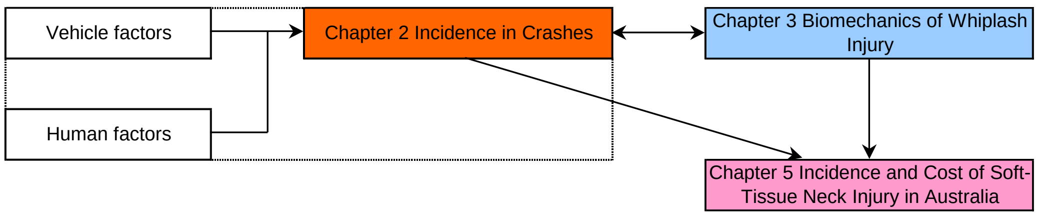


Figure 2.1 The main areas covered and the context of this chapter within the thesis.

2.2 Accident Studies

The analysis of accident data has been useful in providing insight into those human characteristics and vehicle factors, which influence the incidence of whiplash injuries in vehicle crashes. Such research projects are difficult and expensive to carry out if enough precision in the data and a sufficient number of cases are to be gathered to achieve statistically significant results. The following studies have been important in the area and give insight into the characteristics of soft-tissue neck injury associated with whiplash.

One of the earlier field accident studies, which is often referred to and still relevant, is that by States, Balcerak and Williams (1972). This study included all rear-end crashes reported in Rochester, New York over a three-month period. The cases were followed-up through special police forms, telephone interviews and mail questionnaires. Additionally, approximately every 20th case was provided with vehicle photographs and medical examinations. A total of 691 rear-end crashes were collected, and the following observations were made (none of which were regarded as statistically significant):

- Whiplash was the principal injury to occupants of struck vehicles, totalling 99.3% of all injuries and occurring at 38% frequency;
- Head restraints were effective for both the driver and right front passenger. The whiplash injury frequency was 37% for head restraint-equipped vehicles and 42% for vehicles without fitted head restraints;
- The benefit of head restraints appeared to be more noticeable for female occupants, for whom the whiplash injury frequency was 38% in vehicles with head restraints and 51% in those without;
- Occupants with fixed head restraints were better off and had a whiplash injury frequency of 13% compared with an overall rate of 16%;
- Seat back damage showed no effect on whiplash injury frequency;
- The overall whiplash injury frequency for females was 44%, and 35% for males;
- The whiplash injury frequency for rear and centre front seat occupants was very low at 22%, which was associated with the high usage by younger and smaller occupants;
- The use of lap belts seemed to increase the whiplash injury frequency.

Ryan et al. (1993) made a follow-up study of WAD. Thirty-two individuals with neck strain following a car crash were interviewed and given a physical examination, soon after the crash and again after six months. Each case vehicle and crash site was inspected and the crash reconstructed. The severity of the crash was assessed by using the maximum vehicle residual deformation to estimate the change in velocity.

In 22 cases, the impact originated from the rear; the remainder were from the front or side. Neck strain occurred as a result of low severity impacts, with six cases having a velocity change of less than 10 km/h and eight cases resulting in a maximum vehicle residual deformation of less than 50 mm. For rear impacts, maximum residual deformation and velocity change were positively associated with the measures of neck strain severity. Six months after the impact, 19 (66%) of the 29 subjects available for follow-up still showed evidence of injury. There was no statistically significant association between either measure of crash severity and persistence of neck strain at six months. Subjects who were aware of the impending impact had less severe symptoms initially, and were much less likely to experience persisting problems.

A more recent study by Morris and Thomas (1996) on soft-tissue neck injury was based on the Cooperative Crash Injury Study (CCIS) in the UK, which commenced in 1983 and ended in 1992. This database contains 11,866 occupants and 6,973 crashed vehicles. They found the following:

- The incidence of soft tissue neck injury (STNI) for all accident types was found to have been increasing steadily over the data collection period, from 11.2% in 1984 to 22.8% in 1991;
- There was a distinct gender effect for all accident types with the soft-tissue neck injury rate for females increasing at a faster rate than for males. The rate for females had climbed from 14% in 1984 to 31% in 1991, while for males the rise was from 10% in 1984 to 18% in 1991;
- Soft-tissue neck injury occurred in all impact directions with an average injury rate of 16%, but for rear impacts the rate was 38%.
- Seatbelt use increased the overall likelihood of soft-tissue neck injury, with 20% of restrained occupants compared with 8% of unrestrained occupants sustaining neck injury. Seatbelts increased the injury likelihood for males but had no effect on females;
- Rear impacts resulting in neck injuries occur at a lower average speed of 32 km/h as compared to 36 km/h for those without injuries. The critical speed region for these impacts was apparently between 17.5 and 27.5 km/h;

- Occupant and seating characteristics have some effect, the most marked being that neck injured females appear to be using higher head restraints, are younger and lighter, as shown in Table 2.1;
- The type of head restraint showed no differences in effectiveness; and,
- There was some tendency for reduced STNI when the seat back was deformed.

Table 2.1 The relationship between occupant height, weight and age in relationship to seat characteristics in rear impacts, Morris and Thomas (1996).

Characteristic	STNI mean	No STNI mean
Males		
Head restraint height (cm)	77.5	77.7
Age (years)	41.5	38.9
Weight (kg)	79.7	74
Height (m)	1.79	1.75
Females		
Head restraint height (cm)	77.8	75.9
Age (years)	36	43.8
Weight (kg)	61.7	64.4
Height (m)	1.63	1.61

Temming (1998) analysed the Volkswagen accident database to investigate the significance of human factors, such as gender, age, height and weight, on the frequency of occurrence and risk of suffering whiplash injuries in rear collisions. It was shown that for all combinations of parameters (belted, belted with multiple collisions, and belted with single rear collision), female occupants had a higher risk of neck injury (cervical spine distortion) than males by a factor of between 1.9 and 2.4, Table 2.2 and Table 2.3.

Within the age range of 18 to 57 years (88.7% of cases), there was a general reduction in frequency of occurrence of whiplash injuries with increasing age (from about 30% to about 5%). Risk of neck injury ranged between 20 and 30% for all age groups older than 18 years, with no clear trends noted. There were no clear trends regarding the influence of body height either. The trends relating risk by body weight were the same as by gender. Temming suggested the following explanations for the gender-specific differences:

- Men have stronger neck muscles, as indicated by the ratio between head volume and neck cross-sectional area;

- Women have longer necks and larger heads relative to their own body weight than men;
- Women sit farther forward in their seats than men; and
- Women may be more inclined to file an insurance claim for whiplash than men.

Table 2.2 Occurrence of soft tissue neck injury (distortion) in the Volkswagen data investigated by Temming (1998).

Collision damage location	All passenger car occupants	Belted occupants	Belted occupants, injured	Belted occupants with neck injury	Belted occupants with STNI
All collisions	14 276	10 349	4 570	1 621	1 229
Frontal	9 064	6 650	2 773	931	700
Right side	1 368	940	432	106	71
Left side	1 635	1 146	574	171	119
Impact side occupants	1 482	1 067	576	173	118
Opposite side occupants	1 386	953	374	88	58
Multiple collisions (rear most severe)	1 744	1 295	594	358	297
Single rear-end	1 005	836	330	226	186
Rollover	386	281	216	56	46

Table 2.3 Risk by gender of neck injury (distortion) in the Volkswagen data investigated by Temming (1998).

	Total	Male	Female	M/F ratio
Occupants	886	482	337	58.9%/41.1%
Injured occupants	186	73	113	39.2%/60.8%
Injury Risk	22.2%	15.1%	33.5%	

A Swedish study, Krafft (1998), found that women with whiplash injuries are more likely to develop long-term symptoms of whiplash than are men with whiplash. In this study fifty-five percent of the women who sustained whiplash injuries went on to develop longer term symptoms compared with only 38 percent of men.

2.3 Engineering for Rear Impact Injury

As shown in Figure 2.2, engineering intervention aimed at mitigating whiplash injury can be implemented either before the crash, by reducing the likelihood of the crash, or during the crash, by minimizing the injury risk of the crash. The necessary engineering design changes can be applied to the vehicle structure, to change its performance during the crash, or to the individual components of the vehicle such as seats. The process of designing to minimise an injury requires detailed understanding of the likely injury mechanisms and the availability of appropriate injury tolerance data.

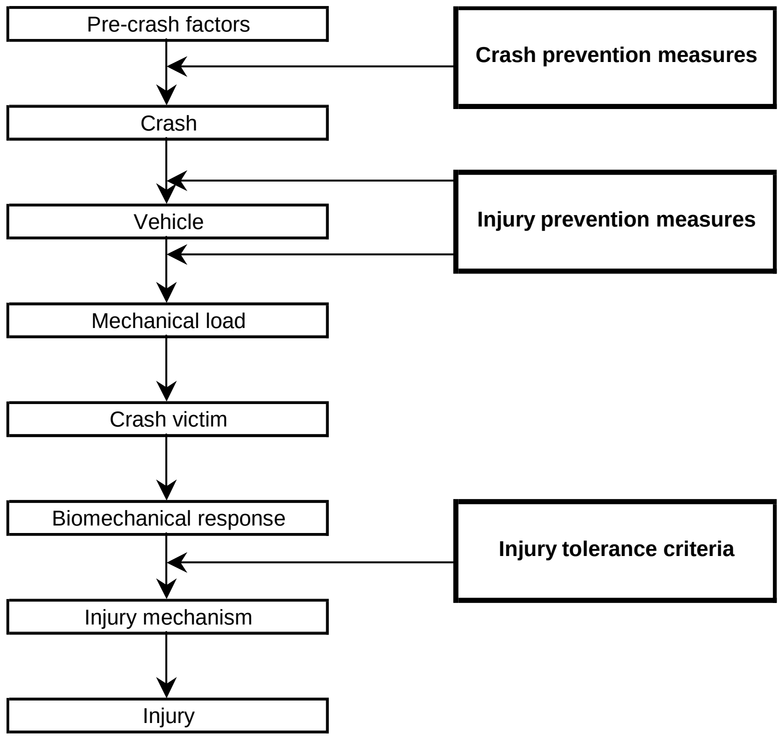


Figure 2.2 The biomechanical injury/load model showing how engineering interventions can reduce crash related injury, after Wismans (1995).

2.3.1 Crash Prevention

Prevention of the crash includes vehicle design features aimed at reducing the likelihood of the crash occurring at all. Road vehicles have undergone numerous improvements in the performance of braking systems and consistency of operation (such as the user ergonomics, road holding and wet-weather performance) since their inception. The areas now open for improvements in crash prevention are mainly tied to developments in intelligent vehicle systems. The crash prevention area, which has the potential to be very effective in reducing whiplash associated disorder is peripheral to the aims of this thesis.

2.3.2 Injury Prevention

Injury prevention includes measures designed to reduce injury once the vehicle has been involved in a crash. These interventions are aimed at reducing the mechanical load applied to the victim during the crash. This is achieved by changing the response of the vehicle to the crash. An example is the incorporation of crushable zones in the rear structure of a vehicle for the purpose of reducing the severity of the crash pulse generated during a rear impact.

The way in which the load from the vehicle structure is transmitted to the victim during a crash interaction may be reduced by:

- Promoting ride down;[1]
- Spreading the energy of the impact in a manner appropriate to the body region loaded;
- Managing the rate of energy absorption by the use of padding material; and
- Preventing excessive relative motions between the body segments of the crash victim.

Applying these principles to the protection of an occupant in a car during a low speed rear impact requires several improvements to typical vehicle designs. The rear structure of the vehicle must be designed to minimize the severity of the deceleration pulse without compromising the structural integrity in a high velocity crash. This must be combined with a seat system designed to promote effective ride down of the occupant of the rear impact crash pulse, while minimising misalignment in the neck and the possibility of rebound following the impact.

The injury tolerance level is the magnitude of loading at the threshold of a specified level of injury severity to a human[2]. There are large variations of tolerance to neck injury between individuals, depending on such factors as gender, age, and anthropometric and physiological differences (Yoganandan, Pintar & Cusick 1997). Injury tolerance applies specifically to living humans and injury tolerance values are of limited use to engineers, as they require human subjects (who will be injured) for testing. The assessment of a specific design during the product development process requires the use of a test methodology based on a surrogate, usually a mechanical crash test dummy, and associated injury criteria. For ethical, reliability and consistency considerations, crash test dummies are used in regulatory tests.

[1] Ride down is the term used to describe the coupling of the occupant to the vehicle structure with a safety restraint system to minimize the acceleration levels applied to the occupant, by using as much of the deceleration distance afforded by the collapse of the front structure of the vehicle as possible.

[2] Injury severity level is usually measured by means of the Abbreviated Injury Scale (AIS) (AAAM 1990).

2.4 Vehicle Factors

2.4.1 Effect of Head Restraints

In a study for the U.S. Department of Transportation (DoT), Kahane (1982) evaluated the effectiveness of head restraints complying with the original Federal Motor Vehicle Safety Standard (FMVSS) 202 in reducing the overall risk of injury in rear impacts. The study estimated this to be 17% for integral restraints and 10% for adjustable restraints. The effectiveness of head restraints in reducing whiplash could not be determined due to lack of recorded data for AIS1 neck injury. The in-use median height for adjustable head restraints was determined to be less than 660 mm as compared to over 710 mm for the integral restraints. When combined with the erect seated height to the base of the skull of a 50th-percentile U.S. male of 700 mm, Kahane hypothesized that head restraint heights above 700 mm should give full injury prevention benefits.

This study is supported by other later studies. Nygren, Gustafsson and Tingvall (1985) found that the use of head restraints decreased the risk of neck injury in a rear-end collision by approximately 20% on average. Fixed head restraints gave a 24% reduction and adjustable ones gave a 14% reduction.

In a study of 33 occupants of Volvo cars, Olsson, Bunketorp and Carlsson (1990) found that neck symptoms lasted longer for increasing horizontal distance between the head and the head restraint. The authors found that a head restraint back-set of more than 10 cm from the back of the head correlated with an increased risk of neck injuries in rear impacts. They also suggested that the risk of whiplash injuries could be decreased by using controlled plastic deformation of the seat back rest integrated with the head restraint, to diminish the relative motion of the head and trunk.

A Dutch study of front-seat occupants found that roughly 40% of male occupants and 50% of female occupants had adjusted their head restraints to the "correct" height – when the top of the head restraint is at least level with the ear or higher (van Kampen 1993). However, it should be noted that a seat that meets the minimum height requirement of the European vehicle regulations only enables correct height adjustment for males shorter than the 25th-percentile and for females shorter than the 90th-percentile of the Dutch population.

2.4.2 Vehicle Structural Effects

Krafft et al. (2000) studied the influence of crash severity in rear impacts that led to short-term or long-term WAD (i.e. injuries lasting less or more than a year, respectively). The study was completed in three stages.

In the first stage, Volvo 240 model vehicles were tested with and without towbars fitted. The fitting of a towbar to a car requires extra structure to be added to the vehicle behind the rear axle and has a stiffening effect. This change to the vehicle structural stiffness at the rear gives a simple method to investigate the effects of the vehicle structure on the incidence of neck injury. Full-scale crash tests were run at 25 km/h to give a change in velocity of 15 km/h to the struck car. It was found that the acceleration due to the rear impact in the towbar-equipped car was higher (9.6 g) than in the car without a towbar (8.0 g). Further, the occupant of the towbar-equipped vehicle was subjected to higher accelerations. The dummy in the towbar-equipped car had a maximum T1 acceleration of 8.9 g, compared to 6.7 g in the car without a towbar.

In the second part, data was collected from the Folksam insurance company between 1990 and 1993. Occupants of Volvo 240, Volvo 700 and Saab 900 (1979-1993) model cars, with a random sample of AIS1 neck injuries, were selected. Cases with short-term consequences, where at least one occupant reported an injury after the accident, were represented by 233 reported car crashes. Cases with long-term consequences, where at least one occupant sustained long-term consequences in the struck vehicle, were represented by 75 reported car crashes. There was a 22% greater risk of long-term WAD consequences in a struck car with a towbar than in one without.

Finally, an additional 28 rear impacts in which a crash pulse recorder (CPR) was mounted under the driver's seat, were collected from Folksam. The vehicles were inspected, the seat-back deformations were investigated, medical notes and questionnaires were obtained, and possible medical symptoms were followed up at least 6 months after the collision. Fifteen occupants in 11 collisions did not suffer symptoms to the neck. The maximum acceleration levels in these collisions were 6 g at most. In fifteen collisions, 20 occupants sustained short-term consequences. In these cases, the maximum accelerations did not exceed 10 g. Two occupants from the 15 collisions sustained long-term injuries. In these cases, the accelerations reached 13 and 15 g.

From the study, Krafft et al. (2000) concluded that the stiffness of the vehicle structure influences the severity of WAD outcomes. In rear impacts, cars fitted with towbars displayed higher peak accelerations than those without.

2.4.3 Vehicle Seat Response Effects

Foret-Bruno et al. (1991) analysed a French vehicle accident database containing 8,000 involved vehicles, and made the following conclusions:

- Deformation of the seat back reduced the incidence of cervical injury in rear impacts;
- Elastic rebound of the seat back following a rear impact leads to increased neck loading, which the authors demonstrated with a series of sled tests involving dummies and a cadaver.

Parkin et al. (1995) used the UK based CCIS crashed vehicle data. The authors looked specifically at the relationship between seat damage and AIS1 neck injury and found the following:

- Rear impacts only made up 6.0% of the total of 5,361 crashes studied.
- The seats with no damage became less frequent as the severity of the collision increased.
- An occupant was significantly more likely to suffer AIS1 neck injury if the seat was undamaged.
- The frequency of AIS1 neck injury was not related to impact severity, which may be related to the greater likelihood of the seat collapsing as the impact severity increased.

Haland et al. (1996) evaluated various seat designs, which were aimed at minimizing the motion between the upper and lower cervical spine (i.e. limiting the S-shape, which occurs in the neck, see discussion in Chapter 3) as a result of a rear impact. For this testing, the recently developed rear-impact dummy (RID) neck was used on a Hybrid III (HIII) dummy. It was found that:

- The Opel Corsa seat was worse than the Peugeot 205 seat because the neck motion during impact occurred at a faster rate;

- A strengthened standard seat performed worst because the limited deflection (40 mm) increased the velocity of the lower neck; and,
- A modified seat with an increased seat back deflection of 130 mm reduced the velocity of the lower neck.

Based on the loading on dummies in rear-impact tests, the authors concluded that seat responses were more important than the vehicle structural responses, in mitigating whiplash injuries.

Another Swedish study by Kraft et al. (2003) was also based on the Folksam insurance data, in this case of 554 occupants in 195 crashes. It was found that females seated in the rear seat were at significantly higher risk of injury in a rear impact than when seated in the front or driving. The effect was not as strong for males. This was found to be consistent with the rear seat being stiffer than the front seats of a vehicle.

2.4.4 Seatbelt Effects

In a series of sled tests, Viano (1992a) investigated occupant retention by the seat during rear impacts. Viano found that the retention of an unrestrained occupant was dependent on the degree of seat back deformation, which was in turn dependent on the severity of the impact. The limit at which an unrestrained occupant would be retained in his seat was found to be at 60° of seat back deformation. Beyond this point, the rearward acceleration of the test dummy was enough to overcome the friction involved, and caused it to ramp up the seat and over the head restraint. For a standard seat designed to meet FMVSS 202 criteria, this was observed at a 15.5 g peak acceleration pulse with a velocity change of 9.6 m/s. In another related study, Viano (1992b) demonstrated that lap belt use improved retention of the dummy in the seat in rear impacts and so reduced ramping. This was particularly evident in slightly offset rear impacts (15°), in which the pelvis was kept engaged with the seat.

The effects of shoulder-belt geometry in rear-end collisions were analysed by Krafft et al. (1996), by comparing the outcomes of 2- and 4-door Volvo 240 cars (1975-1994) and 3- and 5-door Saab 900 cars (1979-1993). In these vehicle models, the 2- and 3-door vehicles have the seatbelt shoulder anchorage mounted 27 cm and 23 cm further to the rear than their sedan counterparts, respectively. Accident data reported by the police to the Swedish National Bureau of Statistics was used in the study. It was found that the weights of both the struck and striking vehicles, the gender of the occupant, and the seatbelt geometry as indicated by whether the cars were hatchbacks or sedans (with the sedans fairing worse), all influenced the relative risk of AIS1 neck injuries. The authors concluded that the influence of the seatbelt geometry on the risk of AIS1 neck injury in rear impacts, added support to the hypothesis that rebound from the seat is an important part of the injury mechanism for whiplash-associated disorders.

2.4.5 Airbag Effects

Otte (1995) reviewed 41 motor-vehicle accidents in Hannover, in which airbags were deployed. He found that half of airbag inflations caused soft tissue neck injury. Otte concluded that inflation of an airbag induces an extreme motion of the head and cervical vertebrae, giving a higher risk of these whiplash injuries. Conversely another German research group (Langweider, Hummel & Müller 1996) found that the deployment of driver side airbags resulted in fewer soft tissue neck injuries. These researchers proposed that the interception of the head motion by the airbag prevented the hyperflexion of the cervical spine.

In a study for the Australian Transport Safety Bureau (ATSB), Morris et al. (2001) evaluated the effectiveness of the Australian Design Rule (ADR) 69, which is the full frontal barrier crash test requirement with restrained dummies. The case-control study of real crashes compared the injuries and HARM to occupants of vehicles equipped with and without SRS airbags. In terms of whiplash-associated injuries, it was found that 19% of drivers in the airbag cases (n = 291) suffered AIS1+ neck injuries, compared with 30% in the non-airbag cases (n = 141). The combination of seatbelts with or without airbags gave similar results: 19% of belted drivers with airbags (n = 253), and 31% belted drivers without airbags (n = 130) suffered AIS1+ neck injuries.

2.4.6 Seat Design Developments

Viano and Olsen (2001) evaluated the field performance of the Saab Active Head Restraint (SAHR) in reducing whiplash in rear crashes. Comparisons of single-event rear-end Saab crashes involving Saab 9-5/9-3 equipped with SAHR and Saab 9000/900 fitted with standard head restraints, over a period of 18 months. The SAHR reduced whiplash injury risks by 75±11%, from an 18±5% incidence in 85 occupants with standard head restraints to 4±3% in 92 occupants with SAHRs. No SAHR-fitted seats required repair or replacement after the crashes. The SAHR was found to be effective in reducing the incidence of medium to long-term whiplash injury in a sample of rear crashes in Sweden.

Recent research has investigated the effectiveness of these new head restraint and seat designs in reducing neck injury in rear impacts. An Insurance Institute for Highway Safety study was based on the claims data supplied by three of the major US insurance companies, Nationwide, Progressive, and State Farm (Farmer et al. 2003). The claims data were supplied by three of the major US insurance companies. Three different seat and head restraint designs were studied:

- <u>Improved geometry</u>, allowing the head restraint to be positioned closer to most occupants' heads. Ford adopted this principal in their Ford Taurus and Mercury Sable models between 2000 and 2002.
- An <u>active head restraint</u>, allowing the occupant's torso to sink back into the seat during a rear-end crash, and engage a mechanism in the seat back, which pushes the head restraint up and toward the back of the head. This design was adopted by Saab in 1997 as the SAHR system and in some General Motors and Nissan models.
- A <u>yielding seat back</u> reduces the forward acceleration of the torso in rear-end crashes. The Volvo WHIPS seat design includes a specially designed hinge for the seat back, which allows rearward movement to reduce forward acceleration, without collapse of the seat (Lundell et al. 1998). The Toyota whiplash injury lessening (WIL) system allows an occupant to sink farther into the seat back during a rear impact (Seizuka 1998).

Overall, neck injury claims have been reduced. The benefits have been greater for women than for men. There was also a 43% reduction in neck injury claim rates for the Saab, General Motors and Nissan models with the active head restraints and an 18% reduction in the Fords with improved geometry. The Toyota WIL system did not show any reduction in neck injuries.

2.5 Minimising Whiplash Injury

2.5.1 Static Head Restraint Requirements

Regulations aimed at minimizing neck injury have focused on the mandatory instalment of head restraints and the control of seat back stiffness, to reduce rearward head motion in crashes and prevent neck injury as a result of hyperextension. Typical of current regulations is the Federal Motor Vehicle Safety Standard, FMVSS 202, which has required that all passenger cars sold in the U.S. be fitted with head restraints in the front outboard seating positions since 1969. FMVSS 202 requires that one of the following conditions are met:

- The head restraint is at least 27.5 inches (700 mm) above the seat reference point when fully extended and the seat back must not deflect more than 4 inches (100 mm) rearward under a 120-lb (54.5 kg) load.
- The rearward angular displacement of the head reference line is limited to 45°, under a forward acceleration of the seat structure of 8g.

In the current European regulations, ECE 25.04, the minimum height requirement has been raised to 29.5 inches (750 mm).

The New Car Assessment Program (NCAP) started with full frontal crash testing of vehicles in the U.S. in 1978, as a means of evaluating the safety of vehicles for consumer information. An increase in the scope of this testing has ensued, with the inclusion of evaluation of head restraints fitted to vehicles (IIHS 2001). The rating is based on static measurements of the head restraint in its lowest position, with respect to a 50th percentile mannequin (Figure 2.3 and Table 2.4):

- The vertical position of the top of the head restraint with respect to the top of the head (V).
- The horizontal position of the front of the head restraint with respect to the back of the head, or its back-set (H).

Table 2.4 Geometric criteria used to rate head restraint position by the NCAP (Estep and Lund 1995).

Rating	Height, V (mm)	Back-set, H (mm)
Good	< 60	< 70
Acceptable	70 ± 10	80 ± 10
Marginal	90 ± 10	100 ± 10
Poor	> 100	> 110

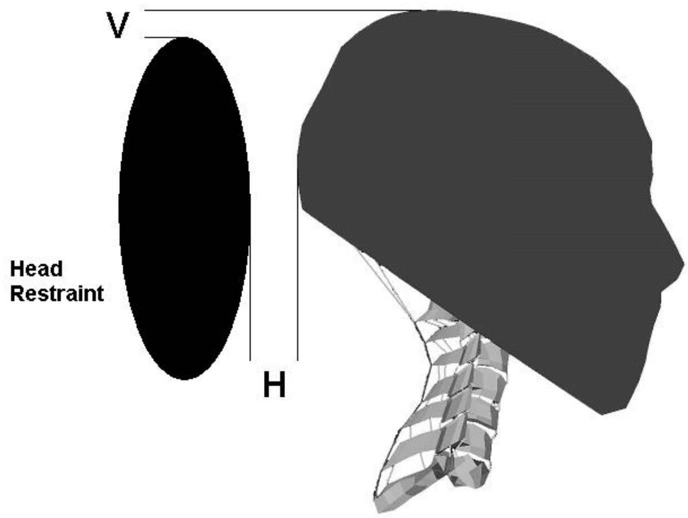


Figure 2.3 The head restraint rating dimensions V and H.

The NCAP criteria were used in the evaluation of 164 vehicles from the 1995 model year (Estep and Lund 1995). Under these criteria, only five vehicles were rated as good, eight were acceptable and the remaining 117 were regarded as poor. In 1995 only 3% of measured head restraints rated good compared with 45% in 2003 (IIHS 2004). The number of poor restraints has decreased dramatically from 82% in 1995 to 10% in 2003. Nevertheless, these criteria are purely geometric and have not been correlated with injury claims. In a study by Bostrom et al. (1997), the researchers concluded that this rating system for vehicles based on seat-system geometry evaluation did not correlate well with available accident data.

2.5.2 Dummy Neck Response Corridors for Extension

When the head is allowed to move with respect to the trunk during a rear impact, the following occurs:

1. The struck vehicle is accelerated forward and the occupants are subjected to a forward push by the seat back;
2. The inertia of the head causes its motion to lag behind the torso forcing the neck into an S shape, with extension at the base (at the level of C7/T1) and flexion at the top (at the level of the occipital condyle).

Mertz and Patrick (1967) demonstrated that a volunteer in a high backed, rigid seat could withstand 30 km/h rear impacts on a test sled without injury. In these tests, the head of the subject remained in contact with the seat back and hence remained stationary with respect to the trunk. The researchers combined the volunteer testing with cadaver tests and derived a dynamic neck response in extension requirement (Figure 2.4).

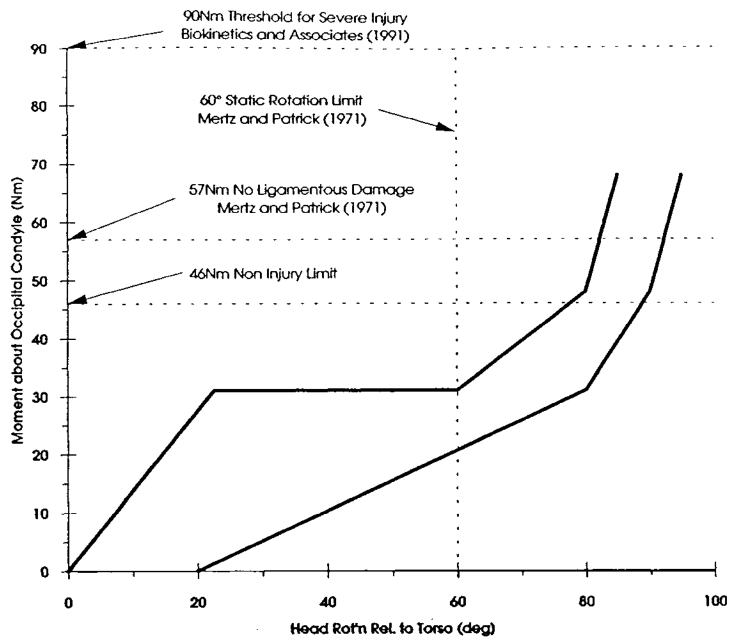


Figure 2.4 The dynamic head/neck moment at the OC in extension, after Mertz and Patrick (1971).

More recent volunteer tests have been aimed at improving the understanding of whiplash injury causation. These tests have been carried out on a variety of subjects, in a variety of seat types and with a variety of pulse shapes and have occasionally been to the threshold of mild cervical strain injury. Szabo (1997) reviewed 12 of these test series, with a total of 298 volunteer exposures in staged vehicle-to-vehicle impacts. Included in his review were some of the studies cited here (Table 2.4). The review was made to determine the level of exposure that has been tolerated by volunteers in rear impacts. The change in vehicle velocity is often used to rate impact severity in low-speed rear impacts, but it makes no allowance for the duration of the impact or the rate of change of the acceleration.

The majority of subjects had no symptoms at all and were seated with a normal posture in seats with head restraints and in three-point restraints. Subjects with symptoms, which persisted for longer than 15 minutes, typically suffered neck stiffness for approximately one day after the test. The results are summarised in Table 2.4 below. The change in velocity at which injury becomes significant is at 5-6 km/h.

Table 2.4 The summary of human subject vehicle-to-vehicle rear-impact tests from Szabo (1997).

Change in velocity (km/h)	**No. of tests**	**Symptoms for longer than 15 min**	
		No.	**%**
0-1	7	0	0
1-2	40	0	0
2-3	60	1	1.6
3-4	56	1	1.8
4-5	57	7	12.3
5-6	47	11	23.4
6-7	18	4	22.2
7-8	3	0	0
8-9	5	1	20.0
9-10	3	1	33.3
10-11	2	0	0
Total	298	26	8.7

To assist the systematic development of a dummy for use in rear impacts, Thunnissen et al. (1996) reviewed the available kinematic response data for the dummy neck and defined performance requirements. Three sets of response data for the neck were found to be the best available from current data: the moment about the occipital condyles as a function of the head angle, after Mertz and Patrick (1971), Figure 2.4, and two head-rotation time history corridors at different accelerations, based on the relationship between maximum head rotation and average acceleration, after Ono and Kanno (1993). These response requirements are designed for relaxed car occupants who are unaware of the impending impact. The corridors selected included the effect of T1 motion during the test, as they were designed to produce a neck to retrofit on Hybrid III dummies for use in rear impact testing. The Hybrid III dummy was developed as a frontal crash test dummy by General Motors in the 1970s and has since formed the basis for most vehicle safety system testing (Mertz 2002). It has a rigid thoracic spine, which reduces the neck motion caused by the interaction of the hips and thorax with the seat during rear impacts. These response requirements were used to finalise the design of the TRID neck, which was developed from the prototype RID-neck developed by Svensson (1993).

A comprehensive set of response requirements for dummy necks in low-speed rear impacts was stipulated by van den Kroonenberg et al. (1998), to evaluate biofidelity. These were based on 43 volunteer tests conducted at the Allianz Center for Technics, Germany.

Davidsson (2000) compared the performance of the BioRID P3 rear-impact dummy prototype in rear impacts to the ten volunteer tests performed at the Japan Automobile Research Institute by Ono et al. (1999). The BioRID dummy was designed to be used as a rear-impact dummy for testing seats and head restraints. It has a flexible spine with lordosis and responds to the pressure from a seat in a rear impact in a biofidelic manner. This dummy is now being manufactured by Denton in the U.S.

2.5.3 A Dynamic Test Method

WAD countermeasures based on the static measurement of head restraints have been shown to be only partially effective (Farmer et al. 2003). This has resulted in a proposal for a standard dynamic test for car seats concerning protection in low-speed rear impacts by the International Insurance Whiplash Prevention Group (IIWPG). This proposal consists of a 16 km/h rear-impact sled test of the seat with a BioRID IIe dummy as the seated occupant and is currently available for comment (IIWPG 2003).

The head restraint rating scheme currently in use by the Insurance Institute for Highway Safety is based on a combination of the static assessment of the head restraint position and a dynamic test, with the following characteristics:

- Uses the 50^{th}-percentile BioRID dummy with specific positioning instructions
- Impact sled test with a peak acceleration of 10 g (5 g mean), and duration of 91 ms;
- Criteria used to assess the seats are:
  - Two seat design parameter measured during the test, time to head restraint contact (must be less than 70 ms) and torso acceleration (must be less than 9.5 g);

- Two evaluation criteria measured on BioRID during the test, the maximum neck shear force and maximum neck tension.

Using this new dynamic test procedure and dummy, the Insurance Institute for Highway Safety rated 73 seat/head restraint combinations available in 63 car models sold in rear-end crashes at low to moderate speeds (IIHS 2004). Seats with head restraints, that had good or acceptable geometry as described above, were tested dynamically. Only 8 of the 73 seat/head restraints that were dynamically tested earned overall ratings of good. Sixteen were acceptable, and 19 rated marginal. The other 30 seat/head restraint combinations that were tested are rated poor, as are 24 seats that weren't tested because of inadequate geometry.

2.6 Summary

Field accident studies have shown that the typical sufferers of chronic neck pain resulting from a motor vehicle accident are female, mid thirties in age and in a rear impact while stationary. Whiplash injury occurs not only in rear impacts, but results from all directions of impact in motor vehicles.

The field accident studies also showed that when the seatback was deformed by the rear impact there was less likelihood of the occupant sustaining whiplash-related injuries and head restraints have proven only partially effective. The use of seat belts seems to increase the likelihood of injury.

Some investigators have linked an increased likelihood of chronic pain to a lack of awareness of the impending impact or to having the head out of position.

The vehicle factors implicated by the field studies have included the stiffness of the vehicle rear structure, the dynamic stiffness of the seat back and head restraint and the design of the seat belt. It has also been shown that a well-designed seat and head restraint is able to overcome the effects of the vehicle structure and the seat belt.

The prevention of injury requires the injury mechanism to be understood, the lack of understanding in this area has been a handicap to the implementation of engineering solutions. The attempts at regulation of the dimensions and static stiffness of the head restraints fitted to vehicles have been only partially effective. The introduction of consumer information based on static testing has also been only partially effective. For the optimisation of the design of the safety systems for vehicles in rear impacts a dynamic test procedure is being finalised and a specialised biofidelic rear impact dummy, the BioRID, is now available. An appropriate injury criterion is still lacking for use in such testing. There is a need for a better understanding of injury whiplash mechanisms to allow the development and acceptance of improved injury criteria. The next chapter of this thesis reviews the biomechanics of whiplash-associated injury.

CHAPTER 3 BIOMECHANICS OF WHIPLASH INJURY

3.1 Introduction

The biomechanical factors involved in whiplash injury are collected together in this chapter. The main areas covered and the context of this chapter within the overall structure of the thesis is shown in Figure 3.1. These areas include the basic anatomy and biomechanics of the lower neck, the clinical data regarding lower neck injury and the insights derived from testing. The testing discussed includes *in-vitro* testing, as well as the testing of volunteers and cadavers. Data from volunteer testing are used to define the phases of the whiplash motion in rear impacts. Finally, the various injury hypotheses that have been proposed are discussed.

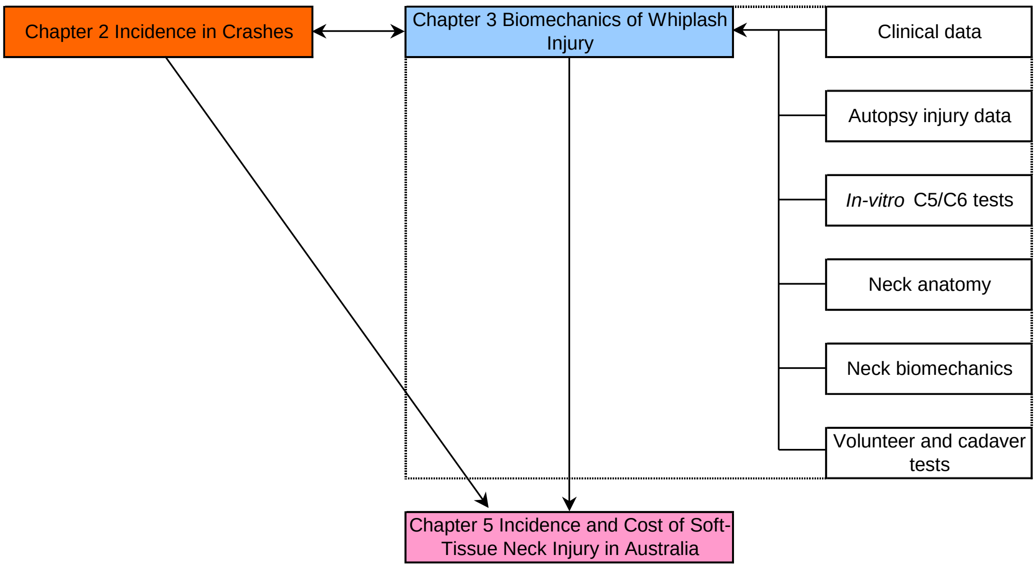


Figure 3.1 The main areas covered and the context of this chapter within the thesis.

3.2 Anatomy of the Lower Neck

3.2.1 Classical Neck Anatomy

The cervical spine is the upper section of the spine, which supports the head and protects the spinal cord. Its articulation allows the head to move relative to the torso. The four basic motions of the head and neck are *flexion* (forward bending), *extension* (rearward bending), *lateral flexion* (sideward bending), and *axial rotation*. The bones making up the neck (superior to inferior) are the base of the skull, the seven cervical vertebrae identified as C1 to C7, as illustrated in Figure 3.2, and the top thoracic vertebra, T1.

The upper cervical spine consists of the *occiput*, which is the base of the skull, the *atlas* (C1), and the *axis* (C2). In flexion and extension, the occiput articulates with the atlas through the occipital condyles (OC), which are convex in shape. The atlas has no vertebral body but consists of a bony ring with anterior and posterior arches on which the articular facets and transverse processes are located. The upper facets are large, concave and elliptical in shape. The axis, like the lower vertebrae, has a body and an arch, but with an additional element known as the odontoid process or dens. The dens protrudes upward from the body of C2 and acts as a pivot about which the head and atlas rotate axially.

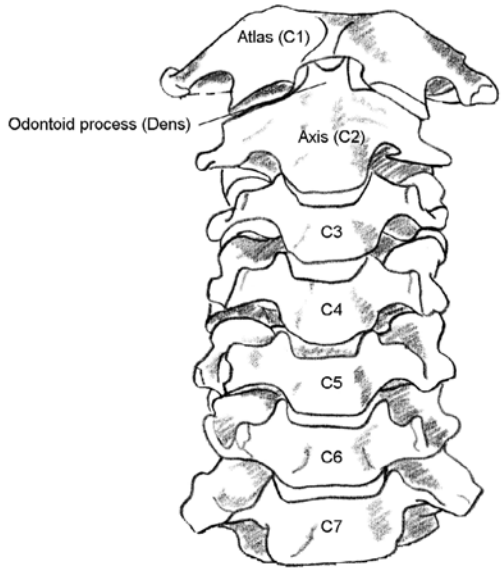


Figure 3.2 Anterior view of the vertebra of the cervical spine, from C1 to C7.

The lower cervical spine consists of vertebrae C3 through C7. Each vertebra consists of a cylindrical body and an arch (Figure 3.3). The lower surface of the vertebral body, the inferior end plate, is concave from front to back, whereas the superior end plate is concave from side to side. The arch includes two pairs of articular facets, a spinous process and two transverse processes. The articular facets are almost flat and covered with articular cartilage, and have a backward inclination of about 45 degrees in the horizontal plane. The transverse and spinous processes are attachment points for muscles and ligaments. The arch and body enclose the vertebral foramen, which form the spinal canal through which the spinal cord and associated structures run.

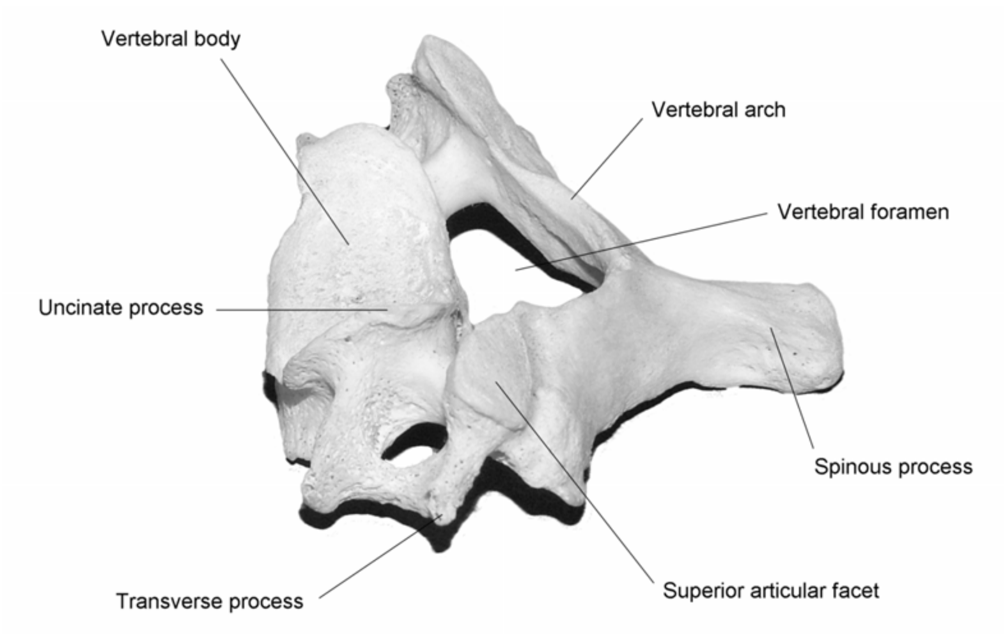


Figure 3.3 View of the C6 vertebra.

The cervical vertebrae are linked by soft tissues, which include intervertebral discs, ligaments, uncovertebral joints, facet joints and muscle. These soft-tissue components all contribute to the relative motion between the vertebrae. The linkage between two adjacent vertebrae is formed from the intervertebral (IV) disc, the facet joints and the uncovertebral joints. The disc allows for motion in all directions, while the uncovertebral and facet joints (zygapophysial joints) guide and constrain this motion.

The five intervertebral discs (from C2/C3 to C7/T1) are the main connecting structures between the cervical vertebral bodies. An intervertebral disc is a fibrocartilaginous pad between the end plates of each two adjacent vertebral bodies. Classically, the healthy disc is described as consisting of concentrically arranged components:

- The outer alternating layer of collagen fibres forming the peripheral rim of the anulus fibrosus (AF);
- A fibrocartilage component forming a major portion of the anulus fibrosus;
- A region where the anulus fibrosus and the nucleus pulposus (NP) merge; and
- The central nucleus pulposus made of a soft, pulpy, highly elastic mucoprotein gel.

The cervical discs are thicker anteriorly, giving the cervical spine a distinct curve in the sagittal plane known as the cervical lordosis, shown in Figure 3.4.

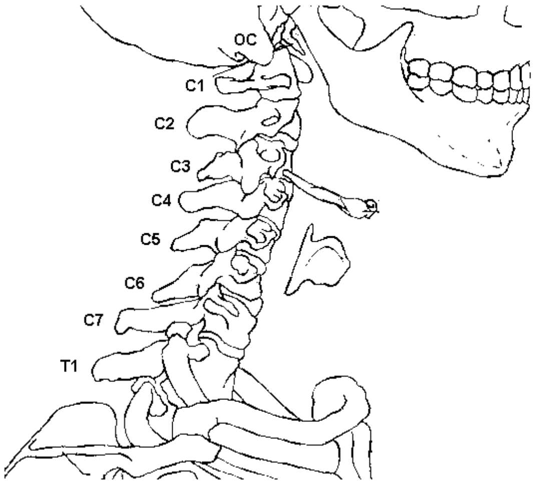


Figure 3.4 Lateral view of the vertebra C1 to T1, showing the lordosis of the cervical spine.

The uncovertebral joints are small joints, formed by the uncinate processes of the lower vertebra and the lower endplate of the upper vertebra, on either side of the intervertebral disc.

The facet capsular joints (FC) are synovial joints formed by the corresponding articular facets of adjacent vertebrae, and are enclosed by capsular ligaments. Usually, synovial joints only permit sliding motions, but within the facet joints other movement is possible due to some slack in the capsular ligaments.

The motion of the cruciate complex of the atlanto-axial joint (surrounding the dens) is controlled by the following major ligaments:

- The transverse ligament is a strong horizontal ligament, which holds the dens against the anterior arch of the atlas, constraining its posterior movement.
- The apical ligament extends from the top of the dens to the occiput.
- The two alar ligaments extend from the dens to the medial aspect of the occipital condyles and the atlas.

In addition to the cruciate complex of the atlanto-axial joint, there are five biomechanically relevant ligaments in the cervical region of the spine (Kleinberger 1993). The positions of these ligaments at the C5/C6 level of the cervical spine are shown in Figure 3.5.

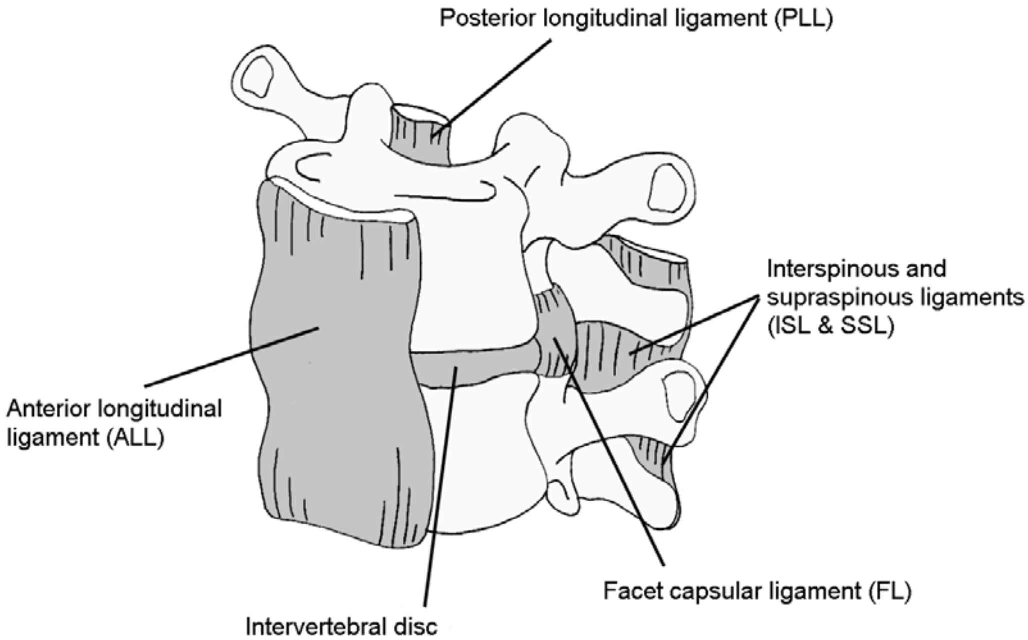


Figure 3.5 Sketch illustrating the C5/C6 cervical spine motion segment and the major ligaments.

- The <u>anterior longitudinal ligament</u> (ALL) extends to the base of the occiput via the anterior occipito-atlantal membrane, and descends along the anterior aspect of the atlas and all of the vertebral bodies. It is firmly attached to the vertebral bodies and is only loosely connected to the intervertebral discs, serving to limit extension and distraction (stretching) motions of the neck.
- The <u>posterior longitudinal ligament</u> (PLL) extends to the base of the skull via the tectorial membrane, and descends along the posterior aspect of the vertebral bodies. It is firmly attached to the intervertebral discs but only loosely to the vertebral bodies, serving to limit flexion motion.
- The <u>ligamenta flava</u> (LF) complex connects the laminae of adjacent vertebra and extends laterally to the articular processes.
- The <u>facet capsular ligaments</u> (FC) are attached around the articular facets, and are part of the synovial joint. They lie perpendicular to the facets and are slack at rest.
- The <u>supraspinous</u> and <u>intraspinous ligaments</u> (SSL and ISL) connect the adjacent spinous processes. While in extension and when the head is upright, these ligaments are slack, tightening during flexion of the neck.

The major muscles involved in controlling neck motion are listed in Table 3.1, with the corresponding points of origin and insertion, and the major functions they control. Any motion of the neck is actuated by a series of these muscles acting in concert. The muscles can be divided into three types:

- *Superficial* muscles, which have no attachments to the vertebrae, but run directly from the thorax to the skull, such as the sternocleidomastoid and the trapezius;
- *Intermediate* muscles, which have several attachments to the cervical vertebrae and run from the vertebrae to the thorax, such as the scalenus muscles; and
- *Deep* muscles, which join one vertebra to another or span several vertebrae, such as the sub-occipital or longus muscles.

Table 3.1 Major muscles of the neck with their attachments and functions, after Stone and Stone (1990).

(a) For Flexion

Muscle	Origin	Insertion	Action
Longus Colli (inferior oblique)	Anterior surfaces of C1-C3	Transverse processes of C5 to C6	Flexes neck
Longus Colli (superior oblique)	Transverse processes of C3-C5	Anterior arch of C1	Flexes neck
Longus Colli (vertical part)	Anterior surfaces of T1-T3 and C5-C7	Anterior surfaces of C2-C4	Flexes neck
Longus Capitus	Transverse processes of C3-C6	Anterior occipital bone to foramen magnum	Flexes neck and head
Scalenus Anterior	Transverse processes of C3-C6	Inner border of first rib	Flexes and rotates neck
Scalenus Medius	Transverse processes of C2-C7	Superior surface of first rib	Flexes and rotates neck
Scalenus Posterior	Transverse processes of C5-C7	Outer surface of second rib	Flexes and rotates neck
Sternocleidomastoid	Manubrium Clavicular head	Mastoid process S-nuchal line (occipital bone)	Bends neck laterally, rotates head Flexes neck, draws head ventrally
Rectis Capitis Anterior	Anterior transverse processes of C1	Anterior occipital bone to foramen magnum	Flexes head

(b) For Extension

Muscle	**Origin**	**Insertion**	**Action**
Trapezius	Occipital process, LM, spinous processes of C7, T1-T12	Clavicle, acromion and scapula	Extends neck, elevates and rotates scapula
Splenius Capitus	Lower LN, spinous processes of C7, T1-T4	Temporal bone	Extends and rotates head
Splenius Cervicis	Spinous processes of T3-T6	Transverse processes of C1-C3	Extends and rotates head
Semispinalis Capitis	Transverse processes of C4-C7 and T1-T7	Occipital bone	Extends and rotates head
Semispinalis Cervicis	Transverse processes of T1-T6	Spinous processes of C2-C5	Extends and rotates vertebral column
Longissimus Capitis	Transverse processes of T1-T5, Articular processes of C5-C7	Posterior temporal bone	Extends and rotates head
Longissimus Cervicis	Transverse processes of T1-T5	Transverse processes of C2-6	Extends and laterally flexes vertebral column
Rectus Capitis Posterior Major	Spinous process of C2	Occipital bone	Extends and rotates head
Multifidus Cervicis	Articular processes C5-T4	Spinous processes of C2-C7	Extends and rotates neck
Rectus Capitis Posterior Minor	Posterior arch of C2	Occipital bone	Extends head
Obliquus Capitis Superior	Transverse process of C1	Occipital bone	Extends and laterally flexes head

(c) For other motions

Muscle	**Origin**	**Insertion**	**Action**
Rectus Capitis Lateralis	Transverse processes of C1	Jugular process of occipital bone	Laterally flexes head
Levator Scapulae	Transverse processes of C1-C4	Scapula	Laterally flexes head
Obliquus Capitis Inferior	Spinous process of C2	Transverse processes of C1	Rotates C1
Hyoideus (lumped)		Hyoid bone	Controls hyoid bone

3.2.2 Functional Anatomy of the Lower Cervical Spine

<u>The rotation axes of the intervertebral joints</u>

In their review of the biomechanics of the cervical spine, Bogduk and Mercer (2000), following the work of Penning (1988), suggested that the cervical intervertebral joints are saddle structures (Figure 3.6). The inferior surface of the upper vertebral body is concave downwards in the sagittal plane and matches the form of the superior surface of the lower vertebral body due to the uncinate processes. This allows rocking of the superior vertebra, sliding in the sagittal plane about Axis 1, and rotation in the transverse plane about Axis 2 (Figure 3.7).

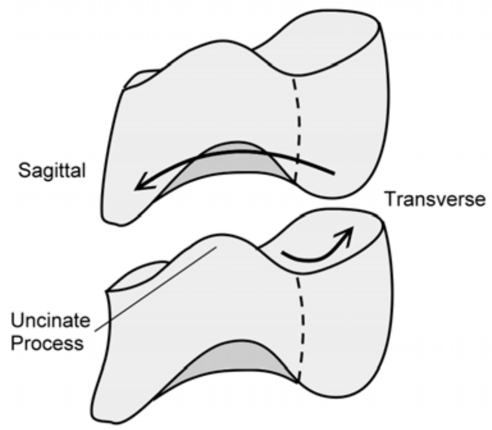


Figure 3.6 The lower cervical intervertebral joints are saddle structures, allowing sliding in the sagittal plane and rotation in the transverse plane, after Bogduk and Mercer (2000).

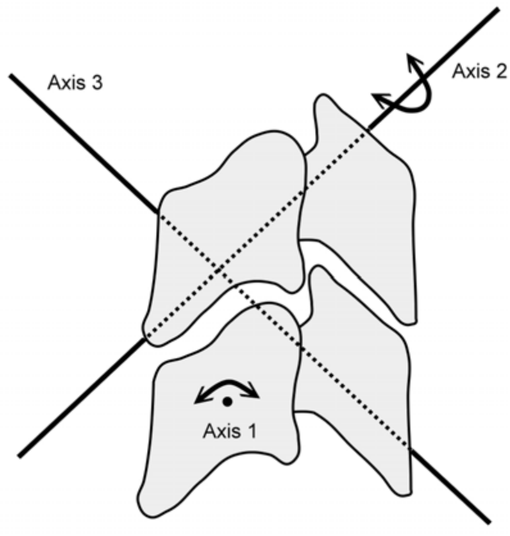


Figure 3.7 A sagittal section of the C5/C6 vertebra showing directions of rotation of the lower cervical intervertebral joints: flexion/extension occurs about Axis 1; axial rotation in the plane of the facet capsule around Axis 2; and, no motion is possible about the remaining orthogonal Axis 3, after Bogduk and Mercer (2000).

The FC permits the sliding and rocking motion of the intervertebral joint in the sagittal plane, but constrains most other directions of motion. The vertebral body is also able to rotate about Axis 2, which is perpendicular to the facet plane (Figure 3.8). The vertebral body is unable to rotate around the third orthogonal axis, Axis 3, due to contact between the faces of the facets (Figure 3.9). Rotation in this plane may occur only if the facet face rises up the 45° slope of the opposing face. For this reason, the only pure rotation of the cervical vertebral joint is in flexion/extension, as axial rotation of the neck must be coupled with lateral flexion and vice versa.

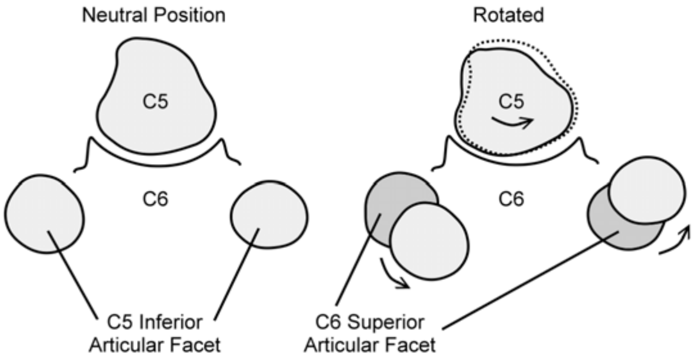


Figure 3.8 The C5/C6 intervertebral joint, from above, sectioned by a plane through the FC and parallel to the facet surfaces. The C5 inferior facet surface is free to glide across the superior surfaces of C6 by rotating around Axis 2, after Bogduk and Mercer (2000).

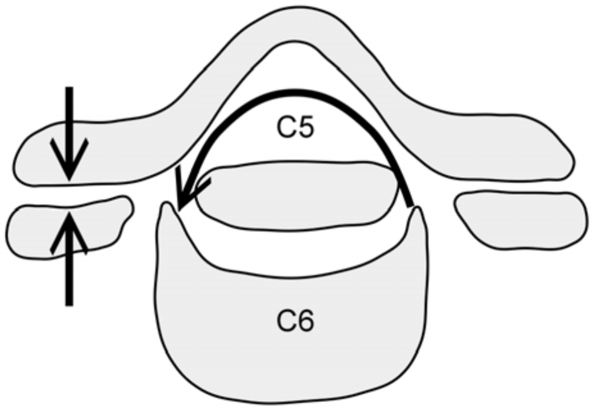


Figure 3.9 A section of the C5/C6 intervertebral joint viewed from above. The section is through the uncinate region perpendicular to the facet surface. C5 is unable to rotate about Axis 3 (in Figure 3.6) due to the inferior facet faces contacting, after Bogduk and Mercer (2000).

Intervertebral Disc Structure

Most anatomy texts indicate that the cervical disc is similar in structure to that in the lumbar spine (White & Panjabi 1990). Mercer and Bogduk (1999) give a detailed three-dimensional description of the cervical intervertebral disc and its surrounding ligaments, based on the study of whole cervical spinal columns from 12 embalmed, human adult cadavers ranging in age from 39 to 83. The authors found that the cervical AF does not consist of concentric laminae of collagen fibres as in the lumbar disc. Instead it forms a crescent shaped mass of collagen: thick anteriorly and tapering laterally to the uncinate processes. The AF is missing posterio-laterally and the posterior AF is limited to a thin layer of paramedian fibres with longitudinal and alar disposition (Figure 3.10). The ALL covers the front of the disc, and the PLL reinforces the rear. The fibres of the anterior AF converge upwards towards the line of Axis 2; approximately 45° to the plane of the intervertebral joint (Figure 3.11). When viewed laterally (Figure 3.12), the fibres in the anterior AF converge forward and upward towards the line of Axis 2 in Figure 3.7. The details of the neck disc anatomy are supported by other researchers (Taylor, Twomey & Levander 2000).

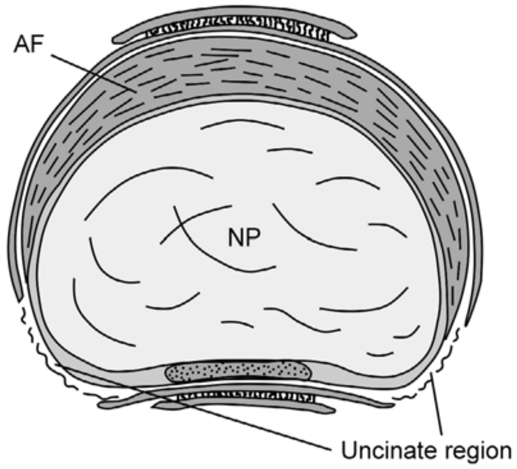


Figure 3.10 Top view of a cervical disc showing the anterior anulus fibrosus (AF) as a crescent and surrounding the fibrocartilaginous core of the nucleus pulposus (NP), after Mercer and Bogduk (1999). The lack of anulus fibrosus (AF) at the uncinate region of the disc is shown.

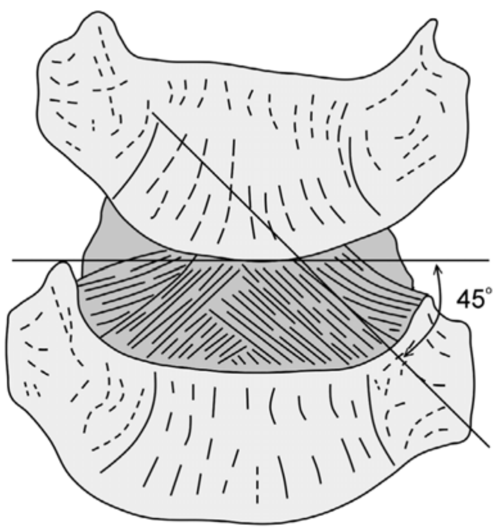


Figure 3.11 Front view of a cervical disc showing the fibres of the anterior anulus fibrosus (AF converging upwards to the midline at an angle of approximately 45°, after Mercer and Bogduk (1999).

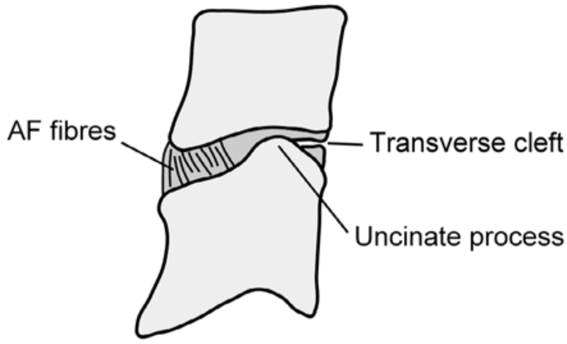


Figure 3.12 Lateral view of a cervical disc with its superior and inferior vertebral bodies showing the fibres of the anterior AF converging upwards to Axis 2 with a transverse cleft of a degenerated NP, after Bogduk and Mercer (2000).

Figure 3.12 also illustrates a transverse cleft in a degenerated NP. The degeneration of the disc is significant in terms of the load sharing and motion of the disc. The clefts originate in the region of the disc covered by the uncinate processes. These clefts are small in a young adult but can become fully developed in the lower cervical spine and replace the NP by the age of fifty (Penning 1988; Taylor, Twomey & Levander 2000). In the testing of the load-displacement behaviour of lower cervical motion segments, Moroney et al. (1988) categorised the amount of degeneration in the discs into 4 grades. The researchers found that there was a 50% increase in mean compression stiffness as the disc degenerated from Grade 1 to 2, and an approximate 50% drop in shear stiffness (Table 3.2). These degenerative changes in the disc occur at around 40 years, the age at which WAD typically occurs (Morris & Thomas 1996).

Table 3.2 Mean disc stiffness in N/mm, with (SD) and (number tested), by degeneration grade measured, by Moroney et al. (1988).

Grade of Degeneration	Compression	Anterior Shear	Right Lateral Shear	Posterior Shear
1	492 (472) (28)	62 (63) (27)	73 (63) (25)	50 (36) (26)
2	737 (885) (4)	31 (13) (4)	31 (13) (25)	18 (7) (3)
3	603 (473) (11)	39 (24)	72 (73) (11)	40 (24) (11)
4	320 (249) (10)	99 (96) (9)	82 (86) (7)	72 (46) (9)
All	328 (236) (3)	76 (18) (3)	114 (19) (3)	53 (26) (3)

3.2.3 Size, Strength and Geometry of the Intervertebral Joint

Yoganandan, Kumaresan and Pintar (2000) assessed the geometry and mechanical properties of cervical ligaments from the C2 to T1 levels. The measurements were made in two procedures.

Firstly, the lengths and cross-sectional areas at mid height of the major neck ligaments (ALL, PLL, FC, LF, and ISL) were measured for 8 human cadavers using micro cryotome images. The data were grouped into middle C2/C5 (n = 4) and lower C5/T1 (n = 4) cervical levels. The ALL and PLL were defined from the mid-height of the inferior vertebral body to the mid-height of the superior body. The LF and ISL were defined from the superior points of attachment to the corresponding inferior points of attachment. The FC were defined from the superior tip of the cephalad facet to the inferior tip of the caudal facet articulation. The geometry was defined based on spinal anatomy and its potential use in complex mathematical models. The geometric lengths and areas of cross section are given in Table 3.3.

Table 3.3 The area of cross section (n = 4) and length (n = 4) of the cervical spine ligaments from micro cryotome measurements (Yoganandan, Kumaresan & Pintar 2000).

	C2/C5		C5/T1	
Ligament	**Area (mm)² (SD)**	**Length (mm) (SD)**	**Area (mm)² (SD)**	**Length (mm) (SD)**
ALL	11.1 (1.9)	18.8 (1.0)	12.1 (2.7)	18.3 (0.5)
PLL	11.3 (2.0)	19.0 (1.0)	14.7 (6.8)	17.9 (0.5)
FC	42.4 (6.4)	6.9 (0.7)	49.5 (12.3)	6.72 (0.5)
LF	46.0 (5.8)	8.5 (0.9)	48.9 (7.9)	10.6 (0.6)
ISL	13.0 (3.3)	10.4 (0.8)	13.4 (1.0)	9.87 (0.7)

The biomechanical properties of the cervical ligaments were measured for a separate group of 25 cadavers (mean age 68 years) using in situ axial tensile tests. The specimens were prepared by excising the individual motion segments to remove the other ligament structures. The biomechanical properties including the stiffness, stress, strain and energy, are presented for each of the five ligaments in Table 3.4 and the Young's Modulus[3] in effect in the response curves, Table 3.5.

Table 3.4 The biomechanical properties of the cervical spine ligaments including the stiffness, energy, and stress and strain at failure are presented for each of the five ligaments (Yoganandan, Kumaresan and Pintar 2000).

Ligament	Number tested	Stiffness (N/mm) (SD)	Energy (NM) (SD)	Stress (MPa) (SD)	Strain (%) (SD)
			C2/C5		
ALL	10	16.0 (2.7)	0.6 (0.3)	8.4 (1.8)	30.8 (5.0)
PLL	7	25.4 (7.2)	0.2 (0.1)	6.3 (2.3)	18.2 (3.2)
FC	8	33.6 (5.53)	1.5 (0.5)	5.7 (1.5)	148.0 (28.5)
LF	12	25.0 (7.0)	0.5 (0.2)	2.6 (0.8)	77.0 (12.9)
ISL	8	7.7 (1.6)	0.1 (0.0)	3.0 (0.8)	60.9 (11.2)
			C5/T1		
ALL	7	17.0 (3.4)	0.5 (0.1)	12.0 (1.4)	35.4 (5.9)
PLL	10	23.0 (2.4)	0.4 (0.1)	12.8 (3.4)	34.1 (8.8)
FC	11	36.9 (6.1)	1.5 (0.4)	7.4 (1.3)	116.0 (19.6)
LF	11	21.6 (3.7)	0.9 (0.2)	2.6 (0.3)	88.4 (13.1)
ISL	8	6.4 (0.7)	0.2 (0.1)	2.9 (0.7)	68.1 (13.8)

Table 3.5 The bilinear Young's Modulus for the cervical spine ligaments (ε_{12} is the strain transition from linear model E_1 to model E_2 (Yoganandan, Kumaresan & Pintar 2000).

	C2/C5			C5/T1		
Ligament	**E_1**	**E_2**	**ε_{12}**	**E_1**	**E_2**	**ε_{12}**
ALL	43.8	26.3	12.9	28.2	28.4	14.8
PLL	40.9	22.2	11.1	23.0	24.6	11.2
FC	5.0	3.3	56.8	4.8	3.4	57.0
LF	3.1	2.1	40.7	3.5	3.4	35.3
ISL	4.9	3.1	26.1	5.0	3.3	27.0

These data, when compared to previous measurements, have several advantages:

[3] The Young's Modulus, or Modulus of Elasticity, of a material is the slope of the stress versus strain curve.

- The complex insertion patterns are retained;
- There is no need to remove the ligaments from their insertion point;
- The true resting length of the ligament is retained;
- The internal load balance, which keeps the spine stable, is retained. This can be seen in the lack of toe[4] effect in the response curves.

Overall, the FC and LF exhibited the highest cross-sectional areas ($p < 0.005$), while the longitudinal ligaments had the highest length measurements. Although not reaching statistical significance, cross-sectional areas were higher in the C5/T1 group than in the C2/C5 group and lengths were higher in the C2/C5 than in the C5/T1 group with the exception of the flavum. Biomechanical failure strains were higher for the ligaments of the posterior (ISL, FC and LF) than the anterior complex (ALL and PLL). In contrast, the failure stress and Young's Modulus were higher for the anterior complex ligaments than the posterior complex ligaments. Longitudinal ligaments exhibited higher magnitudes of the modulus of elasticity than the ligaments of the posterior complex. When comparing the upper and lower cervical groups, however, there were no similar tendencies in the structural responses – stiffness and energy.

Pintar et al. (1986) tested both intervertebral discs and intact motion segments in tension to derive the biomechanical parameters summarised in Table 3.6.

Table 3.6 The strength of discs and intact segments in tension (Pintar et al. 1986).

Level	Number tested	Mean force at failure (N)	Mean deformation at failure (mm)	Mean energy at failure (J)	Mean stiffness (N/mm)
Disc					
C4/C5	3	570	9.3	5.5	66.8
C5/C6	1	391	12.7	2.6	22.0
C6/C7	2	505	9.9	3.5	69.0
Intact Segment					
C4/C5	2	834	6.5	3.0	219.4

Nowitzke, Westaway and Bogduk (1994) examined the relationship between the height and slope of the cervical facet joint (FC) and the patterns of motion of the cervical

[4] Typically, ligaments are thought to have an initial region of low stiffness around the neutral range of the joint. Yoganandan et al. (2000) suggest that this is due to the retraction of the ligament fibres when the ligament is detached from its insertion point.

vertebrae. The geometries of the C3 to C7 superior articular processes were measured by means of lateral radiographs of 40 normal subjects (Figure 3.13). The results of these measurements are given in Table 3.7. It was found that while the slope of the facet had little or no relationship with the position of the instantaneous centre of rotation (ICR), the height of the superior articular process from the base of the vertebra was in a clear linear relationship with the ICR height.

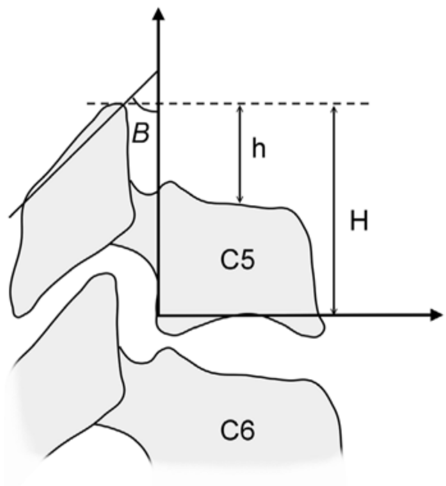


Figure 3.13 Diagram showing the measurements made by Nowitzke, Westaway and Bogduk (1994) of the heights, h and H, and angle *B* of the superior articular process of normal subjects.

Table 3.7 The heights H and h and angle *B* of the C3 to C7 superior articular processes for (n = 40) normal subjects, Nowitzke, Westaway and Bogduk (1994).

Vertebra	*B* (°) (SD)	H (mm) (SD)	h (mm) (SD)
C3	36.4 (7.8)	19.8 (3.4)	6.2 (3.2)
C4	40.3 (6.1)	21.8 (2.9)	8.7 (2.6)
C5	41.7 (8.5)	22.9 (3.0)	10.2 (2.2)
C6	39.7 (8.4)	24.6 (3.2)	11.7 (3.0)
C7	31.4 (7.5)	28.3 (3.5)	13.6 (2.8)

3.3 Motion of the Normal Cervical Spine

3.3.1 Clinical Assessment of Instantaneous Axes of Rotation

In moving from full extension to full flexion, the cervical vertebra each scribes an arc whose centre lies somewhere below the moving vertebra. This centre is called the instantaneous centre of rotation (ICR) and its location can be obtained by superimposing radiographs of the spine taken at the beginning and end of movement. The static ICR, as used clinically, is obtained by drawing the perpendicular bisectors of intervals connecting known points on the two positions of the moving vertebra (Figure 3.14).

Several researchers have shown that an ICR can be reliably and consistently found (van Mameren et al. 1990; Amevo, Worth and Bogduk 1991). Amevo, Worth and Bogduk (1991) derived the positions of the ICR for 40 normal subjects, shown in Figure 3.15.

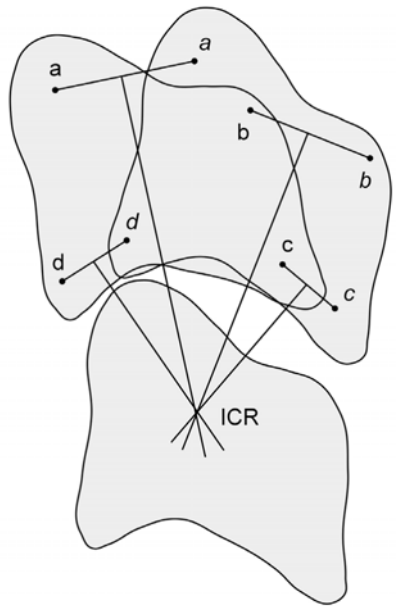


Figure 3.14 The method used for obtaining the instantaneous centre of rotation of a cervical motion segment by drawing the perpendicular bisectors of intervals connecting known points on two positions of the moving vertebra, after Amevo, Worth and Bogduk (1991).

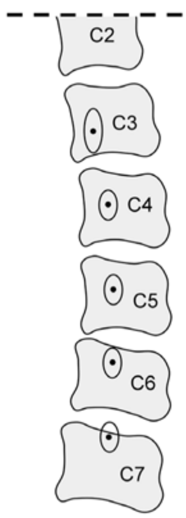


Figure 3.15 The mean positions (dot) and SD (ellipse) of the ICR for 40 normal subjects, derived by Amevo, Worth and Bogduk (1991).

Clinically the ICR is often used to decide whether the neck of a patient has normal or restricted motion. The concept of the ICR is also useful as a means of assessing the capabilities and balance of the ligament constraints in a mathematical model of the neck, and was used for this purpose by de Jager (2000).

3.3.2 Range of Motion of the Cervical Spine

Various researchers have measured the full range of motion of the cervical spine and the effects of combined motions. Schneider et al. (1975) used photogrammetric techniques to investigate aspects of the combinations of two movements of the neck, for a group of 96 volunteers. This study found that when full rotation was followed by flexion, only half of the normal flexion was possible; when followed by extension only one third of the normal extension was possible. Lateral bending following full rotation did not show any effects on the normal range of motion. Full right (or left) rotation, however, was accompanied by slight right (or left) lateral bending.

Ordway et al. (1994) also investigated the effects of the initial starting position of the head and combinations of motions, on the range of motion of the cervical spine. Table 3.8 summarizes the effects of these combined motions. The authors suggested that sagittal flexion/extension, lateral bending and axial rotation measurements were reduced by combinations of head and neck translations and rotations, and that these restrictions were critical to the exact mode of injury. In the case of head retraction, the range of motion of the cervical spine is significantly decreased in flexion, and slightly decreased in axial rotation and lateral bending.

Table 3.8 The effect of head initial position on the range of motion of the cervical spine, after Ordway et al. (1994). (Legend: ↓ slight decrease; O no change; and, ⇓ statistically significant decrease.)

Initial Position	Axial Rotation	Lateral Bending	Flexion	Extension
Protruded	↓	↓	O	⇓
Retracted	↓	↓	⇓	O
Right Rotated	↓	↓	⇓	⇓
Left Rotated	↓	↓	⇓	⇓

White and Panjabi (1990) summarised these and other published measurements of the individual joint range of motion in flexion/extension, axial rotation and lateral bending (Table 3.9).

Van Mameren et al. (1990) used high-speed cineradiography to study the motion of individual vertebra during flexion extension motion of the neck in ten normal volunteers. Twenty-five exposures were taken from full flexion to full extension and the reverse. It was found that the maximum range of motion for a cervical segment did not necessarily occur when the position of full extension was compared to the position of full flexion: so the total range of motion is seldom the sum of the range of motion of the individual vertebra. The researchers also found that the segmental range of motion differs when the motion is from flexion to extension or extension to flexion, resulting in errors of 5 to 15 degrees. They also found that the range of motion for an individual was not stable with time, and would vary each time it was measured.

Table 3.9 Representative values for ranges of motion of the cervical spine, after White and Panjabi (1990).

Neck Interspace	Representative Angle (Degrees)		
	Flexion/ Extension	**Lateral Bending**	**Axial Rotation**
Occipital-Atlanto-Axial			
CO-C1	25	5	5
C1-C2	20	5	40
Middle			
C2-C3	10	10	3
C3-C4	15	11	7
C4-C5	20	11	7
Lower			
C5-C6	20	5	7
C6-C7	17	7	6
C7-T1	9	4	2
Total Range of Motion	136	61	77

A further useful finding by van Mameren et al. (1990) was that the order of movement of the motion segments in the neck is complex and could be divided into three phases. Flexion is initiated in the lower cervical spine (C4 to C7), normally with C6/C7 making the first contribution followed by C5/C6 and then C4/C5. This initial motion was followed by motion at C0 to C2 and then C2/C3 and C3/C4. During this phase a reversal of motion at C6/C7 and sometimes C5/C6 could occur. The third phase was motion in C4 to C7 again. During extension, motion is again initiated in the lower cervical spine (C4 to C7), followed by the start of motion at C0 to C2 and C2 to C4, which is then followed by motion in C4 to C7. This pattern remains consistent for each individual.

Mimura et al. (1989) used bi-planar radiography to directly measure the coupling between axial rotation of a motion segment and the other directions of motion (Table 3.10). For the lower cervical motion segments the axial rotation and the resulting lateral flexion were of a similar magnitude.

Table 3.10 Range of motion of the cervical spine in axial rotation and the coupling with the other directions of motion, determined by Mimura et al. (1989) using bi-planar radiography.

Segment	Coupled Motion		
	Axial Rotation Degrees (SD)	Flexion/Extension Degrees (SD)	Lateral Flexion Degrees (SD)
OC-C2	75 (12)	-14 (6)	-2 (6)
C2/C3	7 (6)	0 (3)	-2 (8)
C3/C4	6 (5)	-3 (5)	6 (7)
C4/C5	4 (6)	-2 (4)	6 (7)
C5/C6	5 (4)	2 (3)	4 (8)
C6/C7	6 (3)	3 (3)	3 (7)

3.4 Clinical Studies of WAD

There is much work investigating whiplash injury and its causation at all levels including crash investigation, clinical studies, radiological studies, biomechanical testing and simulation, and post-mortem studies and testing.

In the report of the Quebec Task Force on WAD, Spitzer et al. (1995) presented a major review of the clinical and related works in this area. The Quebec Automobile Insurance Society (SAAQ) supported the work, which reviewed about 10,000 publications on the subject. Only 346 works were found relevant to the state of the art in diagnosis and treatment of whiplash at the time. In addition to outlining the limits of knowledge in the area, the study proposed that whiplash injury and symptoms be termed *whiplash associated disorder*, and suggested a classification scheme, Table 3.11. The table also has the equivalent Abbreviated Injury Scale (AIS) scores for the injuries to the neck (AAAM 1990). The AIS was designed as a means of ranking and comparing injuries by severity. WAD refers mainly to minor soft-tissue injuries to the neck of AIS1 severity.

The Quebec Task Force noted that in animal models of soft-tissue healing the following stages occur:

1. Acute inflammation lasting less than 72 hours,
2. Repair and regeneration lasting from 72 hours to up to six weeks, and

3. Maturation and remodelling that can last for up to a year.

Table 3.11 The clinical classification of whiplash associated disorders proposed by of the Quebec Task Force (Spitzer et al. 1995), and the equivalent AIS neck injury scores (AAAM 1990).

WAD Grade	Clinical Presentation	AIS Level	Presumed Pathology	Clinical Presentation
0	No neck complaints; No physical signs.	0		
I	Complaints of neck pain, stiffness, or tenderness only; No physical signs.	1	Microscopic lesion; Lesion not causing muscle spasm.	Presents to a doctor more than 24 hours after trauma.
II	Neck complaint AND Musculo-skeletal signs[5].	1	Neck sprain and bleeding around soft tissue (articular capsules, ligaments, tendons, and muscles); Muscle spasm secondary to soft-tissue injury.	Usually presents to a doctor in the first 24 hours after trauma; Non-specific radiation to the head, face, occipital region, shoulder, and arm from soft-tissue injuries; Neck pain with limited range of motion due to muscle spasm.
III	Neck complaint AND Neurological sign(s)[6].	1	Injuries to neurological system by mechanical injury or by irritation secondary to bleeding or inflammation.	Presents to a doctor usually within hours after the trauma; Limited range of motion combined with neurological symptoms and signs.
IV	Neck complaint AND Fracture or dislocation.	2		

Note: The gap in the table indicates the limits of the terms of reference of the Quebec Task Force. Symptoms and disorders that can be manifest in all grades include deafness, dizziness, tinnitus, headache, memory loss, dysphagia, and temporo-mandibular joint pain.

Several studies have described the time-dependent development of WAD symptoms in crash victims. The patients developed symptoms soon after the crash – typically within 24 hours (Barnsley, Lord & Bogduk 1998). The symptoms of WAD are dominated by pain in the neck and headache, followed by pain in the shoulder girdle, paresthesia and weakness in the upper limbs. Less common are symptoms of dizziness, visual disturbances and tinnitus.

Radanov, Sturzenegger and Di Stefano (1995) described the time-dependent progression of the symptoms up to the development of chronic symptoms. These researchers found

[5] Musculo-skeletal signs include decreased range of motion and point tenderness.
[6] Neurological signs include decreased or absent deep tendon reflexes, weakness, and sensory deficits.

that approximately 56% of patients with neck pain due to whiplash recover in the first 3 months after the injury. Thirty percent (30%) develop significant, chronic neck pain (lasting more than 6 months) and 14% of patients suffer pain indefinitely (more than 24 months). The time-dependent progression of the symptoms reported by Radanov, Sturzenegger and Di Stefano (1995) follows closely the development and duration of the pain symptoms following injury to animal subjects reported by Spitzer et al. (1995).

Researchers at the Cervical Spine Research Unit (CSRU) at the University of Newcastle have investigated the causes of whiplash-associated chronic neck pain. The CSRU developed a mapping technique to pinpoint some of the sources of neck pain as an aftermath of whiplash (Dwyer, Aprill & Bogduk 1990; Barnsley et al. 1995). These researchers have also developed the only positive diagnosis tool linking a whiplash-associated symptom (such as neck pain) to a specific area within the neck. Based on patients treated at the CSRU, they found that up to 70% of sufferers of chronic neck pain after whiplash experience cervical FC (or more correctly zygapophyseal joint) pain. For these patients, chronic FC pain was most commonly found at the C2/C3 (44%) and C5/C6 (41%) levels of the neck.

In a comprehensive review of whiplash injury causation, Barnsley, Lord and Bogduk (1998) concluded that there is a converging trend in the result from the many field-, clinically- and experimentally-based studies. The authors conclude that the structures most likely to be injured in whiplash are the facet capsules, the intervertebral discs and the upper cervical ligaments. Injuries to other structures may occur but the available evidence appears to suggest that these are less common. Accordingly, Barnsley and his associates identified the injuries most likely to be associated with WAD (Figure 3.16), and included the following:

- <u>Facet Capsule Injury</u> - ligament tears, cartilage damage, contusion of the intra-articular meniscus haemarthrosis (joint haemorrhage) and possibly extending to microfractures;
- <u>Disc Injury</u> - AF ligament tears, NP cracks and protrusions, and vertebral end plate avulsions;
- <u>Major Neck Ligament Injury</u> - tears to the ALL.

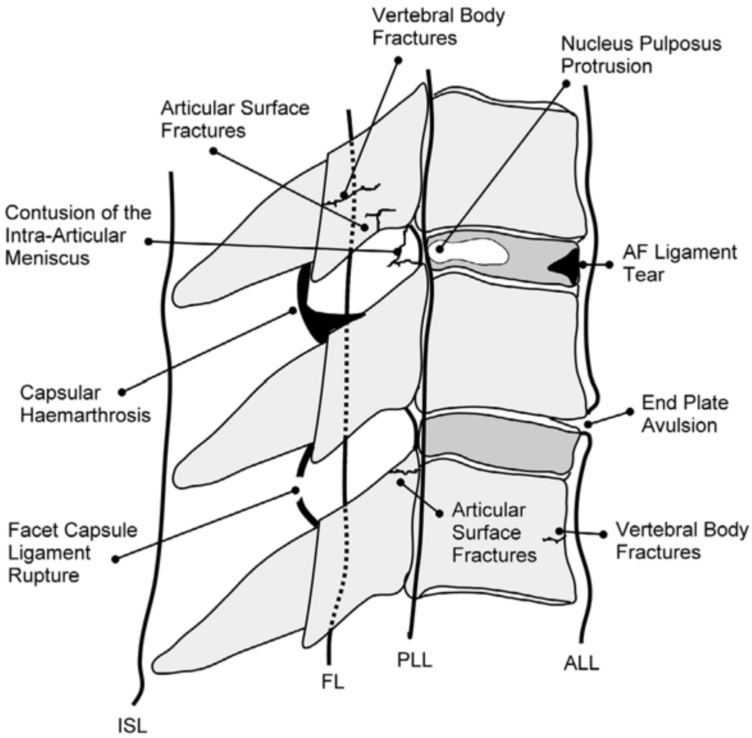


Figure 3.16 A lateral view of a section of the lower cervical spine showing the major neck components including vertebral bodies, neck ligaments, discs, FC and possible WAD injuries, after Barnsley, Lord and Bogduk (1998).

Bogduk (1998) reported on controlled studies on separate populations, which have determined the prevalence of chronic cervical zygapophysial joint pain following whiplash. Collectively, these studies indicated a prevalence of cervical zygapophysial joint pain of 49% with a 40–58% CI. For 53% of the patients with headache as the dominant symptom, pain was traced to the C2/C3 FC joint. The author summarised the known sources of chronic neck pain as follows:

- The source of chronic neck pain was most commonly found in the FC;
- The most commonly affected joints are C2/C3 and C5/C6 followed by C6/C7;
- Most commonly one or both joints at a segment level are affected, followed by two joints at separate levels, usually C2/C3 and C5/C6 or C6/C7, or C5/C6 and C6/C7;
- Headaches arise most commonly from C2/C3 and less frequently from C3/C4.

3.5 Pain Receptors in the Intervertebral Joint

One of the major difficulties in diagnosing, treating or preventing WAD has been the lack of easily discernable injuries. The discussion has been forced to revolve around interpreting symptoms, which may be merely psychosomatic in nature. However, the convergence in the results of the many field, clinical and experimental studies, as suggested by Barnsley, Lord and Bogduk (1998), supports the connection between pain duration and injury. These authors point out that many patients with injuries to the neck ligaments, discs and facet joints associated with whiplash may be expected to have prolonged pain as an outcome, with little chance of healing or spontaneous recovery. This is consistent with the outcome of injury studied in animal models as reported by the report of the Quebec Task Force on WAD, Spitzer et al. (1995).

It is necessary to point out that the purpose of the limited discussion on pain in the intervertebral joint included here is to map out and support some of the issues raised in the latter parts of the study, and not to give a full discussion of the subject. Such a discussion is beyond the scope of this thesis.

It is often worthwhile to investigate the common meaning of a word when trying to understand a difficult concept. Pain has been defined as an unpleasant sensation caused by noxious stimulation of the sensory nerve endings, which under normal conditions signals actual or potential tissue damage (Mosby 1990). It is a subjective feeling and the response to the cause varies amongst individuals. In the case of chronic pain, usually defined as that which continues for more than 6 months, the nervous system itself may become sensitised, and this sensation of pain appears to serve no useful purpose to the organism.

Cavanaugh (2000) reviewed the neurophysiology and neuroanatomy of neck pain. The specialized nerve endings for the sensation of pain are called nociceptors and microscopically they appear as free or finely branched nerve endings. Noxious mechanical and thermal stimuli and certain chemicals can activate nociceptive nerve endings, leading to pain. Tissue damage and inflammation can sensitise nerve endings, causing previously innoxious stimuli to be painful. The pain elicited with only moderate mechanical stress to a sprained ankle or sunburnt skin is an example. The peripheral nerves synapse in the spinal cord. Prolonged nociceptive input to the spinal cord can sensitise the central pain pathway. This is termed central sensitisation. Damage to a nerve itself can result in neuropathic pain. In such cases, areas of the nerve membrane become hyper-excitable.

Nociceptors have been shown to exist in various components of spinal tissues, namely the muscle (Bogduk & Marsland 1988), disc anulus (Bogduk & Windsor 1988) and facet joint ligaments (McLain 1994). Consequently, injury to any of these tissues has the potential to cause neck pain. Schellhas (1996) used both MRI and discography on two groups of patients, one asymptomatic and the other with chronic neck/head pain. It was found that for patients in both groups, normal, healthy discs did not cause pain, while painful discs all exhibited tears to both the inner and outer aspects of the anulus. Facet joint capsules are particularly rich in nerve endings, including nociceptors, and are a primary source of neck pain. It has been demonstrated that pain originating in the cervical facet joints can be referred to areas of the occiput (Dwyer, Aprill & Bogduk 1990). Neck pain can also be radiating, extending into dermatomes of the neck, shoulder or arm.

Typically, disc herniation is part of a long-term process rather than the result of a single traumatic event (White & Panjabi 1990). The nucleus pulposus in herniated discs may irritate the spinal nerve roots, causing radiating pain upon subsequent pressure (Jonsson, Bring & Rauschning 1991). Pressure on irritated nerve roots has been shown to produce prolonged after-discharges in animal models, while mechanical stimulation of nerve roots at the level of pathology has reproduced radiating pain in surgical patients (Cavanaugh 2000). In addition, the dorsal root ganglion (the enlarged areas of the nerve roots that house the cell bodies of peripheral nerves) has unique properties. Compression of even the normal dorsal root ganglion results in sustained after-discharges that can cause radiating pain. In most cases, pain arising from soft-tissue injury resolves after a process of inflammation and repair. The gradual development of some cases of neck pain is nevertheless poorly understood and may reflect disorders in the pain pathway itself, including neuropathic pain and central sensitisation (Cavanaugh 2000).

In the facet joint the adjacent surfaces are covered by cartilage, which is not innervated (Yoganandan et al. 2003). Nerve endings do exist in the underlying subchondral bony processes, and contact between these surfaces may initiate nociceptive firing.

While summarising the clinical assessment of whiplash injury, Bogduk (1998) suggests that standard clinical approaches, such as clinical examination, radiology, CT scans, MRI, functional radiology and psychometrics, are often ineffective in finding sources of pain. This author asserts that as pain is a physiological symptom, a physiological diagnostic test is required. The only two tests being used in the diagnosis of WAD that meet this criterion are:

1. Disc stimulation: Diagnostic disc stimulation is a procedure in which a needle is inserted into an intervertebral disc to stimulate it by distending it with an injection of contrast medium; and,
2. FC (zygapophysial joint) blocks: Blocks allow the monitoring of pain by means of anaesthetising the joint, and more recently its treatment.

3.6 Autopsy Studies of Soft Tissue Neck Injury

The clinical studies of Bogduk and his associates (Barnsley, Lord & Bogduk 1998) are consistent with post-mortem based studies of neck injury in motor vehicle crash victims (Jonsson, Bring & Rauschning 1991; Taylor & Twomey 1993; Taylor & Taylor 1996).

Taylor and Taylor (1996) examined a total of 180 sagittally sectioned spines, including 109 from blunt trauma fatalities and 72 from motor vehicle trauma. Ignoring the more serious injuries, a high incidence of traumatic lesions to the spines such as disc lesions, bleeding into the facet joints and facet fractures, was found. These injuries had been shown in a previous study to be undetected by radiological examination (Taylor & Twomey 1993). The distribution of all the injuries to the cervical discs and facet joints of motor vehicle crash victims in the study is given in Figure 3.17.

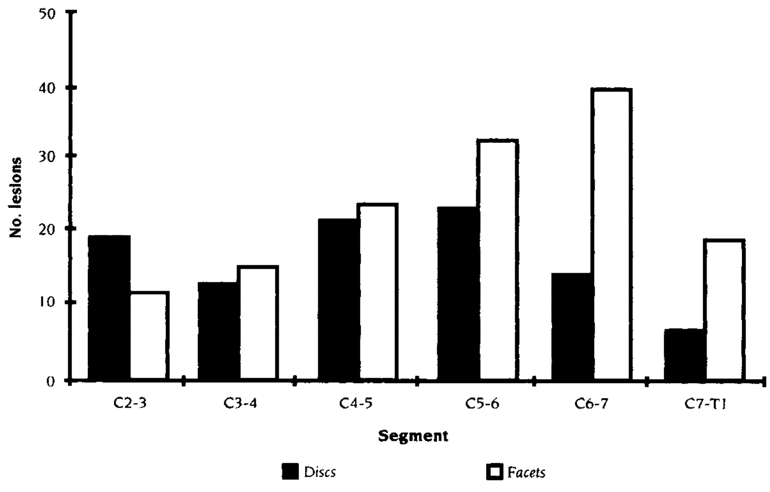


Figure 3.17 The distribution of all neck injuries in motor vehicle crashes (Taylor & Taylor 1996).

The disc injuries found in autopsy were classified as (Taylor & Taylor 1996):

a. *Minor, including:*
   i. tears of the disc from the vertebral rim (rim lesions); or,
   ii. bleeding into the anterior or posterior disc; or
b. *Major*, including:
   i. Complete or partial avulsion of the disc from the vertebra in young subjects; and,
   ii. Traumatic disruption of the disc in older subjects, or traumatic herniation of part of the disc into the spinal canal.

The minor disc lesions were found to be most common at the C3/C4 and C4/C5 levels, while the major lesions were found to have maximum frequencies of occurrence at C5/C6 and C6/C7. In all rim lesions, the longitudinal ligaments remained intact, and in many of the disc avulsions, the longitudinal ligaments were intact, strained or only partially torn. The small anterior muscles remained intact with only intramuscular haemorrhage, even when the anterior longitudinal ligament (ALL) was ruptured. The facet injuries were also classified as: *minor*, including haemarthroses, capsular tears or articular cartilage damage; and, *major*, including facet fractures of the articular processes. Both types of injury occurred mainly in the C5/C6 and C6/C7 levels of the neck, where the facet joint surface has a steeper slope. A comparison of the frequency of disc and facet injuries for the motor vehicle occupants in the study is given in Figure 3.18.

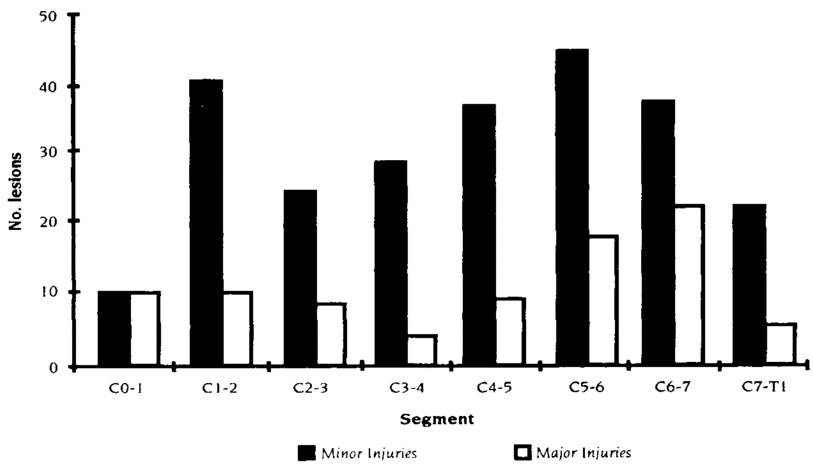


Figure 3.18 A comparison of the severity and frequency of disc and FC injuries for the motor vehicle occupants in the study, after Taylor and Taylor (1996).

In some of the cases, no head injuries had occurred, but an impact to the torso implied that a whiplash-like motion to the cervical spine had been experienced. The Taylors found that in such cases, tears in the discs and bruising in the facet joints were present. Injuries to the disc and FC were also identified in those patients with a history of whiplash injury, who died from other causes. Based on the observations of the neck injuries in autopsy, the authors suggested a hierarchy of injury to the soft tissue of the intervertebral joints. This related the site of the injury to increasing severity of the neck loads and motion. Facet joint injuries, such as ligament tears, cartilage damage, joint haemorrhage and possibly extending to microfractures, were most likely to occur initially, followed by injuries to the disc, such as AF ligament tears, cracks and protrusions in the NP, and end plate avulsions. In more severe cases, injury to the major neck ligaments, such as tears to the ALL or other neck components such as the muscles, spinal cord, ganglia or vertebra, may follow.

3.7 Experimental Studies

3.7.1 Introduction

The earliest experimental studies of WAD were by Severy and Mathewson (1957), who used volunteers to demonstrate the phasing of the body responses in an impact and the seat design requirements for protection of the neck in rear impacts.

Mertz and Patrick (1967) tested a volunteer and several embalmed cadavers using an impact sled. They developed a method for calculating the inertia loading of the neck by the head, using a free body diagram. In a later study, Mertz and Patrick (1971) proposed neck injury criteria, which have been used in many neck injury evaluations. The criteria require that the flexion bending moment at the head/neck junction, or occipital condyle OC, should be less than 190 Nm and that in extension the bending moment should be less than 57 Nm. These criteria were obtained from multiple tests on a group of four cadavers, with no detectable dislocations of the neck vertebrae (i.e. severe ligament damage) by X-ray. Based on this data Newman et al. (1996) constructed the flexion/extension neck injury tolerance curve for an upright, seated person shown in Figure 3.19.

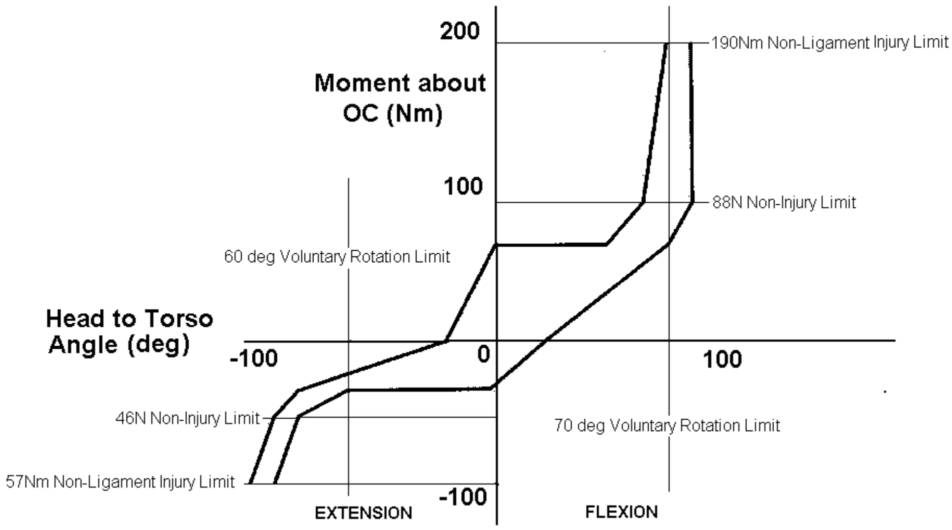


Figure 3.19 Neck injury tolerance and response corridors for the neck flexion/extension of an upright seated person, based on the Mertz and Patrick (1967) corridors from calculated neck reaction moments in cadavers and volunteers, and after Newman et al. (1996).

Clemens and Burow (1972) tested 21 fresh human cadavers in 25 km/h rear impacts, with and without head restraints. The injuries to the neck in these tests only occurred without head restraint. For the subjects without a head restraint, the injuries at autopsy included: injuries to the disc (90%), ALL tears (80%), FC tears (40%) and fractures to the posterior vertebral body or a spinous process (30%). The severity of these injuries is an indication that the mechanism was hyperextension.

In the 1990s, a growing awareness of the increasing numbers of soft-tissue injuries and the lack of effectiveness of available head restraints led to further work in investigating the response of volunteers in rear impacts. Important among these studies were those by Ono and Kanno (1993), McConnell et al. (1993 & 1995), Geigl et al. (1994), Szabo and Welcher (1996), Ono et al. (1997) and Siegmund, Brault and Wheeler (1998). Testing has centred on human volunteers as this supplies the best description, due to posture, muscle tone and muscle reaction, of the kinematics of a live occupant in a rear impact. Consequently, much of the detailed kinematic human responses have since been determined and the biofidelic development of test dummies for dynamic testing of anti-whiplash systems have been implemented (Thunnissen et al. 1996; Davidsson et al. 1999b).

3.7.2 Neck Motion of a Volunteer in a Rear Impact

An important series of volunteer tests were carried out at the Japan Automotive Research Institute JARI to define the motion of the neck in rear impacts (Ono et al. 1997; Davidsson et al. 1999b; Kaneoka et al. 2002).

Kaneoka et al. (2002) tested 10 volunteer subjects seated on a sled, to simulate car rear-impact acceleration (Figure 3.20). An impact speed of 8 km/h was used to study the head-neck-torso kinematics and cervical spine responses. The acceleration pulse generated by the sled in the 8 km/h impact speed is shown in Figure 3.20. A headrest was not used in the experiment. The activity of the sternocleidomastoid muscle and the paravertebral muscles were measured with surface electromyography (EMG). The neck axial and shear forces, and the flexion/extension bending moments at the occipital condyle, were calculated by treating the head as a free body using the method developed by Mertz and Patrick (1967). The results for one volunteer are plotted in Figure 3.21. This study has particular importance because the cervical motion was recorded by cineradiography (90 frames per second X-ray) and analysed to quantify the rotation and translation of individual cervical vertebrae resulting from the impact. This method allowed the motion patterns of cervical vertebrae in the crash motion and in normal motion to be compared. Kaneoka and Ono (1998) divided the motion and head-neck-

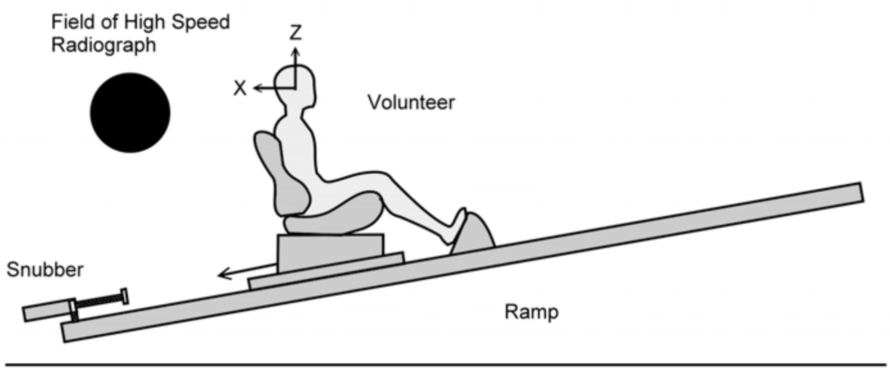


Figure 3.20 Volunteer seated on a sled inclined at 10°, simulating a car rear impact at 8 km/h (Kaneoka et al. 2002).

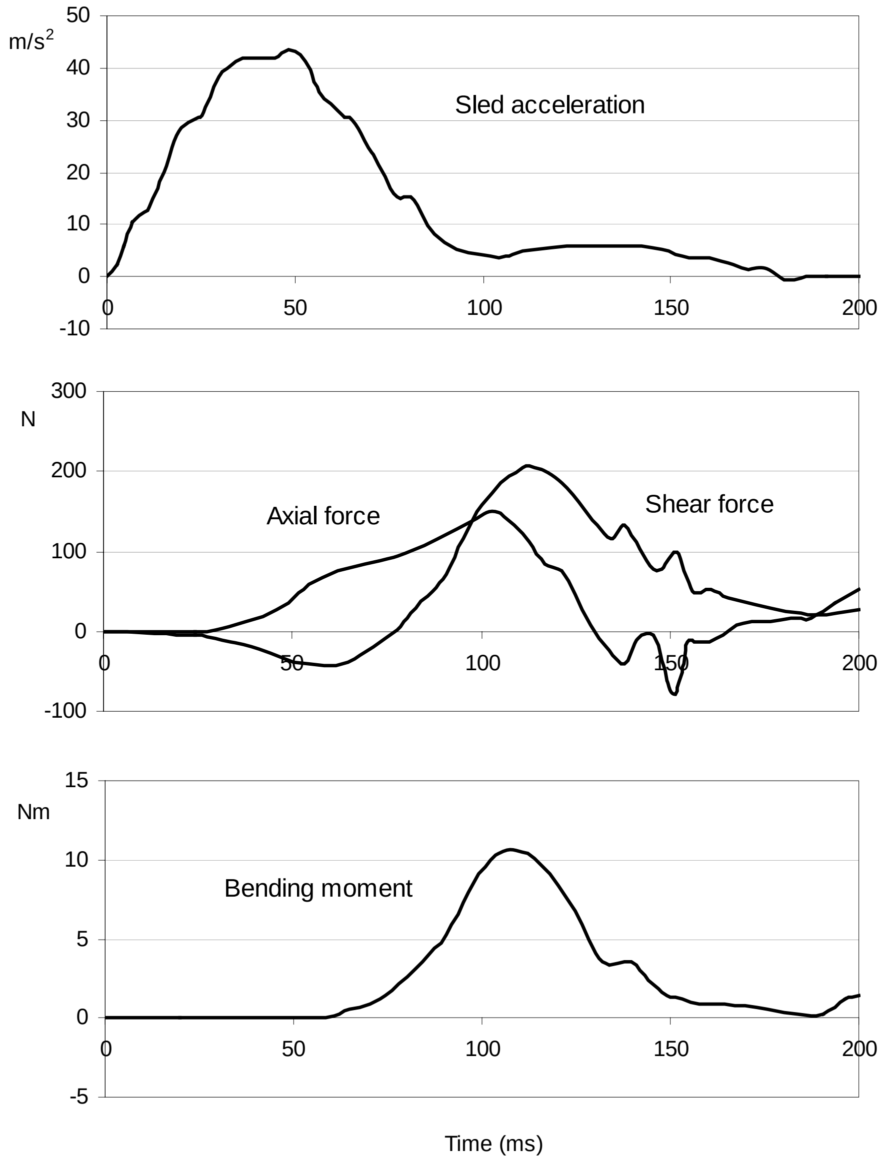


Figure 3.21 The acceleration pulse generated by the sled test of one of the volunteers at the 8 km/h impact speed with the calculated resultant neck axial and shear forces and flexion/extension bending moment at the occipital condyle (Kaneoka et al. 2002).

torso responses of the test subjects into four phases:

Phase 1: Sled motion (0–40 ms)

- The seat begins to press the back of the volunteer;
- The spine begins to straighten;
- Cervical motion has not occurred;
- No muscular response in the neck.

Phase 2: Neck compression (40–100 ms)

- The torso moves forward – pushed by the seat back;
- The torso moves upward – parallel to the seat inclination, causing axial compression of the cervical spine due to the inertia of the head, which reaches a maximum;
- The head remains stationary due to inertia, with a slight initial flexion;
- C6 rotates earlier into extension than the upper vertebral segments (C3, C4 and C5);
- The vertebra of the neck assume an 'S' shape with the upper region in flexion and the lower region in extension;
- No muscular response in the neck.

Phase 3: Neck shear (100–160 ms)

- As the sled slows the torso rebounds and moves forward with some backward rotation;
- The axial force on the neck decreases while the shear force on the neck reaches a peak at about 120 ms;
- The head begins to rotate into extension;
- The cervical spine moves into alignment in extension;
- The EMG of the sternocleidomastoid discharges from about 115 ms.

Phase 4: Full neck extension (150–220 ms)

- The torso moves forward and down;
- The head and neck rotation reaches full extension;
- Shear and axial forces in the neck decrease;
- The muscular discharge finishes by around 220 ms.

The alignment of the vertebra obtained from the high-speed radiography during these phases is shown in Figure 3.22. These phases of the neck motion are supported by the other volunteer tests that have been conducted, and by the whole cadaver and intact human head and neck testing mentioned earlier. The exact timing of the events in a volunteer test is quite variable and depends on the acceleration pulse shape and magnitude, the stiffness of the seat back, the angle of the seat back, the posture and anthropometry of the subject, and whether a head restraint was present.

The S-shaped response in Phase 2 of the neck in a rear impact has been verified by other studies using cadaver head and necks, whole cadavers and volunteers (Yoganandan, Pintar & Kleinberger 1998b; McConnell et al. 1993; Svensson et al. 1993).

If the seat used in the test is fitted with a head restraint, then during Phase 3 the head makes contact and starts to receive additional support. Maximal retraction of the head is most likely to occur before contact with the head restraint (Bostrom et al. 2000). The effectiveness of the head support depends on the geometry and stiffness of the head restraint, its mounting on the seat back and the manner the seat back deflects. A head restraint dynamically located in the correct proximity from the head in terms of offset and height, with appropriate crush stiffness, has the potential to reduce the neck loads in Phases 3 and 4.

In Phase 4, the restrained subject moves forward into the shoulder portion of the seatbelt. The lap portion of a seatbelt also reduces the upward motion of the torso in Phase 2 (Viano 1992a). Field accident data indicates that this rebound from the seat into the restraint system in Phase 4 may be potentially connected with WAD (Krafft et al. 1996).

The ten-degree incline used in this test series (Kaneoka et al. 2002) has some significant effects on the results. The ramp increases the vertical acceleration on the test subject and the compressive load in the neck by 10%. The initial alignment of the cervical vertebra for the test subject presented is actual the first high-speed radiograph and this is at t=44s. At this point the neck of the volunteer is showing no lordosis (Figure 3.22). The volunteer's neck is effectively flexed slightly forward at the time of impact due to the inclination of the ramp and the impact force. This minor change in posture is sufficient to change the motion of the individual vertebra, especially those in the lower neck.

Based on the phases of motion outlined above for the volunteer, there are three distinct periods that have the potential to cause injury to the neck:

- Early in the impact event during the head retraction period and leading to the 'S' shape of the neck (Phase 2);
- Due to the impact with the head restraint, if it is poorly positioned with respect to the head and neck at the time of contact (Phase 3);
- Due to hyperextension for a severe impact with a poorly fitted head restraint or without one (Phase 4); and,
- During the rebound into the seat belt (Phase 4).

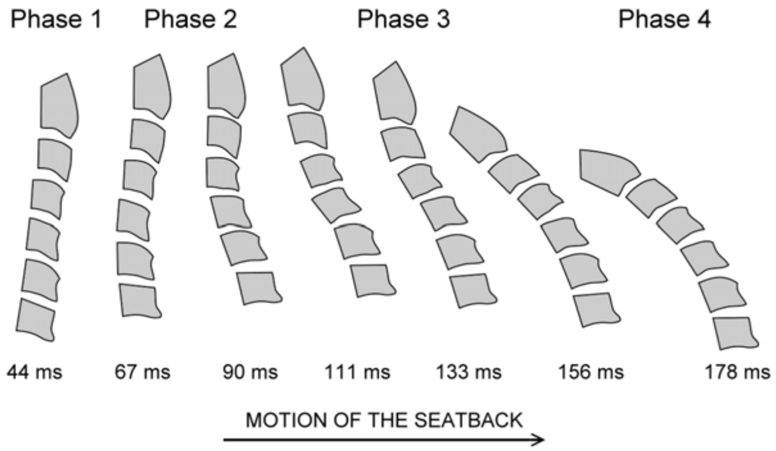


Figure 3.22 The alignment of the C2 to C7 vertebrae of a volunteer during a rear impact obtained by high-speed radiography for the 4 phases described by Kaneoka and Ono (1998). The alignment at 111 ms also includes the FC and spinous processes to illustrate the possibility of impingement of the facet surfaces.

3.7.3 Recent Testing of Cadavers

Volunteer testing must be strictly limited in severity for ethical reasons. As a result, cadaver testing to investigate specific injuries has also continued in more recent times and in various forms. Deng et al. (2000a) and Geigl et al. (1994) used intact cadavers to investigate injury in more severe impacts. The use of whole cadaver tests has been criticised because the effects of active musculature are ignored. It is based on the premise that the neck muscles require some time to activate and that the initial injury occurs early in the event. This over simplifies the actual situation where some muscles must be partially activated in the neck of an upright human to maintain head and neck posture. It has been confirmed by Deng et al. (2000a) in a series of whole cadaver tests with cineradiography, which showed that in a typical rear impact the peak strains to the facet joint ligaments occur before the head contacts the head restraint and before muscle activation could occur.

Two groups began testing human head and neck models in parallel to the volunteer testing described in Section 2 (Yoganandan et al.1998a; Panjabi et al. 1998). Cusick, Pintar and Yoganandan (2001) used the cadaver head and neck model to delineate the ‘S’ curve of the neck and other kinematic factors. The study advanced a compression injury mechanism for the facet surface impingement. Testing of cadaver head and necks has been criticised due to the fixation of the distal end of the neck, which precludes the effect of upward motion and rotation of T1. However, the two types of studies have been confirmed to have similar motion of the lower cervical spine (Yoganandan et al. 2001). This result itself has confirmed that the most important element of the rear impact is the horizontal motion.

Yoganandan et al. (1998b) showed compression of the dorsal region of the facet capsule in whiplash using the experimental human cadaver head and neck. This is consistent with injuries to the facet region found in the detailed neck dissection at autopsy (Taylor & Taylor 1996).

Further definition of possible neck injury in rear impacts was achieved by this group using the cadaver head and neck model (Yoganandan et al. 2001). In this study, 4 intact cadavers were tested with 4.4 m/s (low) and 6.9 m/s (high) velocity changes due to rear impacts to generate minor soft tissue injury to the neck. The injuries and head and neck motion were compared to the detailed motion of the C5/C6 facet joint obtained from 8 cadaver head and neck model tests at velocity changes of 1.3, 1.8, 2.6 and 3.5 m/s. The injuries were obtained from the four intact specimens by cryomicrotomy. The injuries (n = 16) were mainly in the region of the C5/C6 motion segment. The types of injury at C5/C6 were facet capsule ligament tears, anterior longitudinal ligament tears (including one avulsion) and disc disruptions including anulus ligament tears. The other injuries were 2 flaval ligament tears at C6/C7, 1 facet capsular tear at C4/C5 and 2 ligament tears at C1/C2). The motion of the C5/C6 motion segment was found to be complex and non-uniform, which is consistent with the vertebral motion of normal volunteers in flexion extension measured by van Mameren et al. (1990). Sliding motions (x direction shear component) were found between the anterior and posterior regions of the facet and corresponded to the local extension of the neck, but the z axis motion was variable between the anterior and posterior regions of the facet joint. The combination of these motions was found to lead to local stretching of the ligaments, tearing occurring when this stretch was beyond elastic limits.

At the neck motion segment level, several investigators were using *in-vitro* testing of excised motion segments to investigate specific injury mechanisms suggested by other studies. Winkelstein et al. (2000) investigated the strains on the FC ligaments from combined torsion and shear loading to the neck, while Siegmund et al. (2000b) studied the effects of shear and compressive loading to the neck. These two injury mechanisms originated from hypotheses developed from the results of field studies and volunteer tests.

3.8 Role of Muscles in Subject Response

In addition to the work of Kaneoka et al. (2002) described earlier, several other researchers have also investigated the effects of muscular response on the head and neck motion of volunteers in rear-impact tests.

Szabo and Welcher (1996) measured the EMG activity of volunteers during low-speed rear impacts. Ten vehicle impacts were conducted using male and female subjects aged 22-54 years and with a target vehicle velocity change of nominally 10 km/h (from an impact speed of 16 km/h). Accelerometers were affixed to the target vehicle's static centre of gravity and the occupant's head, cervical spine, and lumbar spine. The test protocol was designed to inhibit the subjects from bracing in anticipation of the impacts. The tests were run in a relaxed manner with the subjects not expecting the impact. EMG readings were taken from the superficial neck and back muscles of volunteers, including the superior trapezius, sternocleidomastoid, suboccipital cervical extensors, and the paralumbar muscles.

Typically, initial muscle activity was found to occur 100 to 125 ms after the moment of bumper contact – when the occupant's cervical spine extends during the initial phase of impact. Full muscle tension only developed 60 to 70 ms after the onset of muscle activity – when the cervical spine undergoes flexion. Muscle onset commenced while the neck continued to extend and full muscle tension was not achieved until well into the flexion phase. The cervical flexor, cervical extensor and lumbar paraspinal musculature demonstrated similar points of activity onset. Consequently, the researchers hypothesised a centrally generated response for the initial onset of muscle activity. This response was consistent with being triggered by lumbar spinal acceleration and typically occurred 90 to 120 ms following the onset of lumbar spine acceleration. The authors found difficulty in estimating the active muscle forces.

In a more recent study, Siegmund, Brault and Chimich (2000a) tested 42 male and female subjects (aged 20 to 40 years old) in rear impacts at 2 km/h and 4 km/h. The responses of the sternocleidomastoid and the cervical paraspinal muscles (at the C4 to C6 levels) were investigated using EMG. It was found that at 2 km/h the response time for the sternocleidomastoid muscle was 91(±9) ms while the 4 km/h impact velocity yielded a response of 81(±8) ms. The females in the group had slightly faster onset times for both muscle groups, but neither the magnitude nor time of the peak muscle-lengthening velocity varied with gender. The researchers made the following conclusions:

- The cervical muscles become active in the early phases and are capable of generating forces, which modify the head and neck dynamics later in Phases 3 and 4 of the motion;
- The sternocleidomastoid muscle undergoes lengthening contraction during cervical extension, which is consistent with possible contraction-induced muscle injury;
- The arrangement of the neck muscles provides little resistance to the horizontal shear motion between the head and neck pertaining to whiplash; and,
- The predominantly vertical alignment can lead to axial compression loads as a result of muscle contraction.

In seated low-severity subject-perturbation tests, Kumar, Narayan and Amell (1998) showed that the peak head accelerations of subjects who were aware of an impending horizontal perturbation were approximately half as large as those in subjects who were unaware.

3.9 Role of Gender

The accident studies in Chapter 2 consistently indicated that gender is implicated in whiplash associated pain (States, Balcerak & Williams 1972; Morris & Thomas 1996; Temming 1998). Radanov, Sturzenegger and Di Stefano (1995) found that females were 62% of patients who remained symptomatic after 2 years following the injury. Gibson et al. (2000) found that 55% of the drivers of vehicles in rear impact with chronic pain outcomes were female.

The causes of this susceptibility of females to whiplash injury appear to have multiple factors. Stemper, Pintar and Yoganandan (2004) review possible factors, including genetic, hormonal, structural and injury tolerance. The formation of the cartilage on the facet surface has been recently shown by Yoganandan et al. (2003) to be different between the genders, in females the cartilage covering the facet processes to a lesser extent. The susceptibility of females to soft tissue injury is also shown with other injury types, such as anterior cruciate ligament tears (Ireland 2002).

Siegmund et al. (1997) tested forty-two human subjects (21 male and 21 female) in two rear vehicle-to-vehicle rear impacts with speed change of 4 km/h and 8 km/h. A statistical comparison was made of 31 common peaks in the kinematic responses of the subjects. The position of the head restraint in the testing was with a back-set of less than 100 mm and with the top of the head restraint above the ears. In comparison to the general driving population this was regarded as optimal setting of the head restraint. An associated observational study showed that only 10% of drivers had the back-set of the head restraint set to less than 100 mm and had the top above the ears. Not only were significant effects of pulse severity found for the amplitudes and timing of the peaks but the gender difference was also found to be significant. In particular the trends were for female subjects to have greater and earlier peak horizontal accelerations of the head and C7/T1 joint axis and for males to have greater and later peak head excursion. These findings were consistent with the greater body mass and head size of the male subjects.

Stemper, Yoganandan and Pintar (2003a) tested 10 cadaver head and neck complexes (5 male and 5 female). Each specimen was exposed to velocity changes of 0.58, 1.28, 1.83 and 2.58 m/s with the Frankfort plane horizontal and the occipital condyles vertically above T1. The loading was applied from the rear to T1, which was kept horizontal. The tests were able to produce the ‘S’ shape. It was found that segmental angulation in the lower cervical spine increased with increased test severity. The magnitude of the mean female segmental angulation was greater than for the males at all levels of the neck. Further analysis of the test demonstrated that the female specimens had significantly greater shear motion in the facet regions in the upper regions of the lower neck (Stemper, Yoganandan & Pintar 2004).

The greater amplitude of the female head and neck motion due to head inertia and muscle combined with the probable greater susceptibility of the female to neck injury makes females the at risk group when assessing the risk of whiplash associated neck injury.

3.10 Hypotheses of WAD Injury Mechanisms in the Lower Cervical Spine

There have been many attempts to relate the phenomenon of soft tissue injury to the neck following a rear impact. The direct linkage between the mechanical loading from the crash and the injury leading to the observable symptoms is yet undefined. The clinical data regarding chronic pain outcomes related to WAD strongly supports the hypothesis that over 50% of the injury is located within the FC of the cervical spine. The exact timing and mechanism of this injury to the FC has yet to be determined, let alone other possible injuries. Consequently, neither injury-mechanism nor criteria have been fully established and hence it is useful to review some of the main theories of possible injury in the literature.

3.10.1 Hyperextension of the Neck

Early studies tended to relate WAD injury to hyperextension of the neck. These included primate studies (MacNab 1965), volunteer and cadaver studies (Mertz & Patrick 1967) and field accident studies (States, Balcerak & Williams 1972). The introduction of head restraints as a result of motor-vehicle safety regulation in the 1980s was only partially effective in reducing WAD (see Section 2.2), the introduction of head restraint eliminating some of the injury. The significantly increasing levels of WAD in the last decade combined with the results of the volunteer testing, which suggests possible injury in the early phase of motion, are indications that simple hyperextension of the neck is not the problem.

In a related area, a study by Margulies et al. (1992) investigated how the motion of the neck affected the spinal cord. The researchers used MRI techniques to investigate the motion of the vertebra and the strain in the cord during quasistatic flexion and extension motion. The study indicated that the relative motion of the vertebra could cause deformation in the cord due to occlusion of the foramen and by changes in length of the canal. This mechanism has direct relevance to whiplash-associated injury.

3.10.2 Muscle Strains

The motion of the head leading to extension of the neck stretches the anterior muscles including the sternocleidomastoid muscles (see Table 3.1). One hypothesis is that these muscles are at risk of injury from attempting eccentric contraction during Phase 3 of whiplash motion. Eccentric contraction occurs when a muscle contracts as it stretches. Studies have shown that muscle failure occurs at forces much larger than maximal isometric force and stretch is necessary to create injury (Garret et al. 1997). The contraction is due to the stimulation of muscle spindles in the flexor muscles that are being stretched as the neck and head move into extension – Phase 2. At this stage, the large extensor muscles in the back of the neck are moving into compression and are hence unlikely to contract at the time of impact.

A second hypothesis is that the extensor muscles are injured during rebound of the head and neck as they undergo eccentric contraction during the rebound phase of the impact in Phase 4 (Tencer & Mirza 1998, Hell et al. 2002). Hell et al. (2002) regarded the rebound into the belt system as a possible additional injury source, because the measured head velocities in this phase have been shown to reach higher values than previously expected. This mechanism is consistent with the findings of Garrett et al. (1997) but fails to explain the significant number of belted occupants in severe frontal impacts who do not have neck pain following a crash.

Further, the muscle strain mechanism may explain short-term muscle stiffness following the impact, but such injuries typically last only a few days.

3.10.3 Spinal Column Pressure Pulses

Svensson et al. (1993) conducted an animal study, to investigate the spinal column pressure pulse theory based on an injury mechanism proposed by Bertil Aldman. The necks of pigs were exposed to rapid flexion-extension motion in simulated rear impacts. Pressure pulses of up to 150 mmHg were found in the lower cervical spinal canal (Figure 3.23) during neck motion and were greater in magnitude across the vertebral

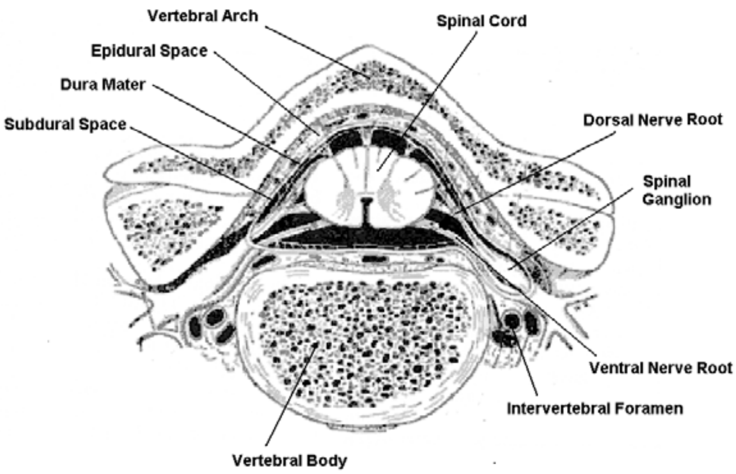


Figure 3.23 Cross-section of the cervical vertebrae with the soft tissues of the spinal canal and intervertebral foramen, after Svensson et al. (1993).

foramen than along the canal. Microscopic analysis of the nerve cells in the spinal dorsal root ganglia (DRG) revealed a leakage of dye from the cerebrospinal fluid (CSF) across the cell membranes, indicating membrane damage.

To investigate this injury mechanism in humans, Eichberger et al. (2000) conducted a total of 21 tests including pressure measurements with 5 cadavers. The sled tests were performed using a test set-up similar to real rear-end collisions. Impact velocities of approximately 9 km/h and 15 km/h were chosen. The subjects were fitted with 2 triaxial accelerometers on the head and chest, one biaxial accelerometer at the height of T1, and one angular accelerometer at the head. Pressure measurements in the CSF were performed using 2 catheter-tip pressure transducers, placed sub-durally in the spinal canal. The upper transducer was placed at the C1/C2 level and the lower transducer at C6/C7. The researchers found pressure peaks reaching 220 mmHg at approximately 100 ms in the cadavers tested. This confirmed the pressure pulse amplitudes and times obtained in the animal experiments by Svensson et al. (1993). Injuries to the nerve tissue in the neck resulting from these pressure effects could not be observed due to the use of cadavers.

3.10.4 Facet Impingement

In a series of related studies by Ono et al. (1997), Kaneoka and Ono (1998) and Kaneoka et al. (2002), volunteer subjects were seated on a sled simulating actual car rear-impact acceleration. The motion patterns of cervical vertebrae in the dynamic crash motion and in normal motion were compared using high-speed radiography. As discussed earlier in this chapter, the forward and upward motion of the torso combined with the inertia of the head leads to an S-shape formation of the cervical vertebrae. The motion leads to compressive and shear loading of the cervical spine. In this phase of the neck motion, Phase 2 in Figure 3.22, the lower cervical spine becomes extended while the upper spine has moved into flexion. Based on the neck radiography from the volunteer tests, the researchers found that the lower motion segments had the larger relative rotation angle. The rotation between the fifth and sixth vertebral segments is the largest and earliest (Figure 3.24).

To quantify this motion, the position of the IAR was analysed for the C5/C6 motion segment (Ono et al. 1997). Volunteer neck measurements provided the expected positions of the IAR within the C6 vertebral body, in normal cervical extension. When the S-shape of the neck occurs in the whiplash motion, the IAR moves upward to a position within the C5 vertebral body (Figure 3.25). This upward motion of the IAR

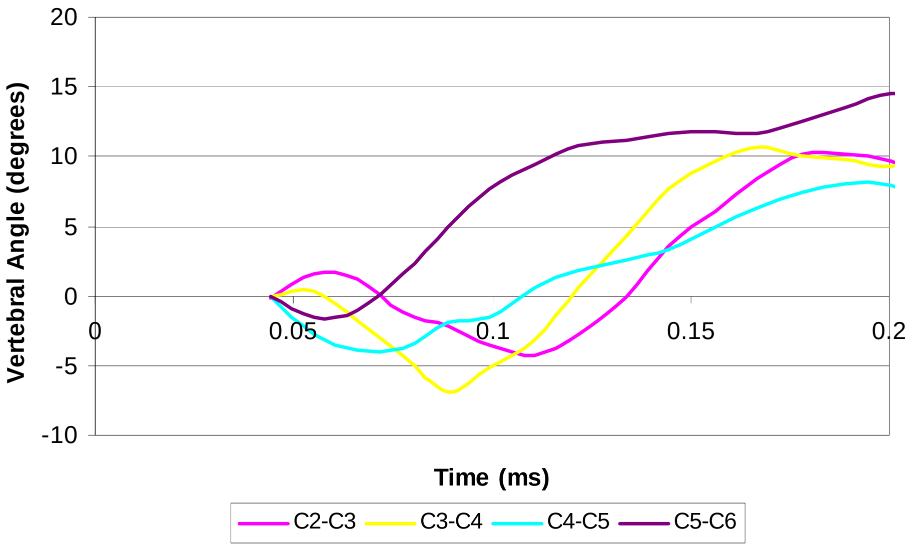


Figure 3.24 Relative rotation of the cervical vertebra for a volunteer (S6) in a rear impact, from Ono et al. (1997).

indicates that the C5 motion at this point is largely rotation rather than shear. The effect of the compression load on the neck is also indicated by the reduction in the intervertebral space in Figure 3.25, which is shown by the relative positions of the static and dynamic C5 vertebral body.

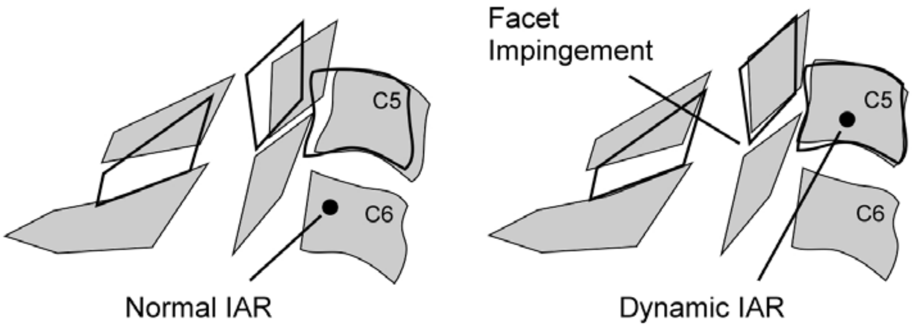


Figure 3.25 With normal cervical extension motion the IAR is positioned in the C6 vertebral body. When the S shape is reached in the whiplash motion, the IAR moves upward to a position within the C5 vertebral body, after Ono et al. (1997).

This upward shift of the IAR was only observed in the C5/C6 motion segment during the crash motion (Kaneoka and Ono 1998). This shift was thought to be the cause of the articular facet surfaces to collide, resulting in mechanical impingement of the synovial fold or meniscoid in the FC (Kaneoka et al. 2002). Further, the researchers hypothesised that if this torque is large enough, there was the possibility of tearing the anterior longitudinal ligament or separating of the anulus fibrosus from the end plate of the associated vertebrae (a rim lesion).

Subsequent testing of cadavers, both head and neck complexes (Yoganandan et al. 1998a; Yoganandan et al. 1998b; Yoganandan et al. 2001; Pearson et al. 2004; Stemper, Pintar & Yoganandan 2004) and whole cadavers (Deng et al. 2000a), has supported the impingement motion of the FC and the possibility of dorsal facet surface compression. In each of these studies significant shear displacement was observed in the FC as well as the rotation. Pearson et al. (2004) characterised the facet displacement during whiplash motion in the following way (Figure 3.26):

A. In the neutral position, FC ligament fibres are perpendicular to the joint surface and have no strain;

B. In the middle of Phase 1, the upper facet slides posteriorly relative to the lower facet and the posterior region of the FC was compressed;

C. At peak intervertebral extension (end of Phase 1), the peak FC compression occurs with the peak FC sliding shortly after;

D. In the middle of Phase 2, the peak FC ligament strain occurred in the anterior as the facets separated while the upper facet was still posterior to its neutral position; and,

E. At the end of Phase 2, the FC ligaments were again perpendicular to the face but strained due to separation of the facets.

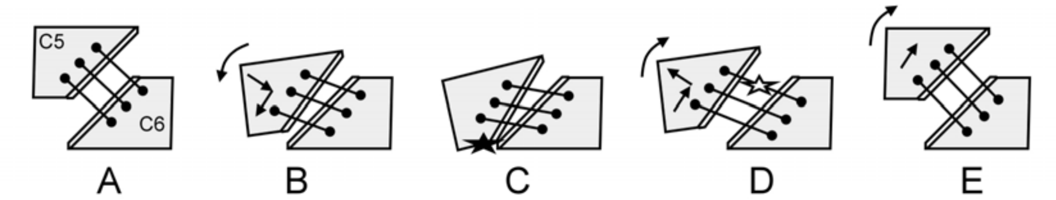


Figure 3.26 Facet motion during whiplash motion as described by Pearson et al. (2004). The black star marks the most likely point of FC impingement and the white star, the peak FC anterior ligament strain.

For a peak acceleration of 6.5 g, Pearson et al. (2004) measured a peak FC compression of 1.8 mm and a peak FC sliding of 4.0 mm at C5/C6, with an average FC ligament strain of 35.9%. This motion of the FC in both rotation and shear, where the posterior edge of the C6 facet surface becomes impinged by the C5 surface (as described by Pearson for the intact cadaver head and neck), varies from that found by Ono et al. (1997) for a volunteer. In the Ono volunteer test, the impingement recorded by the high-speed radiographs (illustrated in Figure 3.22 and Figure 3.22) involved the posterior edge of C5 encroaching on the C6 surface, due to the rotation of the FC alone with no shear displacement. This variation may have been the result of the initial posture of the volunteer in the test.

To investigate the FC impingement hypothesis further, Inami, Kaneoka and Ochiai (2000) dissected 20 cervical spines to gain anatomical data of the cervical facet joint meniscoid. Although cervical facet joints consist of three types of inclusions: fat pads, fibro-adipose meniscoid, and capsular rims, Inami, Kaneoka and Ochiai (2000) were not concerned with the capsular rims and fat pads in their study as they were too short to be impinged. Three types of meniscoids were identified:

1. Type 1 meniscoids consist principally of adipose tissue with a small amount of fibrous tissue. These are less exposed to mechanical stress as they are crescent-shaped and exist only around the peripheral space;
2. Type 2 meniscoids are thicker around the free borders compared to Type 1 and contain dense fibrous tissue around the apical region. This suggests that even small structures have the potential to be exposed to mechanical stress; and,
3. Type 3 meniscoids are thin and their free borders are ragged with the middle and apical regions formed exclusively by fibrous tissue.

There are no data on the size of meniscoid required for impingement, but five large examples of elliptic-shaped Type 2 meniscoids in the sample were thought by the researchers to project sufficiently to be impinged by the articular facets.

The facet impingement has the support of the injury found from the testing of volunteers, cadavers and the accident based autopsy data.

3.10.5 Shear

In a rear impact the torso is pushed forward by the seatback while the head remains stationary, straightening the thoracic spine. The inertia of the head converts this vertical motion of the spine into a compression loading to the cervical spine. This compression has been observed in volunteer and cadaveric tests simulating whiplash. A shear force is generated at each level of the cervical spine to pull the head forward. This shear force was suggested by Yang and Begeman (1996) to be a likely candidate to cause soft tissue injury to the intervertebral joints of the cervical spine. While under compression, the cervical vertebrae slide relative to each other as a result of the shear and the FC stretched and possibly torn, resulting in inflammation and pain.

Deng et al. (2000a) carried out 26 low-speed rear-end impacts on six human cadavers in a rigid seat. It is important to note that there were multiple tests on individual cadavers in these tests. The study showed that the upper cervical vertebrae go into relative flexion with respect to the lower cervical vertebrae during whiplash motion, while the entire neck is in extension (the S-shape). In addition, the upper neck is under flexion when the head contacts the head restraint, while the facets reach peak strain prior to head contact with the head restraint. It was concluded that if stretching of the facet capsular ligaments were the reason for the high incidence of neck pain, the upper cervical spine would sustain a flexion injury while injury to the lower cervical spine would be due to a combination of shear and compression. Deng et al. (2000a) also reported that a 20-degree seatback compared to a 0-degree seatback resulted in less cervical lordotic curvature, more upward ramping motion of the thoracic spine, and greater relative rotation of each cervical motion segment.

3.10.6 Head Impact with the Head Restraint

Several researchers have mentioned the possibility that the actual impact with the head restraint may be injurious. Croft (1998) suggested that there was a risk of cervical injuries at the moment of the first contact between head and head restraint. Such injuries may be likely even if the restraint is properly positioned. Intuitively, the phasing of the 'S' response of the neck in a rear impact leading to the stretching of the major ligaments to near the physiological limits combined with interacting with a poorly designed head restraint may increase the likelihood of injury. "Poorly designed" in this case could be poor positioning with respect to the centre of gravity of the head or of inappropriate stiffness (Winkelstein & Myers 1998). Immediately following head contact the upper cervical spine will be forced into acute as the inertia of the neck continues to draw it rearward, since there is no contact with either seat back or head restraint (Geigl et al. 1994).

Hell et al. (2002) suggest that for females the lighter head and lower muscle mass may drive higher head rebound accelerations following contact with the head restraint.

3.11 Neck Injury Assessment Criteria

3.11.1 Introduction

Many studies have proposed vehicle related crash parameters such as delta V (the speed change of the vehicle in the crash) and the change in vehicle acceleration for use in assessing the risk whiplash associated neck injury. These vehicle related parameters have been shown to not correlate well with the load applied to the neck of the vehicle occupant, but are modified by the seat response in particular (Haland et al. 1996; Bostrom et al. 1998).

Kullgren et al. 2003 used real crash data from the Swedish Folksam study to investigate proposed injury criteria. The study used 79 rear end crashes with 110 front seat occupants who were followed up to assess any pain outcomes. The study used simulation techniques using a validated BioRID II dummy and seat model to reconstruct the impacts and found that mean vehicle acceleration was a useful means of assessing the likelihood of whiplash associated injury risk, with a threshold of about 7 g the risk approached 100%. The study also reviewed whether the following neck injury criteria correlated with the pain outcome:

- NIC proposed by Bostrom et al. (1996), based on the relative horizontal velocity between the bottom T1 and top C1 of the cervical spine (see below for more discussion);
- N_{km} proposed by Schmitt et al. (2002), based on a combination of the upper neck shear and the flexion/extension moment of the neck (see below for more discussion);
- NDC proposed by Viano and Davidsson (2002), based on the angular and linear displacement response of the head relative to T1 of the HIII dummy; and,
- Lower Neck Moment proposed by Prasad, Kim and Weerappuli (1997) and based on the HIII dummy.

The study found that both NIC and N_{km} correlated well with the short and longer term pain outcomes of the occupants. NDC and Lower Neck Moment performed poorly, mostly likely due to being based on the HIII responses. The HIII dummy is a poor representation of a person in a rear impact.

Siegmund et al. (2005) applied the seven proposed whiplash neck injury criteria to a structured test series (n = 90 tests) using the BioRID dummy seated on a programmable sled. Six of the criteria showed graded responses that were most sensitive to the average acceleration of the sled. Two criteria N_{ij} and N_{km} were best able to distinguish between the 15 pulse shapes in the series. This study confirmed that NIC correlated well with early head motion (in the first 85 ms).

3.11.2 The Neck Injury Criterion, NIC

Bostrom et al. (1996) developed a neck injury criterion (NIC) based on a mathematical model of the transient pressure pulses measured by Svensson et al. (1993) in the spinal canal of pigs. These pulses were due to volume changes resulting from forcing the head and torso to translate horizontally relative to each other. Bostrom et al. hypothesised that a soft tissue injury to the neck with long term consequences would occur during the initial head/thorax motion, when the spine takes the 'S' shape as the thorax is pushed forward. Anatomically, this is a retraction motion of the neck and it occurs in the first 100 ms of the rear impact, before the head begins to rotate. Injury was thought likely to occur if:

$$NIC = a_{rel} * 0.2 + v_{rel}^{\ 2} > 15\, m^2/s^2$$

where a_{rel} and v_{rel}, are the relative acceleration and velocity between the head (C1) and the upper torso (T1).

The pressure amplitude for the pig and the human should be very similar as the constants in the predictive model are scale insensitive and the dimensional factor is similar. The criterion of 15 m^2/s^2 was felt to be appropriate for humans.

Darok et al. (2000) validated the NIC for volunteer tests (n = 70), cadaver tests (n= 28) and dummy tests. The testing confirmed aspects of the use of NIC:

- For the volunteers, the peak NIC correlated well with the maximum retraction of the head;
- For the volunteers, no complaints of pain were made at an NIC below 8, while some complaints of pain were made at NIC values of about 10;
- For the cadavers, a ligament rupture occurred at an NIC of 18.6; and,
- For the cadavers, NIC correlated with the magnitude of the peak pressure readings in the spinal canal.

A simplified version, NIC_{max}, has been accepted for use (Bostrom et al. 2000).

$$NIC_{\max} = \text{maximum}_{\text{first 150ms}}\left(a_{rel} * 0.2 + v_{rel}^{\ 2}\right)$$

This simplification uses the head centre of gravity motion instead of C1, based on the assumption that there is little head rotation until full retraction occurs.

Kullgren et al. (2003) used a group of 79 rear-impact crashes with known injury outcomes and a crash-pulse recorder fitted to the vehicle to validate the NIC_{max}. The crashes were all reconstructed with a BioRID II dummy and seat validated with sled testing. The study found that a NIC_{max} threshold of 15 showed relatively high positive predictive values and very high negative predictive values for neck injury with long lasting symptoms (greater than 1 month).

NIC_{max} has been shown to be sensitive to the major risk factors of a rear impact such as crash pulse, seat deflection characteristics and head-to-head restraint distance (Bostrom et al. 2000).

3.11.3 The N_{km} Criterion

The N_{km} is based on the N_{ij} criterion, which is a linear combination of compression load (F_z) and flexion/extension moment (M_y) developed to predict serious injury to the neck in frontal impacts (Kleinberger et al. 1998). The N_{km} criterion was developed for rear impacts and uses shear force (F_x) and flexion/extension moment (M_y) (Schmitt et al. 2002).

$$N_{km}(t) = \frac{F_x(t)}{F_{\text{int}}} + \frac{M_y(t)}{M_{\text{int}}} < 1.0$$

The shear force (F_x) and flexion/extension moment (M_y) are both obtained from the upper load cell in the dummy neck and the F_{int} and M_{int} are the critical intercept values used for normalisation. N_{km} characterises all phases of the neck motion in a rear impact, where NIC_{max} is based on maximum retraction and occurs early in the motion in Phase 2.

Kullgren et al. (2003) also used the results of the crash reconstructions to validate N_{km}. The study found that an N_{km} was also applicable to the prediction of neck injury with long lasting symptoms (of greater than 1 month), but the best predictor was a combination of NIC and N_{km}.

3.12 Summary

In this chapter, many of the factors influencing the biomechanics of cervical spinal soft-tissue injury have been brought together. The factors include aspects of the neck anatomy, clinical data, autopsy data, and the results of many experimental studies using animal, human cadaver and human volunteer models to investigate these types of injury. The converging trend, in the results of the many field-, clinically- and experimentally-based studies as noted by Barnsley, Lord and Bogduk (1998), is becoming more apparent.

Barnsley, Lord and Bogduk (1998) have pointed out that many patients with injuries to the discs and joints associated with whiplash may be expected to have prolonged pain as an outcome, with little chance of healing or spontaneous recovery. The connection between chronic pain outcomes and the proven injury to structures of the neck, such as the facet joints, is strengthening. The structures of the neck most likely to be injured in whiplash are the FC, the intervertebral discs and the upper cervical ligaments. The FC regions of the neck have been shown to be major sources of post-crash pain. There are two main hypotheses of how injury may occur and these are related to events early in the impact. To investigate these theories, it is necessary to develop an investigative tool to establish the causal links connecting the mechanical load to the neck in a motor-vehicle crash and the symptoms of WAD sufferers.

CHAPTER 4 SELECTION OF INVESTIGATIVE PROCESS

4.1 Introduction

There exists a significant volume of research investigating WAD from the perspective of its incidence in field accident studies, clinical investigation, post-mortem examination and testing, and volunteer testing. What is absent is a means of connecting these areas in order to link the causes and effects of the WAD injury mechanisms. The focus of this chapter is to select an appropriate methodology to examine these soft-tissue neck injury mechanisms. The main areas covered, and the context of this chapter within the thesis, are shown in Figure 4.1.

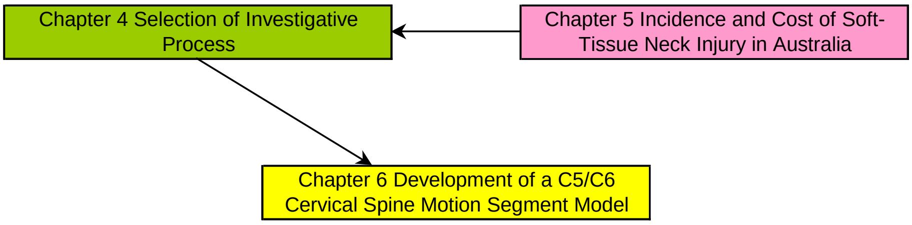


Figure 4.1 The main areas covered and the context of this chapter within the thesis.

4.2 The Available Investigative Models

Numerous models have been used in determining human injury response and injury criteria.

The simplest model to interpret is where a human volunteer is instrumented and subjected to a controlled series of impacts at increasing levels. The threshold of the impacts is set to avoid injuring the volunteer. This method has certain drawbacks:

- The measurements taken can only indicate the threshold of minor injury;
- Data on the causation of more severe injuries are not generated;
- Problems with attaching the instrumentation so as not to injure the subject may produce less reliable data;
- Individual differences among volunteers lead to problems with determining injury thresholds for the average population;

- The effects of muscle tension, learned behaviour and involuntary reactions are also difficult to control; and,
- A bias may be introduced since the majority of volunteer tests have been run on groups not representative of the average population, such as the young and fit or military personnel.

A second method uses cadavers, or post-mortem human subjects (PMHS) in the form of whole body or body sections. These specimens are instrumented and subjected to impact forces. Autopsies show the injuries incurred, which are then correlated with the measurements taken. This type of testing has the advantage that at least the framework of the surrogate has some resemblance to a live human. The problems that exist with this test methodology include:

- Available subjects are generally older and tend to be more degenerated than the average population and so may not give a representative response;
- Subjects are in short supply;
- It is very difficult, if not impossible, to position an intact neck in a reasonably life-like posture;
- The response of the cadaver depends on the pre-test treatment (frozen, embalmed, or fresh, etc.);
- The effects of pressurisation of body systems such as the airways, vascular system and the central nervous system are missing without special techniques;
- The effects from active musculature are absent unless special techniques, such as embalming, are used to simulate muscle tension; and,
- Some signs of injury on a living subject (eg. muscle strain, pain, or loss of consciousness) are undetectable.

The third method uses animal surrogates or models to estimate human responses to impact. In the past, primates and pigs have been used to study automotive injury and it has been very useful in defining injury mechanisms. Using anaesthetized animals can also provide more information on how injury and vital signs are related. Acquiring ethical approval to test animals in this way is becoming increasingly difficult. The biggest drawback of this method is the difficulty of transforming the animal response measurements into human injury criteria due to the effect of the differences in the animal anatomy, responses and injuries.

Injury criteria may also be developed through accident reconstruction using an anthropomorphic test device (ATD) or dummy. If the accident parameters are well recorded, and the victim's injuries are fully documented, then accidents can be reconstructed by testing with a surrogate for the victim. This technique is more useful in developing dummy-based injury protection reference values, rather than human injury criteria. The measurements of the dummy responses can be paired with the injuries (or lack thereof) recorded for the victim. In this way it is possible to develop injury criteria based on the specific dummy. Issues to be considered in relation to this method include:

- Assessment of the adequacy of the reproduction of impact conditions is highly subjective;
- Dummies are only available in a limited series of sizes, so the anthropometry of the victim may not be closely matched;
- Measured dummy responses are dependent on the instrumentation capabilities of the dummy;
- Dummies available do not have perfect biofidelity, and can only approximate human response;
- This lack of biofidelity of the dummy becomes more marked where serious injury occurs, for example when fractures occur;
- Injury criteria developed are not necessarily applicable to humans; and,
- Injury criteria may apply only to the specific dummy type used in the reconstruction.

A variation of this reconstruction technique is the use of mathematical models of the crash victim rather than the physical dummy simulations, to reproduce a well-documented accident. The increase in computing power has supported major advances in the application of mathematical modelling to biomechanics.

The difficulty in carrying out realistic mechanical experiments is due to the lack of effective surrogates for live humans. This has encouraged researchers to use mathematical modelling. While volunteers may only be tested at sub-injurious levels, the use of cadavers and animals raises significant ethical and compatibility issues.

There is a growing trend in investigating human tolerance by using simplified human models to reconstruct field accident data with known injury outcomes (Gibson et al. 1985; Kullgren et al. 2003). The reliability of the procedure is limited by the capability of the computer simulation to reproduce the impact response of the victim. An advantage of computer models over dummies is the ability to match the physical characteristics and anthropometry of the accident victim more closely. When these are applied to the analysis of biomechanical experiments, the quantification of mechanical parameters, which cannot be assessed in other ways, is possible (Wismans 1995). However the accuracy and reliability of a mathematical model is dependent on the availability of biomechanical information on the system being represented as well as the assumptions made in formulating the model.

A mathematical model allows "surrogate experiments" with absolute repeatability (Goel & Clausen 1998). The user is able to vary parameters to observe the effects in both kinematics and internal loading. A mathematical model is able to enhance the results of an experiment by explaining those results and quantifying estimates of internal loads, which are not measurable experimentally. A mathematical model can be used in the design of experiments by indicating critical parameters and allowing the experimental work to be focussed.

4.3 Mathematical Modelling of the Crash Victim

4.3.1 Historical Development

Whole body simulation of the vehicle occupant as a structure in crash conditions has been extensively investigated since computers of sufficient processing power became available in the late 1960s. Two programs in particular have been widely used for crash victim simulation: the ATB (Articulated Total Body) model (Obergefell et al. 1988), and the MADYMO (Mathematical Dynamic Modelling) program (TNO 1999).

In their original forms, these whole body simulation models, such as MADYMO, were formulated to describe a crash test dummy in planar or three-dimensional motion in a crash environment. The dummy was simulated by a set of rigid body segments – with prescribed masses and moments of inertia linked by various types of joints – in an open loop system or tree structure. The complexity of the models can be varied with the number of rigid bodies used. The governing equations of motion for these multiple tree structures of rigid bodies connected by joints in an acceleration field are derived automatically using Lagrangian methods (Prasad & Chou 2002). The relative motions between segments of the system are resisted by non-linear springs, viscous dampers and Coulomb friction. The shape and stiffness characteristics of the model body segments are described by use of ellipsoids or hyper-ellipsoids, which are attached to the rigid body links. The system interacts with the environment through contact planes and ellipsoids that develop resistive forces on the segments in contact. The amount and rate of penetration between ellipsoids into planes, and ellipsoids into ellipsoids, are used to develop non-linear spring forces on the segments in contact.

Such lumped mass or multi-body models are fast in terms of model preparation and computer run-times. This makes them particularly suitable for use in development as many alternatives can be evaluated in a relatively short time. These types of models are well suited to dealing with the kinematics or gross motions of the occupant. On the other hand, the finite element (FE) models require long model preparation and computer run-times, but are better able to model contact situations and material strains more accurately. This gap between multi-rigid body simulation and finite element modelling approaches is gradually disappearing as hybrid models are established (Rzymkowski 1999). This move is being supported by recent developments in the available software such as MADYMO, which has integrated finite element and multi-body approaches.

4.3.2 Attributes of Multi-body and Finite Element FE Modelling Methods

Historically the dynamic modelling of crash victims was an area dominated by multi-body methods due to the capability of such models archiving adequate precision, or biofidelity, with simplified systems and small numbers of parameters. These simplified systems allow this type of mathematical model to be adapted for the specific problem being investigated, with easier physical interpretation of the results. Hence multi-body models are capable of achieving good kinematics similitude whilst being both computationally efficient and easy to interpret. At the same time, these simpler systems have difficulty coping with certain types of injury mechanisms, for example contact stresses and the internal stresses and strains caused by the applied loading within the rigid body components. The rigid body components are usually equated with the bony structures in a biomechanical model. At injury causing levels of stress and strain this assumption is not necessarily correct.

With the continuing improvements in computing power in the last decade, finite element (FE) models have been regarded as the more capable method for biomechanical modelling. An FE model could be very detailed and take into account minor details in internal structures, even down to the level of cell structure if necessary. Such models also gave detailed quantification of the stresses and strains in the system. This capability came at a cost however, with detailed models being very time consuming to set up and run. To adequately represent the system to the required level of complexity, it is necessary to incorporate significant details in the parameters that describe the system. This level of detail is not yet found in the data available for many biomechanical systems although it is improving as the need for modelling drives the experimental work. Lack of available data for many parameters involved in dynamic loading of the human body has made it difficult to achieve reasonable levels of biofidelity of the FE model responses. Better results can still often be achieved more simply by well-designed multi-body methods.

Kleinberger (1998) reviewed the mathematical modelling of the cervical spine. He suggested that the discrete parameter or multi-body models were most suitable for kinematic analyses of head and neck motion and the FE models were more appropriate for investigating tissue stresses and strains. He rated the major limitation of mathematical models for biomechanics as the poor quantification of the material properties because of the lack of both knowledge and of well-supported biomechanical material models in commercial FE analysis programmes. The definition of the muscles and their active responses was the next major limitation and there was a need for even faster hardware on which to run the models. He proposed that the most important recent development in biomechanical modelling was the ability to directly scan actual components of the body to generate the structural components for a model.

At the current state of development of human body models it is not entirely possible, or desirable, to have a single all-purpose model capable of complete human impact response. This has led to the development of a compromise approach – the hybrid model. In this approach, a simple, multi-body kinematic model is enhanced in the critical regions by a model with the required level of detail (Rzymkowski 1999). This enhancement may be achieved through combining the mathematical whole-body model with a more detailed multi-body or FE model in the region being investigated. These developments have led to adaptations of modelling software to support the hybrid approach.

4.4 Development of Human Head and Neck Models

The examination of soft-tissue neck injury mechanisms requires the use of a mathematical model as an investigative tool. The complexity of a simulation model must be related to the level of detail that is available to define the task and the accuracy required of that task (Currie & Gibson 1996). The mathematical simulation of dynamic human head and neck responses can be categorised into three different approaches:

- Simple two-pivot models;
- Multi-body models; and,
- Finite element models.

The two-pivot models are simple three-segment, two-pivot models of the head, neck and torso, and are used to describe the global motion of the head and neck (Figure 4.2).

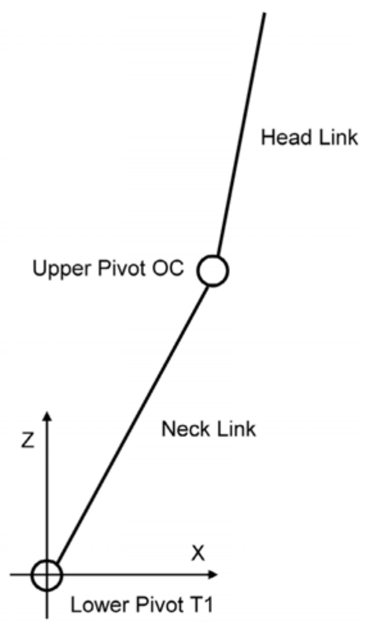


Figure 4.2 A simple three-segment, two-pivot model of the head, neck and torso used to describe the global motion of the head and neck of a volunteer (Wismans et al. 1987).

Applications of these models include motion analysis of volunteers and dummies in crash situations (Wismans et al. 1987). Whilst being a useful tool for analysing the kinematics of volunteers in impact testing, such simple models are incapable of describing vertebral kinematics and soft-tissue loads. To acquire the necessary detail to simulate the internal forces and motions in the neck, more complex solutions are required. In multi-body models, the head and vertebrae are depicted as rigid bodies with their masses lumped at their centres of gravity, and connected by massless, deformable elements, which represent the intervertebral soft tissue and muscles.

Deng and Goldsmith (1987) developed such a lumped-mass model with highly detailed representation of the cervical spine geometry and materials (see Figure 4.3).

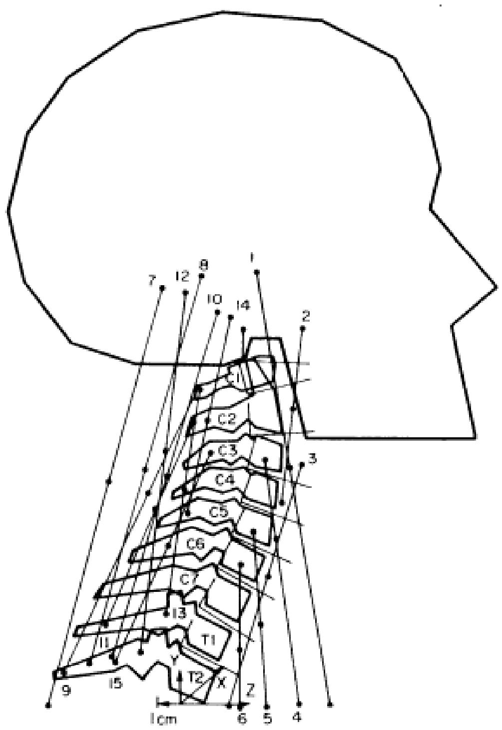


Figure 4.3 The multi-body mathematical model of the head and neck developed by Deng and Goldsmith (1987).

This model was rewritten for use with the MADYMO program by de Jager (2000), who further developed the model capabilities and carefully validated it for frontal and lateral impacts. This basic model proceeded through several stages of development. Initially it was simplified as the global model, which did not have active muscle capability. The global model consists of nine rigid bodies for the head (C0), the seven cervical vertebrae (C1 - C7), and the first thoracic vertebra (T1). The rigid bodies are connected by non-linear, viscoelastic elements with load displacement characteristics derived from current test data. The lower joints in the neck model have similar characteristics, but the two upper joints for reasons of fidelity have unique characteristics. The joint stiffness included the effects of muscle behaviour.

Once this global neck model had been satisfactorily validated against the Naval Biodynamics Laboratory (NBDL) volunteer tests in frontal impacts, further development was undertaken to improve the model responses. The development began by including separate representations of the intervertebral discs, ligaments and facet joints for each motion segment (Figure 4.4). The final step was the inclusion of the Hill-type muscle elements. An animation of the de Jager model with muscles of a cadaver rear impact test is included in the CD-ROM attached in the Appendices.

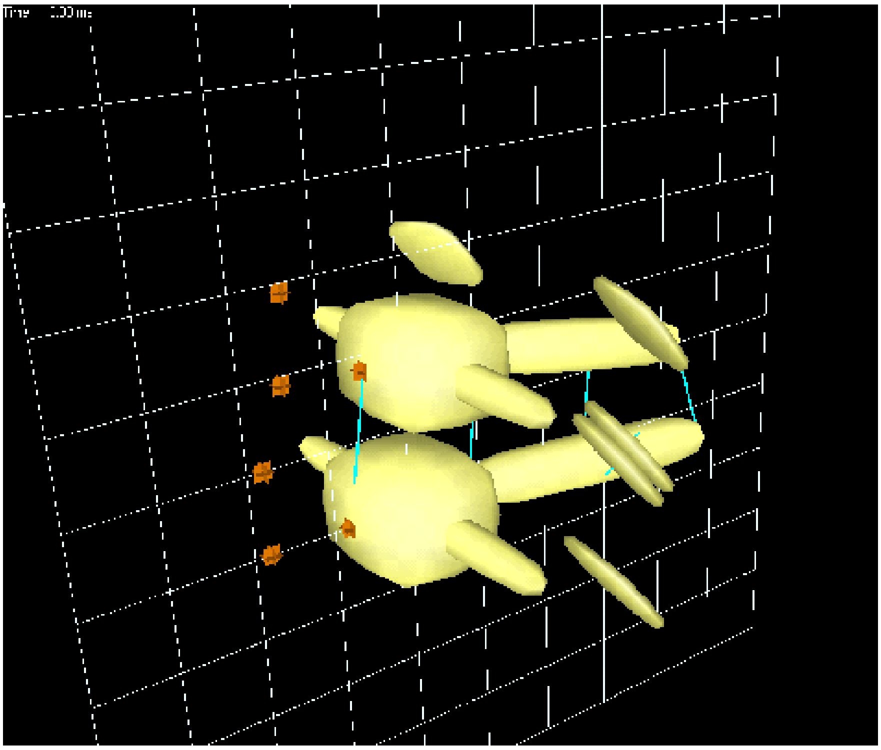


Figure 4.4 The detailed C5/C6 motion segment including separate representation of the intervertebral discs (by a six degree of freedom joint), ligaments (by spring/damper units in blue) and facet capsules (by sliding surfaces) of the de Jager (2000) head and neck model.

De Jager included a total of 14 mid-sagittally symmetrical pairs of the stronger and more superficially located muscles in the model. The muscles were given a simplified geometric representation with a straight line between the origin and insertion points, a method often used in biomechanics to simplify the calculation of muscle reaction forces. These muscles were able to model both passive and active states. The model was then validated for both frontal and lateral impacts with the re-analysed response corridors derived from the NBDL volunteer sled tests by Thunnissen et al. (1996). The de Jager model became the basis for the next stage in the development of a MADYMO-based, mathematical human body model with the necessary realism in its kinematic responses.

Van der Horst (2002) developed a major enhancement to the capability of the de Jager head and neck model by improving the active muscle capability to include curved lines of action. The development was continued into other areas of the neck model and included further work on the validation (van der Horst 2002). The resulting model is described in more detail in the following section of this Chapter.

Finite element models of the head and neck require highly detailed representations of the neck geometry and material behaviour. The model by Kleinberger (1993) was developed to investigate the mechanisms of cervical spine injury in motor vehicle crashes. This model has had only nominal validation published and has been used mainly for investigating the effects of parameter changes. It was based on a single motion segment, which was replicated throughout the cervical spine with only minor dimensional changes, and contained many simplifying assumptions. The author notes that the model required considerable computational resources.

Nitsche et al. (1996) generated a FE human neck model, which was validated for frontal and lateral impacts. Two major difficulties with the model were found, including the quantification of the material properties for the model, and with the validation of the discrete neck model, which required the use of artificial boundary conditions. Several FE models of individual cervical motion segments have also been developed. The Kumaresan, Yoganandan and Pintar (1998) representation of the C4/C6 unit is the most anatomically correct and is based on a single specimen using CT scans. This model was used for a parametric study of the spinal components under static compressive loading. The model was further developed to study the effects of surgical modifications, age-related spinal degeneration and paediatric spinal responses by scaling (Yoganandan et al. 1998c). Similarly, Goel and Clausen (1998) developed a C5/C6 motion segment model, which included ligaments and a composite disc, and was validated for a variety of quasi-static loads. These models are not suitable for dynamic loading situations.

Yamazaki, Ono and Kaneoka (2000) developed a dynamic finite element model of the head and neck, which was validated with respect to the relative vertebral angle at the C5/C6 level.

The FE models of the human neck tend to fall into one of two types: kinematic models or detailed motion segment models. The kinematic models such as that by Choi et al. (2000), which can be validated dynamically, generally offer little gain over the equivalent multi-body models. The detailed motion segment models are only capable of dealing with quasi-static loading of a detailed neck structure. When these model types are combined, the required computational times increase greatly (Kleinberger 1993).

4.5 The van der Horst Human Head and Neck Model

The van der Horst (2002) model is a detailed multi-body model of a fiftieth-percentile human head and neck. It consists of a rigid head, rigid vertebrae (C1, C2, C3, C4, C5, C6, C7 and T1), non-linear visco-elastic discs and ligaments, frictionless facet joints, and controllable, segmented contractile muscles.

The vertebral shapes in the van der Horst model are based on the scanned vertebra of an individual's neck, and are represented as lumped masses. The inertial and geometric representations were based on those used in the de Jager model (de Jager 2000). The inertial properties of the neck were derived from NBDL volunteer data analysed by Thunnissen et al. (1995). The initial positions of the vertebrae were based on lateral x-rays of a group of young men standing erect, performed by Nissan and Gilad (1984). The parameters used in the model are summarised in Table 4.1. The position and orientation of each body is described relative to the adjacent lower body (in the direction of the thorax) using a right-handed coordinate system (see Notation). The position of the centres of gravity for each body is given in the local coordinate system while the principal moments of inertia are defined with respect to a coordinate system parallel to the local system with its origin at the centre of gravity (CofG) of the body. The orientation and size of the facets were derived from Panjabi et al. (1993). The centre of the FC was assumed to be symmetrically disposed around the vertebral CofG in the x and y planes and at a distance behind the CofG equivalent to half the facet diameter. The facet orientations are given in Table 4.2.

Table 4.1 Inertial properties and positions of the rigid bodies of the neck with respect to the lower body, from de Jager (2000)

No.	Neck Segment	Mass	Inertia			Origin		CofG		Initial Position
			I_{xx}	I_{yy}	I_{zz}	S_x	S_z	G_x	G_z	
		(kg)	($kg.cm^2$)			(mm)		(mm)		(°)
1	T1	-	-	-	-	0.0	0.0	-	-	0.0
2	C7	0.22	2.2	2.2	4.3	6.4	16.8	-8.2	0.0	20.8
3	C6	0.24	2.4	2.4	4.7	-2.0	18.4	-8.3	0.0	-5.6
4	C5	0.23	2.3	2.3	4.5	-2.8	17.4	-8.1	0.0	-5.2
5	C4	0.23	2.3	2.3	4.4	-3.3	17.2	-7.9	0.0	-4.7
6	C3	0.24	2.4	2.4	4.6	-4.0	17.8	-7.8	0.0	-5.3
7	C2	0.25	2.5	2.5	4.8	-3.3	18.7	-7.7	0.0	0.0
8	C1	0.22	2.2	2.2	4.2	0.0	16.5	-7.7	0.0	0.0
9	C0	4.69	181.0	236.0	173.0	-4.0	20.0	27.0	43.0	0.0

Table 4.2 The orientations of the facet surfaces with respect to the horizontal plane, van der Horst (2002).

No.	Segment	Orientation of Upper Facet	
		α_z (°)	α_y (°)
1	T1	2.3	-24.0
2	C7	4.0	-28.8
3	C6	3.6	-28.3
4	C5	4.0	-30.6
5	C4	4.0	-40.5
6	C3	45.8	-47.7
7	C2	-28.6	0.0
8	C1	-	-

The intervertebral disc is represented as a simple, six degree of freedom joint, modelled as a parallel spring/damper for each degree of freedom – translation and rotation, see Figure 4.5. The local coordinate system originates at the centre of the disc, with the x-axis lying anterior to posterior.

The stiffness of the disc in axial rotation, lateral bending and shear are based on the *in-vitro* experimental work by Moroney et al. (1988), while Pintar et al. (1986) provided the disc stiffness in tension. The non-linear compression stiffness was based on that calculated for the lumbar disc by Eberlein, Frohlich and Hasler (1999). The non-linear flexion-extension stiffness was based on the values used by Comacho et al. (1997) in a non-linear mathematical model used for near vertex impacts. The stiffness of the motion segment was equally attributed to the ligaments and the disc by assumption. The values used for the motion segment stiffness are summarised in Table 4.3. Damping for the model was based on that used by de Jaeger (2000), however this did not account for dynamic stiffening. Hence the dynamic stiffness was estimated as twice the static stiffness and was applied by a multiplication factor selected by the user.

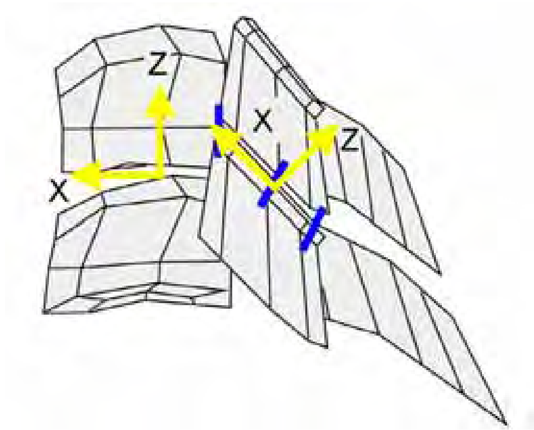


Figure 4.5 Lateral view of the isolated C5/C6 motion segment from the van der Horst (2002) head and neck model including representation of the intervertebral discs by a six degree of freedom joint (large yellow x/z axes) and facet capsules by ligaments (by spring/damper units in blue) and PR sliding surface (by small inclined yellow x/z axes).

Table 4.3 Stiffness and damping values for the intervertebral discs in van der Horst (2002).

Direction of Load	**Stiffness (N/mm)**	**Damping (Ns/m)**	**Source**
Anterior Shear	62	1000	Moroney et al. (1988)
Posterior Shear	50	1000	Moroney et al. (1988)
Lateral Shear	73	1000	Moroney et al. (1988)
Tension	53	1000	Pintar et al. (1986)
Compression	822-2931	1000	Eberlein et al. (1999)
	Stiffness (Nm/rad)	**Damping (Nms/rad)**	
Flexion	0.022-5.4	1.5	Comacho et al. (1997)
Extension	0.022-8.2	1.5	Comacho et al. (1997)
Lateral Bending	0.33	1.5	Moroney et al. (1988)
Axial Rotation	0.42	1.5	Moroney et al. (1988)

In the van der Horst model, Figure 4.4, the facet capsule is represented by a point restraint (a simple sliding joint by a single three-dimensional translation spring/damper) at the centre of the surface. The z-axis is directed perpendicular to the facet surface and the x-axis is aligned in the anterior to posterior direction. The frictionless surface of the synovial joint is represented by a uni-directional stiffness in the z direction, which is arbitrarily assumed to be twice the stiffness of the disc. Four tension-only spring/damper units represent the capsular ligaments and the stiffness values are derived from Yoganandan et al. (1998c), see Table 3.6 and Table 3.7 in Chapter 3.

The major neck ligaments (including the ALL, PLL, LF and ISL) also have stiffness characteristics derived from the work of Yoganandan et al. (1998c) (Table 3.6 and Table 3.7 in Chapter 3).

The representation of the muscles in the van der Horst head and neck model follows the curvature of the neck, with realistic lines of action of the muscle forces and the ability to be actuated as active muscles (van der Horst 2002).

The motion segment component of the van der Horst model was validated quasi-statically with respect to the low load *in-vitro* test data of Moroney et al. (1988). The entire head and neck model was dynamically validated in rear impacts with volunteer test data. In this validation process the MADYMO human body model was used, as it was found that the neck response depended on the seat response and the response of the spine.

A separate study, Meijer et al. (2001), was aimed at further validating the MADYMO 50th-percentile male (1.74m, 75.7 kg) human body model, which incorporates the van der Horst (2002) model as the head and neck. This study used the complete human model to simulate the JARI volunteer sled tests (Ono et al. 1997). The sled test series involved 9 healthy male volunteers (average 26 years, 1.76m, 71 kg) subjected to low speed rear-impact sled impacts in both an upright rigid seat and a standard vehicle seat. The seats were mounted on the sled on a ramp of 10 degrees to the horizontal. The maximum sled acceleration was approximately 3.6g. The human model was found to soundly predict the response of head and neck in slow speed rear impact. The area, which was found to be least satisfactory in the model, was that the ramping up of the spine in the human model was too small in comparison with the JARI volunteers. The researcher found that the results of this study verified that the van der Horst (2002) human head and neck model was:

- Capable of the necessary biofidelity with respect to its kinematic responses;
- Capable of being used as part of a whole human body model.

The study by Stemper, Yoganandan and Pintar (2004) validated the van der Horst model over a range of relatively low velocity experiments with cadaver heads and necks. The same researchers conducted experiments, which provided more certainty with the validation (Stemper, Yoganandan & Pintar 2003b). The global head-neck angle, segment angle and local facet joint regional kinematic responses from the model fell within experimental corridors. This was shown for impact velocities of 1.3, 1.8 and 2.6 m/s. The non-physiologic 'S' shape curve was found by the tests to have duration of 100 ms, which matched the experimental data. The experimental data did not have the effect of active muscle response or the vertical acceleration caused by straightening of the thoracic spine of the subject nor did it have the compression effect of spine straightening.

4.6 Discussion

The study, of the causation of soft tissue neck injury in rear impacts, required an appropriate investigative tool. A significant amount of data is already available from animal, volunteer and cadaver testing as well as from field accident and clinical studies. What is required of such a model is the ability to link together the available data. The aim of the development and use of the model is to find causative links between the biomechanical test data and the injury data to allow the refining and testing of the various injury hypotheses. Mathematical models of the head and neck have reached a point of development where they are capable of making these links. The investigation of whiplash injury mechanisms required the model to possess a high level of dynamic biofidelity of the human body at various levels, as part of a whole human body model, as a head and neck model and as a neck motion segment model. High computational efficiency was required of the model to manage the multiple runs, which are required to reconstruct real life accident case data. Further the model had to be able to predict the injury at the level of the soft tissue components of the motion segment.

The various mathematical models described here were insufficiently developed to be able to meet the requirements of this study. The multi-body type of model was more suited to studying the kinematics of the head and neck in impact situations. The available FE models were better suited to the investigation of the fracture mechanics and stresses in the materials of the components of the neck under quasi-static load conditions.

For the combination of the response of the human body, the kinematics of the head and neck during a rear impact and the strains in the soft tissues of the neck required here, a multi-body mathematical model with detailed soft tissue for the motion segment appeared to be the best option. Even with the limitations imposed by the multi-body format, such a model could be developed to investigate soft tissue neck injury causation at the motion segment level. The detail of the available information regarding both the anatomy and the properties of the soft tissue components were suited to such an approach. It was necessary that this detailed motion segment be integrated into a head and neck model for the purposes of dynamic validation and application in realistic dynamic loading situations. A head and neck model is the smallest viable component to be genuinely able to be used in this manner. To be useful, it was also necessary for the head and neck model to be able to be integrated into a full model human body model, again for the purposes of validation with volunteer tests and for the application of the model for the investigation of soft tissue neck injury in rear impacts in motor vehicles. The model, which best fulfils these requirements, is the van der Horst head and neck model. This model has been validated with respect to both volunteer and cadaver data, including to the level of vertebral angulation. It also has active muscle capability and a compatible whole body model is available for future development. It was chosen to be used as the basis of the development of the detailed motion segment model described in Chapter 6.

CHAPTER 5 THE INCIDENCE AND COST OF SOFT-TISSUE NECK INJURY IN AUSTRALIA

5.1 Introduction

This chapter presents a preliminary investigation into the incidence and cost of WAD in Australia. This work also assisted in validating the approach selected for the investigation process discussed in Chapter 4. It places the incidence and outcomes of the long-term symptoms (chronic pain) of whiplash-associated injury into the context of Australia. The main areas covered, and the context of this chapter within the thesis, are illustrated in Figure 5.1.

The investigation gave a useful understanding of the characteristics of the long duration pain outcomes of whiplash-associated injury. The first stage of this part of the study uses the chronic neck pain database developed by the Cervical Spine Research Unit (CSRU), University of Newcastle, Australia (Gibson et al. 2000). The crash characteristics of these cases were investigated to gain a better understanding of the circumstances leading to chronic neck pain in the drivers. The results were then combined with other published incidence data to estimate the incidence and cost of these injuries in Australia separate from the insurance claims made (Ryan & Gibson 1998). The limitations in these estimates are discussed and a comparison is made with insurance claim costs (MAA 1999).

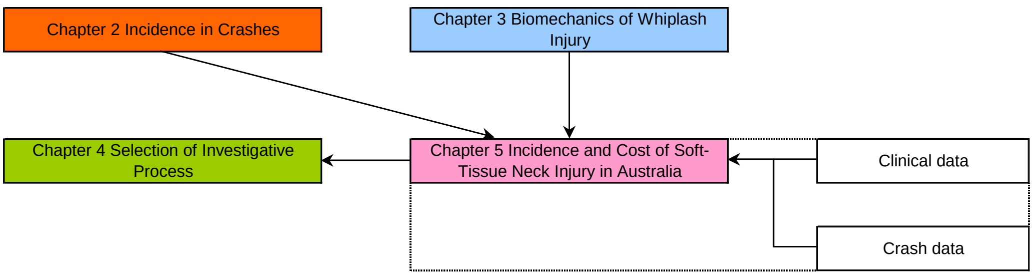


Figure 5.1 The main areas covered and the context of this chapter within the thesis.

5.2 Extension of the clinical data to include the crash

5.2.1 CSRU Clinical Data

Researchers at the CSRU successfully devised and validated the use of fluoroscopically controlled, double blind, differential diagnostic anaesthetic blocks for the objective diagnosis of cervical FC pain (Barnsley et al. 1995). The technique involves anaesthetising or "blocking" the FC by injecting local anaesthetic directly into the joint space, blocking the nerve supply to the joint – the medial branches of the dorsal rami.

The CSRU collected data during the course of these studies on the clinical investigation and treatment of patients with chronic neck pain (Gibson et al. 2000). Many of these patients were drivers, other vehicle occupants, pedestrians and bicyclists. To gain knowledge of the accident circumstances, the cases were matched with the P4 Collision Report Forms from the NSW Police records, which include a description of the accident and a sketch of the crash scene; the age and make of the vehicle, an estimate of its speed on impact, the damage incurred, and whether it required towing from the scene; a post-accident injury description, details of any treatment received at the scene, and whether seat-belts were worn.

The RTA database was used to trace accidents in which the patient was the driver of the vehicle. The RTA only encodes a subset of the accidents from the P4 forms to use as a basis for the NSW road accident statistics. The criteria used to encode an accident are as follows:

- It was reported to police;
- It occurred on a road open to the public;
- It involved at least one moving vehicle;
- At least one person was killed or injured; and,
- At least one vehicle was towed away.

The search was by patient surname, date of birth and date of accident for the years from 1985 to 1996 (Gibson et al. 2000).

The subjects in this study are all patients with a proven source of chronic pain in the facet region of the neck, who were drivers of a motor vehicle in a crash. The biases in the data have no effect on this thesis.

5.2.2 Results of the Analysis

Accident reports for 92 drivers were found in the NSW accident records, consisting of 51 females (55%) and 41 males (45%) (Gibson et al. 2000). The police accident data for these 92 cases was combined with the diagnostic data from the CSRU database. The age of the subjects ranged from 19 to 68 years (mean age of the females = 36.4 years and of the males = 39.4 years). All except two of the car drivers in the group were encoded as wearing seat belts. The majority of the accidents occurred in the Newcastle metropolitan area: some occurred on rural roads and in the Sydney metropolitan area. Seventy accidents (76%) involved a single impact. There were 16 cases (17%), which involved two major impacts and six (6.5%) with multiple impacts – including three rollovers (3%). Thirty seven (40%) of the impacts were from the rear and 31 (34%) were frontal. The lateral impacts were almost even with 10 (11%) from the left and 11 (12%) from the right. Twelve (32%) of the collisions from the rear involved a double impact, with rear followed by a frontal impact. The proportions of males and females in the lateral and rear impacts in the sample were similar (see Table 5.1).

A measure of accident severity available in the police reports was whether the vehicle damage was great enough to require towing from the scene. Seventy vehicles (76%) were towed following the accidents, 15 (15%) were not towed and the rest were unspecified. Twenty-one (68%) of the vehicles involved in frontal impacts and 28 of those in rear impacts (76%) required towing from the scene.

Table 5.1 Gender and impact direction for the 92 cases from Gibson et al. (2000).

Impact Direction	**Female**		**Male**		**Overall**	
	N	**(%)**	**N**	**(%)**	**N**	**(%)**
Front	19	(37)	12	(29)	31	(34)
Left	5	(10)	5	(12)	10	(11)
Rear	18	(35)	19	(46)	37	(40)
Right	6	(12)	5	(12)	11	(12)
Rollover	3	(6)	0	(0)	3	(3)
Total	51	(100)	41	(99)	92	(100)

The police accident records contained only limited injury data. Thirty-six cases (39%) were not treated, 50 (54%) were treated but not admitted to hospital, and 6 (6.5%) were admitted to hospital. The injuries listed for those treated at hospital were fractured neck, lacerations, spinal injuries, fractured ribs, lower back pain, neck injury, and fractured foot. With regard to injuries listed for the cases not admitted to hospital, no injury was reported for 53 (58%) of the cases, and neck pain for 33 (36%) of the cases.

The 92 cases matched with the Police accident reports included a total of 88 positive diagnoses of FC pain (Gibson et al.2000). A 'positive' diagnosis of cervical FC pain was made when the patient had been diagnosis based on the double blind, differential diagnostic anaesthetic blocks (Barnsley et al. 1995). Sixty-eight (74%) cases had at least one positive diagnostic block, no positive blocks were found in nine (10%), and in one case the pain resolved during the study. The remaining 14 (15%) had not completed the procedure at the time. The bilateral positives were treated as single positives on the assumption that they were associated with the same injury-exposure event. The distribution of positive blocks by level of the neck is shown in Table 5.2.

Table 5.2 Positive diagnostic FC blocks (n = 88) by level of the neck for the 92 CSRU drivers (Gibson et al. 2000).

Segment	Position of the Positive Blocks			
Level	Bilateral	Left	Right	Total
C2/C3	6	7	17	30
C3/C4	1	1	8	10
C4/C5	1	1	3	5
C5/C6	1	11	16	28
C6/C7	1	8	6	15
Total	10	28	50	88

In this group with positive blocks, females were slightly more represented than males (55% to 45%); rear-end impacts were more common (40%); and, few of the subjects had injuries requiring hospitalisation (6.5%).

The positive diagnosis of symptomatic FC occurred more at C2/C3 (34%) and C5/C6 (32%) levels of the neck with right side symptomatic Z-joints more predominant. Right-side positive blocks are more apparent in the males than females at the C2/C3 and C5/C6 joints in the sample, see Table 5.3.

The authors of the study (Gibson et al. 2000) found that the leakage of cases was due to the following factors: records were only available for drivers: with licenses; only 70% of the police accident forms met the RTA encoding criteria; women who changed their maiden names could not be traced; some cases occurred prior to 1985; hyphenated names were not available from the search; errors in the accident dates; work site accidents; and, interstate accidents.

Table 5.3 Distribution of the 88 positive blocks (n =92) by gender and level of neck (Gibson et al. 2000).

Intervertebral Joint		Position of the Positive Blocks			
		Bilateral	Left	Right	Total (%)
C2/C3	*f*	5	5	9	19 (21.6)
	m	1	2	8	11 (12.5)
C3/C4	*f*	-	-	3	3 (3.4)
	m	1	1	5	7 (8.0)
C4/C5	*f*	1	1	2	4 (4.5)
	m	-	-	1	1 (1.1)
C5/C6	*f*	1	9	9	19 (21.6)
	m	-	2	7	9 (10.2)
C6/C7	*f*	1	5	3	9 (10.2)
	m	-	3	3	6 (6.8)
Total (%)		10 (11.4)	28 (31.8)	50 (56.8)	88 (100)

5.3 Estimates of Incidence and Cost of WAD in Australia

5.3.1 Introduction

The results from the previous section were combined with several other Australian field studies to estimate the numbers of WAD only cases in Australia per year and their costs separate from more usual data collected from insurance claims (Ryan & Gibson 1998). The aim of this estimate was to examine the occurrence and persistence of symptoms of this type of injury in Australia unbiased by the differences in claim outcomes in the various state jurisdictions.

5.3.2 Australian Field Studies

The estimates are based on a set of Australian field studies, which are briefly described below. Each of these studies was based on a selected and possibly unrepresentative sample. The aim of the study was to make an estimate using non insurance claims based incident data, and so this limitation had to be accepted due to the lack of alternatives.

Ryan et al. (1993) identified individuals (n = 32) with neck strain after a car crash from physiotherapy and general medical practices in Adelaide. The subjects were interviewed and physically examined by a manipulative physiotherapist soon after the crash and again after six months. For each case the vehicle and the crash site were examined and the crash reconstructed.

Fildes and Vulcan (1995) studied 120 whiplash-injured car occupants presenting at a teaching hospital in Victoria. A vehicle examination noted impact direction, damage, and headrest and seat position. The injuries were monitored to measure the timing of the WAD, with chronic defined as symptoms six months or later after the event.

Dolinis (1997) obtained driver characteristics, crash circumstances and injury data from structured interviews of 246 Adelaide drivers, who had recently experienced a rear impact of sufficient severity to be reported to the police.

5.3.3 Estimate of incidence

Dolinis (1997) interviewed 246 subjects of which 86 (35%) suffered neck symptoms to some degree. Of those with symptoms, 34 (39.5%) were restricted in their activities of daily living for at least one day, 57 (66.3%) consulted a medical or other health practitioner, three (3.5%) attended a hospital emergency department, and for 20 (23.2%) symptoms persisted for three months or more.

Ryan et al. (1993) identified thirty-two cases from general and physiotherapy practices as having neck symptoms. Six (18.7%) had attended a hospital emergency department and one was admitted, who also had a number of other injuries.

The 92 drivers with chronic neck pain, as a result of motor vehicle crashes from the CSRU, consisted of 36 cases (39%) with no medical treatment immediately post-crash, 50 (54%) treated but not admitted to hospital, and six (7%) admitted to hospital (Gibson et al. 2000). For the cases not admitted to hospital, no injury was reported for 53 (62%), and neck pain for 33 (38%).

In the four financial years from 1993/94 to 1996/97, there were 2,238 claims for neck injury to the Insurance Commission of Western Australia, of which 76 (3.5%) were admitted to hospital (Dyke, Ryan & Hendrie, 1999). This suggests that for those with insurance claims, the ratio of non-admitted to admitted cases is about 30:1 (Ryan & Gibson 1998).

These studies provide a measure of the spectrum of intensity of whiplash symptoms, allowing a rough estimate of the relative frequency to be made at each level of treatment (Ryan & Gibson 1998). From Ryan et al. (1993), there was one hospital admission in every 32 cases – a similar ratio to that observed in the Insurance Commission claims data. In addition, it is recognised that a medical practitioner had examined all cases in this study. From Dolinis (1997), the ratio of persons with symptoms to those attending a medical practitioner was 1.5:1. The ratio of GP attendances to emergency department attendances for Ryan et al. was 5.3:1 compared with 19:1 for Dolinis. This suggests that the latter cases were less severe (Ryan & Gibson 1998).

Finally, for the subjects with chronic neck pain from Gibson et al. (2000), the ratio of non-admitted patients to admitted was 15:1, of treated to admitted was 8.3:1 and the ratio of those just with symptoms to those treated at the time of the event was 0.67:1. This suggests that the events contributing to these cases were significantly more severe than those in the previous studies (Ryan & Gibson 1998).

5.3.4 Estimate of number of cases per year

Between 1993 and 1997, Western Australia had an average of 65 admissions to hospital for whiplash injuries, Dyke et al. (1999). By extrapolation and using the ratios derived above, it is possible to estimate that there were around 3,000 whiplash cases occurring each year in Western Australia, with about 2,000 consulting general practitioners (Ryan & Gibson 1998). This is summarised in Table 5.4.

Table 5.4 Estimated annual number of WAD cases in Western Australia (Ryan & Gibson 1998).

WAD events	Estimated number
Hospital admission	65
Emergency department attendance	103-368
General practitioner attendance	1,950
Persons with symptoms	2,925

From Ryan et al. (1994), about 66% of GP attendance cases demonstrated persisting symptoms at six months post injury, which is defined as *chronic* according to Fildes and Vulcan (1995). This permits an estimate of 1,287 new chronic cases per year in Western Australia (Ryan & Gibson 1998). The Insurance Commission of WA averages 560 new whiplash claims per year, suggesting that not every case of whiplash results in an insurance claim (Ryan & Gibson 1998). Further, as Western Australia comprises only 10% of the Australian population, there could be as many as 30,000 new cases of whiplash symptoms – of varying severity – each year, including up to 13,000 new chronic cases (Ryan & Gibson 1998). Moreover, this suggests an annual incidence rate of WAD in Australia of approximately 167 per 100,000 (based on a population of 18 million).

5.3.5 Costs

Hendrie et al. (1998) estimated that the average total cost of a neck injury of low severity on the Abbreviated Injury Scale AIS 1 (equivalent to WAD cases) was \$18,114. The major components of this total were general damages, pain and suffering (31.1%), legal and investigation costs (15.9%), economic loss or loss of wages (14.4%), and hospital and medical costs (8.8%).

For Western Australia the total cost of WAD cases could be 3,000 x \$18,000 = \$54,000,000 annually. This is about 8.5% of the estimated annual \$637 million cost of all road crashes in Western Australia (Hendrie 1999). For Australia, the total cost of WAD cases could reach \$540 million each year (Ryan & Gibson 1998).

By comparison in NSW the claims where whiplash was the only injury accounted for 11.2% (n = 14,196) of all claims and 3.8% (\$222.9 million) of the total insurance cost (MAA 1999).

5.4 Summary

The police accident data were combined with the diagnostic data for 92 drivers suffering chronic neck pain from the CSRU database. The accident characteristics associated with these drivers suffering chronic neck pain are similar to those found for more general WAD discussed in Chapter 2. There were more females than males and symptomatic facet joints were most commonly found at the C2/C3 or the C5/C6 levels of the neck.

The incidence of neck pain as a result of rear impacts in Australia was estimated and found to be greater than the insurance claims data.

The best supported of the available injury causation hypotheses are the shear and the facet impingement hypotheses based on volunteer and cadaver neck responses, as reviewed in Chapter 3. These results confirm that the most productive area of the neck to investigate for whiplash injury causation in rear impacts is at the C5/C6 motion segment level. Chapter 6 outlines the development and validation of a multi-body model of the C5/C6 motion segment to form the basis of this investigation.

CHAPTER 6 DEVELOPMENT OF A C5/C6 CERVICAL SPINE MOTION SEGMENT MODEL

6.1 Introduction

The discussion in Chapter 4 concluded that a mathematical model of the human head and neck was required to investigate the mechanisms of injury resulting in neck pain as a result of rear impact in a motor vehicle. The head and neck model must not only have kinematic rear impact response biofidelity, but also reflect the detailed anatomy of the soft tissue of the neck motion segments. Mercer and Bogduk (1999) have pointed out that a typical adult disc structure departs from the image commonly presented in anatomy texts. These researchers also hypothesised that the kinematics of a cervical motion segment and its soft tissue injuries mechanisms are dependent on this structure of the disc (Bogduk & Mercer 2000). The development of a joint model incorporating this disc structure and able to reproduce the motion at the segment level would be of use in improving our understanding of the injury mechanisms involved. Such a model would be able to assist in the development of soft-tissue neck injury criteria.

The C5/C6 level motion segment was chosen as the most appropriate starting point for the development for the following reasons:

- C5/C6 is very similar to C3/4 and C4/5 and would allow easy development of a complete neck;
- It is one of the two most commonly injured levels of the neck in rear impacts (Gibson et al. 2000);
- Injury at the C5/C6 level of the neck has been shown to be the source of chronic neck pain by clinical studies (Barnsley, Lord & Bogduk 1998); and,
- An injury mechanism hypothesis, based on volunteer testing (Ono et al. 1997) and cadaver testing (Yoganandan, Pintar & Kleinberger 1998b; Cusick et al. (2001); Panjabi et al. 1998), is linked with this specific level of the neck.

The aim of this part of the study is the development a mathematical C5/C6 motion segment model, which, by having both biofidelity and injury sensing capability, would be able to investigate the causation of soft-tissue neck injury in rear impacts.

With the multiple models being described here, it is important to understand which model is referred to. Henceforth the word model will refer specifically to the mathematical models being discussed. There are three models, which will be regularly mentioned: the original van der Horst human head and neck model and its component parts; the newly developed version of the C5/C6 motion segment model; and the van der Horst human head and neck model with the new C5/C6 motion segment, which will be referred to as the ‘head and neck model’.

6.2 Model Development

6.2.1 The Development Process

The development stages of the new C5/C6 motion segment model are shown in the flow chart in Figure 6.1. The starting point for the development of a new head and neck model was to isolate the C5/C6 motion segment from the original van der Horst (2002) human head and neck model, see Chapter 5.

A review was undertaken to ensure that the data used to define the C5/C6 motion segment were the most appropriate available, to make a change required direct experimental data to be available. A parameter study of the motion segment geometry and the physical parameters such as ligament stiffness and damping was conducted. Based on the parameter study several areas of the motion segment were found to be capable of improvement, specifically the representation of the disc and the bearing surfaces of the facet capsule (FC). Changes were developed to ensure:

- Improved detail response biofidelity;
- Full multi-directional response capability of the motion segment;
- Improved representation of the soft tissue elements for injury sensing purposes; and,
- That the overall responses remain compatible with the van der Horst head and neck model.

At each stage in the development, the new C5/C6 model motion segment model was examined to ensure that it still met the multi-directional response requirements based on the *in-vitro* quasi-static low load tests by Moroney et al. (1988) (described in detail in Appendix A4). Where these requirements were not met then the modifications were either further developed until the model response was improved or returned to the original van der Horst parameters.

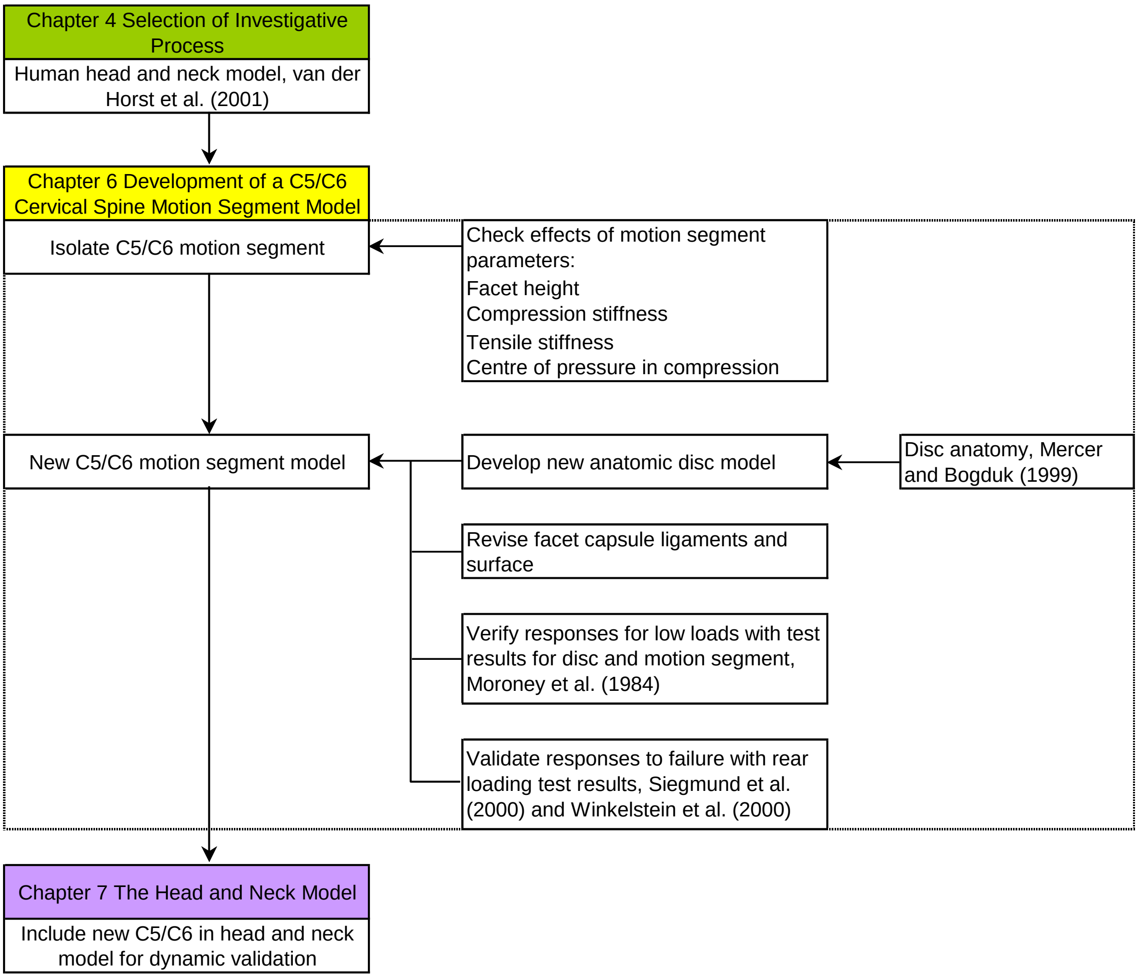


Figure 6.1 Flow chart of the steps in the model development and validation.

The model C5/C6 motion segment responses were then compared with *in-vitro* testing based on quasi-static loads to failure. The tests selected for this final quasi-static validation of the motion segment model were the two studies of complex loading of a single motion segment designed to represent possible rear impacts by Siegmund et al. (2000b) and Winkelstein et al. (2000). A detailed description of the *in-vitro* test methodology used in these studies for the excised motion segments is included as Appendix A4.

The final stage of the validation process was to incorporate the new C5/C6 motion segment model into the van der Horst head and neck model for dynamic response comparisons with test data obtained from volunteers and cadavers. This stage of the model validation process is described in Chapter 7.

6.2.2 Development of the New C5/C6 Motion Segment

Once the C5/C6 motion segment was isolated, the specific modelling parameters were reviewed (see description of the van der Horst head and neck model in Chapter 4). As the intent was to remain as consistent with the original van der Horst head and neck model as possible, the review was limited to the C5/C6 motion segment parameters. Where alternative data was available for the model parameters a check was made of the sensitivity of the model to these parameters. The parameters investigated, the sources of the new data and their effects on the model are summarised in Table 6.1. If the parameters did not have an appreciable effect on the response of the motion segment, the original values used in the van der Horst motion segment model were retained.

Table 6.1 The motion segment model parameters investigated, the sources of data and the effect on the model responses.

Parameter	van der Horst (2002) source	Effect on model	Alternative data source
Disc height	de Jager (1999)	No	CSRU data (Appendix A2)
Facet surface angle	de Jager (1999)	Yes	Nowitzke, Westaway and Bogduk (1994)
Facet surface height	de Jager (1999)	No	Nowitzke, Westaway and Bogduk (1994)
Facet surface compression stiffness	As for the disc x 2	No	None found
Facet surface area	Point restraint only	Yes	Panjabi et al. (1993)
Disc compression stiffness	Eberlein et al. (1999)	No	Shea et al. (1991)
Disc surface area	Point restraint only	Yes	CSRU data (Appendix A2)
Anulus fibrosus ligaments	Point restraint only	Yes	Mercer and Bogduk (1999)

The van der Horst (2002) head and neck model uses the ligament stiffness data measured by Yoganandan, Kumaresan and Pintar (2000), which was the most appropriate data available (Chapter 3). The same source provided the neck ligament length for the calculation of relative strain in the new C5/C6 motion segment model (Chapter 3). The ligament stiffness data was extrapolated linearly past the point of failure.

The disc in the van der Horst (2002) model consists of a 6-degree of freedom joint at the centre of the intervertebral joint. The motion of a cervical motion segment is due to the interaction of the disc, the ligaments and the facet capsule, as described in Chapter 3. Mercer and Bogduk (1999) found that the adult cervical anulus fibrosus does not consist of concentric laminae of collagen fibres as in the lower discs, but is arranged in a crescent shape. This crescent is thick anteriorly and tapers laterally towards the uncinate processes. Posteriorly the anulus consists only of a thin layer of vertically oriented fibres. Anteriorly the fibres are arranged at approximately 45° to the vertical axis of the neck. Based on this anatomy described by Mercer and Bogduk, a model of the disc was developed to be included in the C5/C6 motion segment model. The sources for the material properties used in the components of the motion segment model are given in Table 6.2.

Table 6.2 The material properties of the C5/C6 motion segment components.

Parameter	Modelling Technique	Material Properties
Ligament	Kelvin Restraint	Pintar et al. (1986) Yoganandan, Kumaresan and Pintar (2000)
Anulus Fibrosus	Kelvin Restraint 3D point restraints (x2)	Eberlein et al. (1999) Pintar et al. (1986) Moroney et al. (1988)
Nucleus Polposa	3D point restraints (x4)	Eberlein et al. (1999) Pintar et al. (1986) Moroney et al. (1988)
Facet Joint	1D point restraints (x4)	van der Horst (2002)

In this new disc model, the anulus fibrosus is represented by nine tension-only, spring/damper units inclined at 45°. A double anterior row represents the depth and the inclination of the ligament fibres within the crescent and a vertical spring damper unit for the posterior region of the anulus. The stiffness characteristic for each of these spring damper units was set to 11% of the axial tension (positive z direction) stiffness of the original 6 degree of freedom joint. The overall tension stiffness of the motion segment was unchanged. These tension-only ligaments are illustrated in the drawing of the motion segment components in Figure 6.2, and pictorially in Figure 6.3.

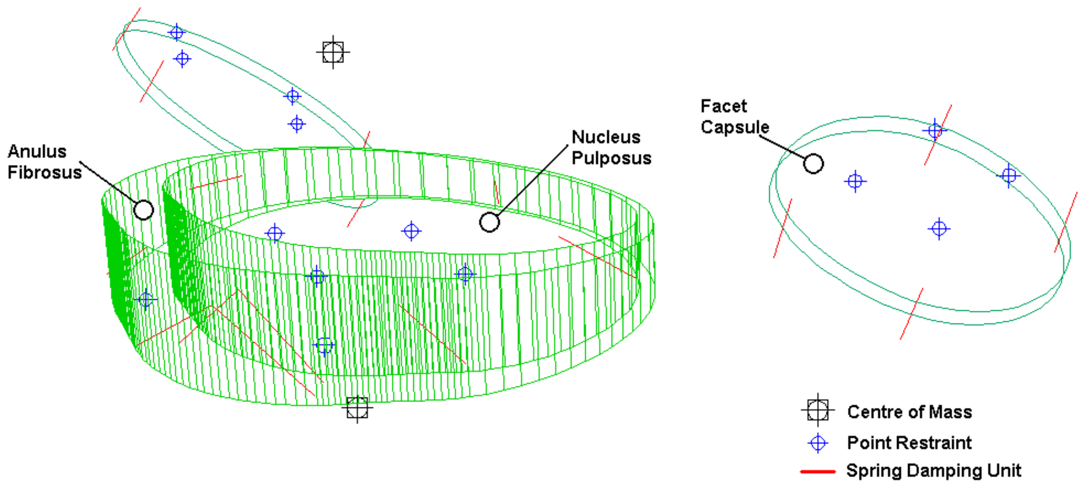


Figure 6.2 The soft tissue components of the new C5/C6 motion segment model, showing the disc, with the tension-only ligaments (in red) and the compression point restraints (in blue).

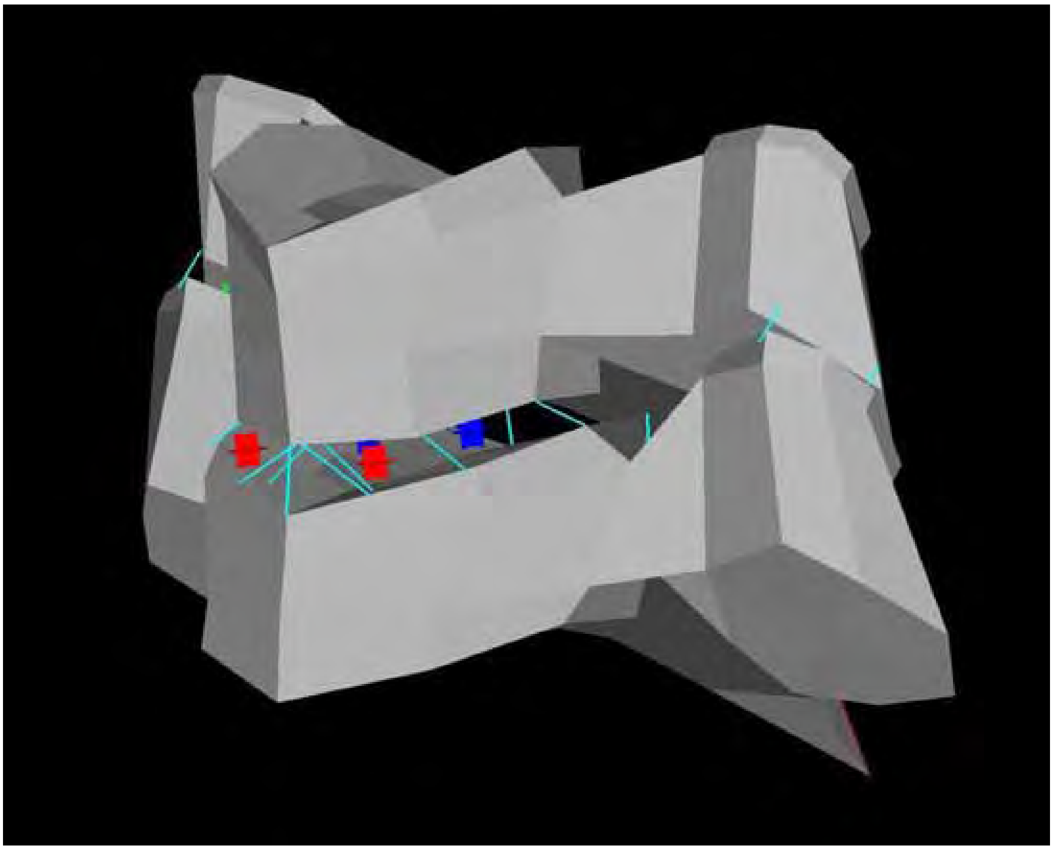


Figure 6.3 Oblique pictorial view of the C5/C6 model motion segment, showing the scanned vertebral bodies (in grey), the ligaments (in light blue) and the compression point restraints for the AF (in red) and NP (in blue).

Moroney et al. (1988) demonstrated that the stiffness of fully degenerated discs decreased by 50% as the nucleus pulposa (NP) aged. In a fully degenerated disc, the nucleus pulposa appears to loose its compressive capability and the stiffness of the disc in this loading direction is due to the anterior section of the anulus fibrosus. The bearing surface of the new disc model is represented by six compression-only point restraints (Figure 6.2). Two are for the crescent shaped anulus and the remaining four represent the circular pad of the nucleus pulposus. A point restraint in MADYMO is a three degree of freedom joint, which can translate in the three axes x, y and z. For compression (negative z-direction), only the x and y axes are defined to have zero stiffness. The overall compression stiffness of the van der Horst disc remained unchanged, but instead of acting at the centre of the disc, it was distributed over the bearing surface of the disc itself.

By separating the compression and tension stiffness to representations of the specific soft tissues in the joint, the disc model was then able to indicate relative component strains in the disc. These component strains can be used as indicators of possible injury. The new disc model was combined with the ALL and PLL to represent the disc-only configuration in the tests performed by Moroney et al. (1988) for low loads (Table 6.3). The multi-axial responses of this disc unit were then adjusted by replacing some of the stiffness in the 6-degree of freedom joint to correctly simulate the test results. The 6-degree of freedom joint retains the damping for the motion segment (see Chapter 3).

Table 6.3 The loading applied to both the intact motion segment and disc-only segments by Moroney et al. (1988).

Loading	Symbol	Load direction	Intact segment	Disc only
Compression	COM	- F_z (N)	73.6	73.6
Anterior shear	AS	+ F_x (N)	19.6	15.7
Posterior shear	PS	- F_x (N)	19.6	15.7
Right lateral shear	RLS	+ F_y (N)	19.6	15.7
Flexion	FLEX	+ M_y (Nm)	1.8	1.6
Extension	EXT	- M_y (Nm)	1.8	1.6
Right lateral bending	RLB	- M_x (Nm)	1.8	1.6
Counter clockwise torsion	CCW	+ M_z (Nm)	1.8	1.6

The new disaggregated disc model was placed within the C5/C6 motion segment model with the original van der Horst (2002) facet capsule structure and other ligament definitions. Initial verification simulations of the Moroney et al. test procedure on this complete motion segment indicated the need for further development. In particular, the motion segment model gave excessive rotation during the posterior shear test, which is of some importance in whiplash loading. These responses were improved by shifting the centre of pressure of the model motion segment. This was accomplished by including an array of four compression-only point restraints representing the surface of the facet capsule and extending it to the rear. These changes give the facet capsule a sliding surface rather than a single point (see Figure 6.2 and Figure 6.3). These changes also improved the ability of the motion segment model to detect the type of facet surface impingement injury hypothesised by Ono et al. (1997), Yoganandan, Pintar and Kleinberger (1998) and Panjabi et al. (1998). The geometries of the four facet capsular ligaments were also made more consistent.

The final geometry of the soft tissue in the motion segment model is shown in Figure 6.2 and Figure 6.3. During the development of the internal structural representation of the motion segment, the overall stiffness of the van der Horst C5/C6 motion segment in tension and compression were maintained to ensure that the responses of the new motion segment were compatible with the rest of the neck. The next step was to complete the validation of the C5/C6 motion segment model responses by comparison with experimental results.

6.2.3 C5/C6 Motion Segment Low Load Response Verification

The responses of the disc model were developed to meet the multi-directional, quasi-static small load responses of the *in-vitro* experimental results for excised disc-only motion segment responses measured by Moroney et al. (1988). To measure the equivalent responses of the disc-only C5/C6 motion segment model, the experiments were set up and simulated with the model following the same loading and test methods used by the authors. The experimental methodology is described in detail in Appendix A4. The lower vertebra (C6) was clamped and the loads were applied to the upper vertebra (C5) of the model. The loading cases measured are given in Table 6.3. The simulation was repeated for the intact motion segment.

6.2.4 C5/C6 Motion Segment Large Load Response Validation

Following the completion of the development process for the new C5/C6 motion segment, the model was validated for larger quasi-static combined loads up to failure by comparing the simulation results obtained from two experimental studies by Siegmund et al. (2000b) and Winkelstein et al. (2000).

Siegmund et al. (2000b) tested (n = 7) excised C3/C4 and C5/C6 motion segments. A combination of compressive preloads (0 N, 45 N, 197 N and 325 N) and posterior shear loads (0 N, 27 N, 49 N, 79 N, 98 N and 127 N) were applied to the motion segment. The loads in the experimental procedure were chosen to reflect those calculated for the shear and compression in the necks of volunteers during rear impact testing by Ono et al. (1997). The compression load is due to the inertia of the head during *upward* motion of the torso, while the shear is caused by the inertia of the head during *forward* motion of the torso. Upward and forward motion of the torso results from seat back pressure during a rear impact. The translation and rotation of the loaded motion segment were measured and the motion segments were loaded to failure. A more detailed discussion of the test methodology used by Siegmund et al. is given in Appendix A4.

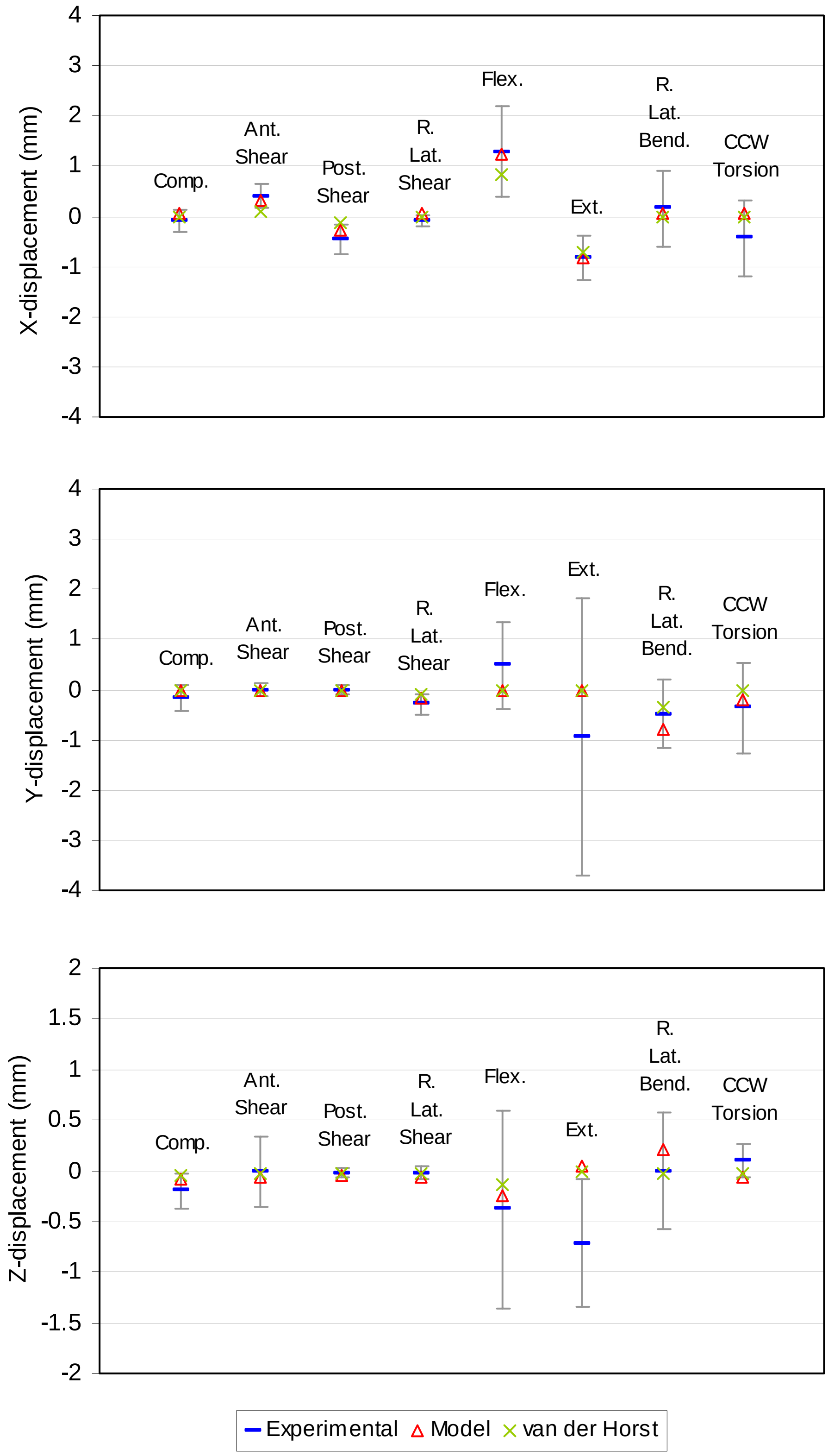
In a related investigation, Winkelstein et al. (2000) tested six excised C3/C4 and C5/C6 motion segments. Three types of axial preload were applied about the z-axis: ipsilateral (clockwise) torsion (+1 Nm), no preload (0 Nm) and contralateral (counter-clockwise) torsion (-1 Nm). In this set of experiments, the aim was to look at the effects of combined loading of the neck. The axial preload was equivalent to a person in a rear impact with the neck axially rotated such as might occur if they were looking at the rear view mirror at the time of the impact from the rear. The response of the motion segment in flexion and extension was measured. A more detailed account of the test methodology used by Winkelstein et al. is given in Appendix A4.

6.3 Results

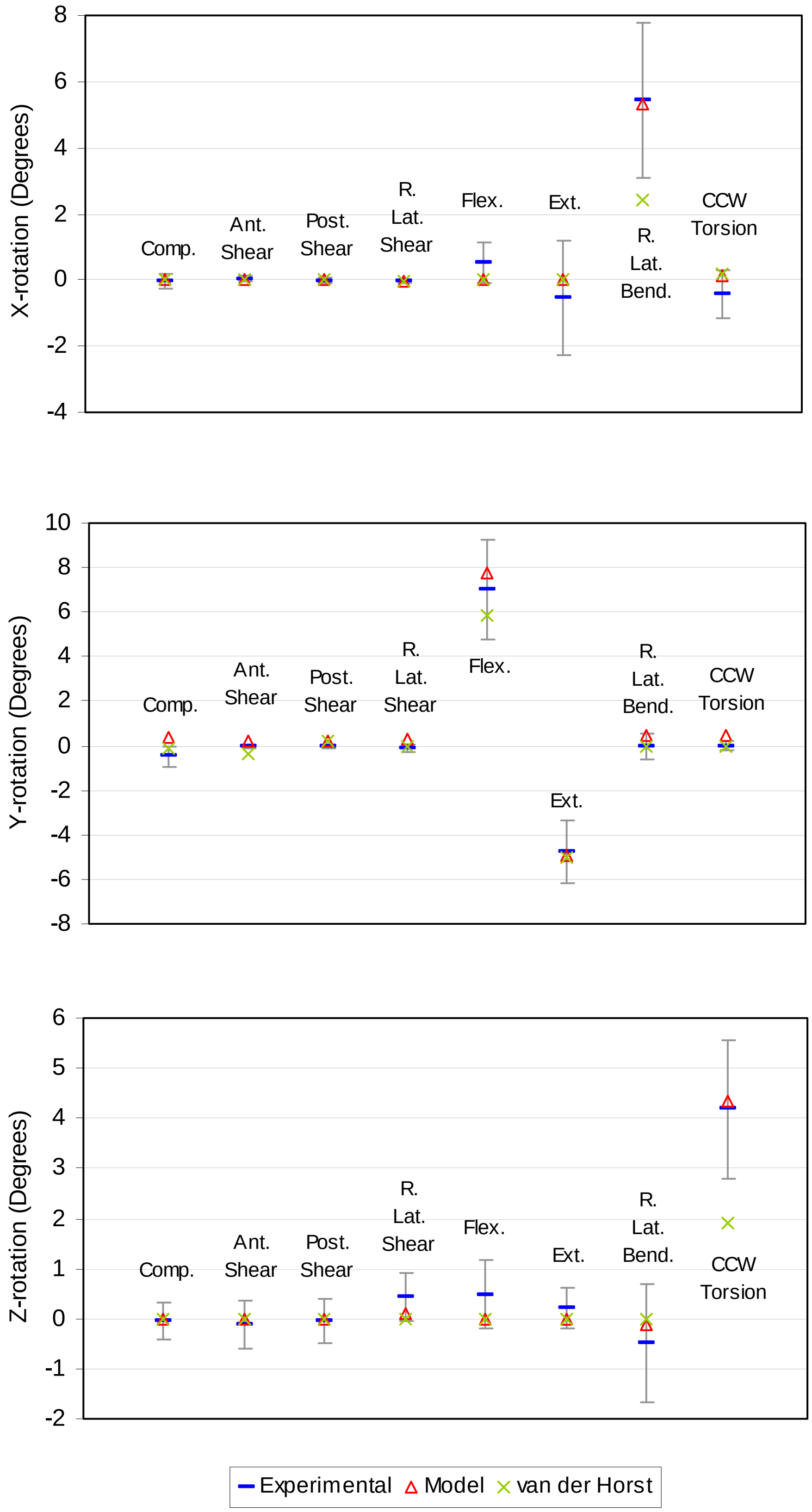
The simulated small load responses of the model, the displacement and rotation of C5 with respect to C6, are compared with the experimental results obtained by Moroney et al. (1988) for the disc alone (Figure 6.4 (a) and (b)). The figures illustrate the results of the Moroney et al. tests as vertical bars representing the spread of the standard deviation for each loading configuration. The horizontal bars show the experimental means while triangles (Δ) indicate the predicted model results. For whiplash motion the critical motions are a combination of shear displacement and rotation, which (in rear impacts) are displacement in the x direction and the rotation about the y-axis.

Figure 6.4 (below) Comparison of the (a) small load displacement, and (b) rotational responses of the new head and neck model, the original van der Horst model, and the experimental results obtained by Moroney et al. (1988) for the disc alone.

(a) C5/C6 disc only segment displacement.



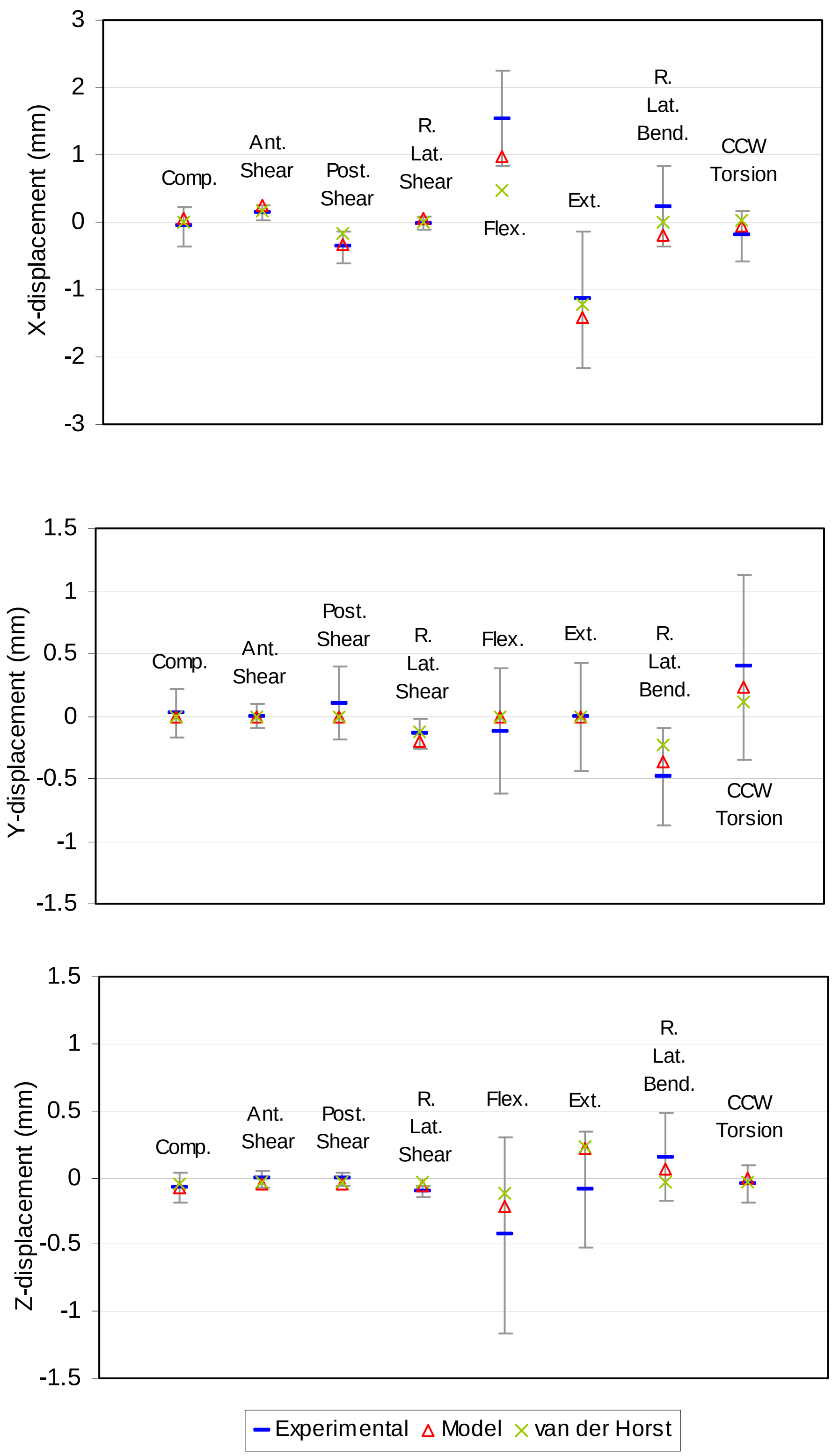
(b) C5/C6 disc only segment rotation.



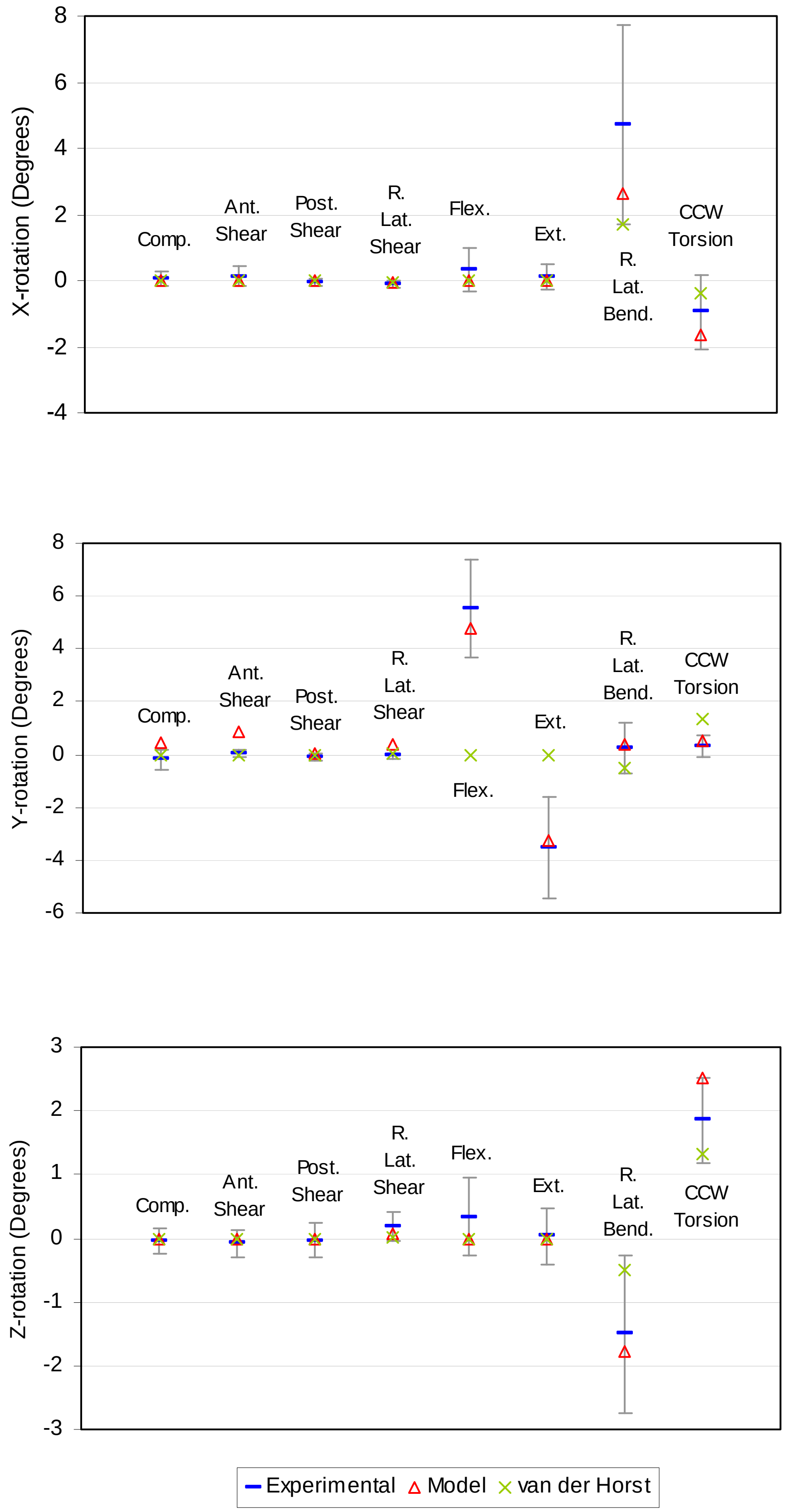
Overall, the responses of the disc model are generally within ± one standard deviation of the experimental results – especially with respect to the x-axial displacement and y-axial rotations. Particularly for the applied motion, the model does not fully reflect some of the coupled motions occurring in the real neck motion segments, such as the z-axis displacement motion. Similarly, the simulated small load responses of the model are compared with the experimental results obtained by Moroney et al. (1988) for the intact motion segments (Figure 6.5 (a) and (b)). Overall, the responses of the model for the intact motion segment are within ± one standard deviation of the experimental results for all directions of loading and this is an improvement on the original van der Horst (2002) motion segment.

Figure 6.5 (below) Comparison of the small load displacement (a) and rotational responses (b) of the new head and neck model, the original van der Horst model, and the experimental results obtained by Moroney et al. (1988) for intact motion segment.

(a) C5/C6 intact motion segment displacement.



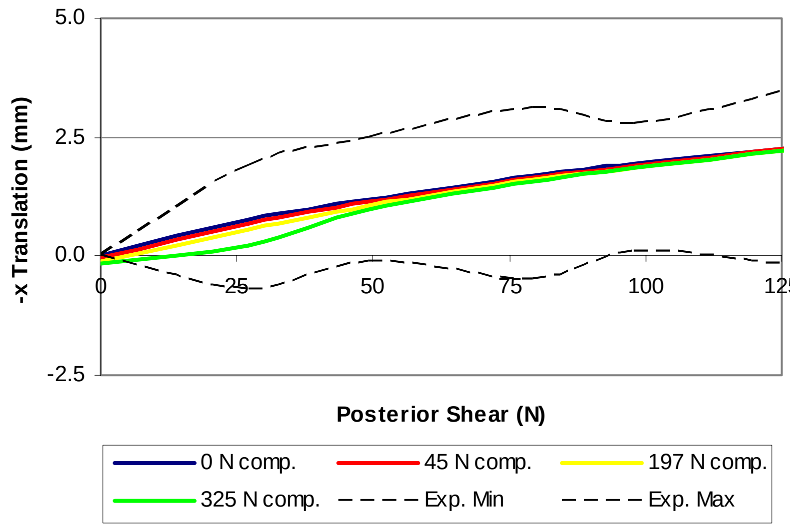
(b) C5/C6 intact motion segment rotation.



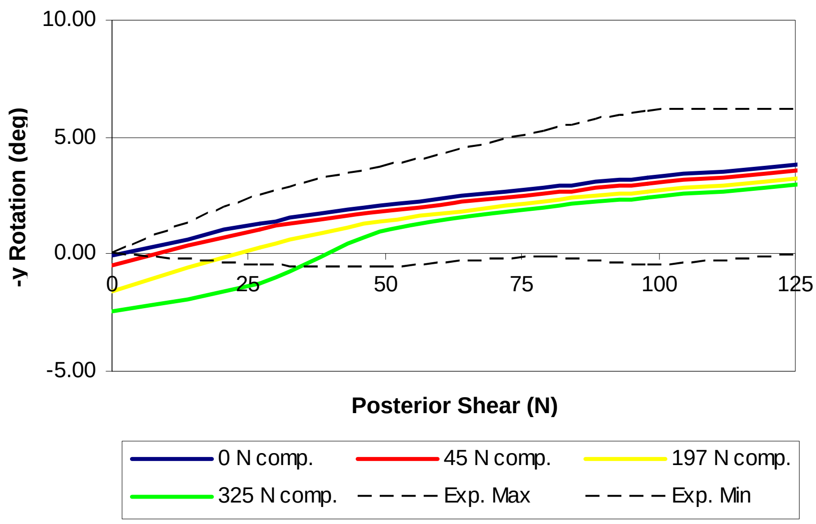
The model predictions for large loads were compared with the measurements taken in posterior shear of C3/C4 and C5/C6 motion segments (n = 13) under compression loading by Siegmund et al. (2000b) (Figure 6.6 (a) and (b)). The means and maxima and minima of the experimental results at each level of posterior shear and compression loading of the segment are shown along with the model predictions.

Figure 6.6 (below) Comparison of the model motion segment responses and posterior shear compression and loading experimental results (dotted corridors) obtained by Siegmund et al. (2000b) for excised motion segments.

(a) Displacement



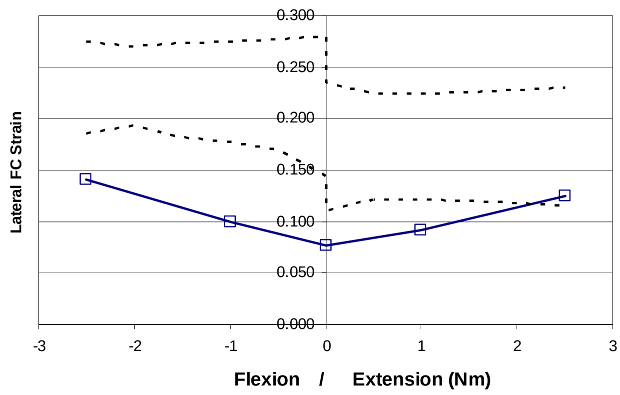
(b) Rotation



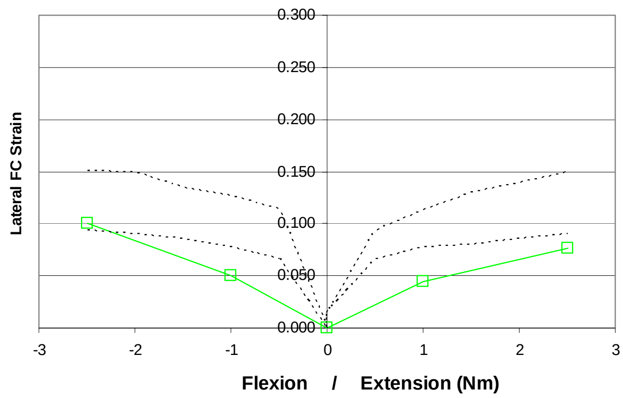
The model predictions were also compared to the related study by Winkelstein et al. (2000), in which the flexion and extension bending of excised human neck motion segments, while being loaded in axial torsion, were investigated. The results of this comparison are plotted in Figure 6.7 (a), (b) and (c) respectively.

Figure 6.7 (below) Comparison of the predicted model facet capsular lateral ligament strain with experimental principal strains (dotted corridors) from Winkelstein et al. (2000) for flexion and extension bending of excised human neck motion segments loaded in axial torsion.

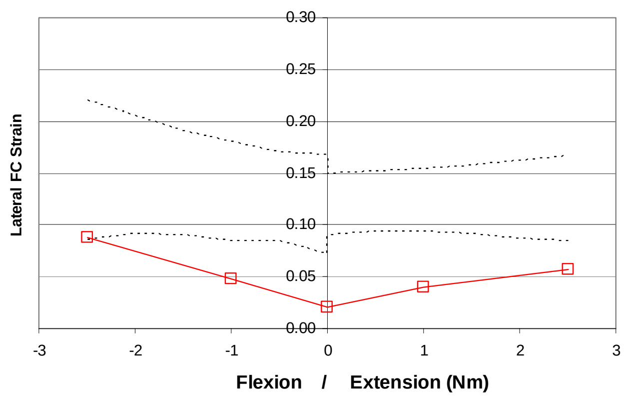
(a) +1 Nm axial torsion



(b) 0 Nm axial torsion



(c) -1 Nm axial torsion



The strains plotted are for the experimental results, the principal strains for the facet capsule calculated from the individual displacements measured during the test; and, for the model, the relative elongation for the lateral facet capsular ligaments.

6.4 Discussion

The simulated responses of the model were compared with the experimental results obtained by Moroney et al. (1988) for the intact motion segment in Figure 6.5. The neck responses important for WAD loading are translation in the x direction and rotation about the y-axis. The small load responses of the model motion segment fell within the spread of experimental results obtained by Moroney et al. for all of the loading directions. The motion segment responses in extension are 4 degrees rotation and 1.5 mm rear displacement for the intact motion segments. To place these responses in perspective, it is useful to compare these results with the C5/C6 rotations measured in the volunteer tests conducted by Ono et al. (1997) (see Chapter 3). In these dynamic rear impacts at 8 km/h, with a peak sled acceleration of approximately 4g, the measured C5/C6 rotation in extension was between 10 and 15 degrees. Testing of intact head-neck units by Pearson et al. (2004) produced a 3 mm sliding displacement of the C5/C6 facet capsule. The Moroney et al. results are well within the physiological limits for the motion segment.

For validation purposes, the model response predictions for larger combined loads were then compared with the results of two experiments involving loading at a level and type that was more consistent with that, which occurs in actual rear impact situations. These studies had been performed on excised motion segments and were aimed at investigating specific possible neck motion segment injury mechanisms associated with WAD.

The first set by Siegmund et al. (2000b), investigated the effect of posterior shear of a motion segment under compression loading on excised human necks (n = 13). This work was based on the neck loading observed for volunteers subjected to rear impacts (Ono et al. 1997). The experimental loading took the neck segments to the point of failure. The posterior translation and extension responses predicted by the model were compared with results from the experiments. The predicted translation and rotation of the upper vertebra (C5) rose to the upper edge of the experimental corridors. The principal shear strain on the facet capsular ligaments was measured during the experiments. It was found not to vary with the compression load and this is reflected in the model predictions. For the 137 N posterior shear cases, the researchers found that the principal facet capsular ligament strains varied between 11% and 23% for the motion segments measured, and were aligned with the direction of loading (x-axis). The model predicted a shear strain of 24% for the same load case, which is consistent with the slightly more flexible joint characteristics demonstrated in Figure 6.6.

The second study by Winkelstein et al. (2000), investigated the effect of flexion and extension bending on excised human neck motion segments (n = 12) while being loaded in axial torsion to the point of failure. This set of experiments was aimed at investigating the effect of having the head rotated at the onset of the loading in a rear impact, on strains within the facet capsule. This has been reported as a significant factor in several field accident studies of whiplash related accidents.

The model motion segment predictions are able to show the correct trends in the preloading effects on the ligaments, as a result of the axial rotation, but the magnitude of the strain appears low. The experimental corridors shown in Figure 6.7 are the principal strains in the facet capsule ligaments, calculated from the strains on the capsule measured in the experimental study. These experimental principal strains, in terms of both magnitude and position within the capsular ligaments, were found by the researchers to be quite variable. They suggest that this may be due to individual variations in the structure and geometry of the facet capsule. It was not explained by the researchers why the neck motion segment responses were asymmetrical in the two directions of rotation about the z-axis. These experimental results are therefore affected by the experimental error and possibly also by the difficulty of defining the neutral zone of the motion segment being tested. In Figure 6.7, these experimental principal strain corridors are compared with the maximum uniaxial strains in the lateral facet capsule ligament predicted by the model. The averaged model strains were lower than the measured principal strains.

A reason for the underestimation stems from one of the drawbacks of a multi-body model due to the representation of the ligaments as spring-dampers. These ligaments give relative elongation (and hence uniaxial strain of the ligament) based on the direct line of action between the two insertion points on the upper and lower vertebra (Figure 6.8). In reality, the ligament fibres are clumped together and interwoven in structures such as the anulus fibrosus or the facet joint capsule. In the facet capsule for example, the ligaments are perpendicular to the facet surfaces. If the superior and inferior vertebral bodies, which form the motion segment, move in shear, the ligament fibres are elongated in the 'S' shape shown in Figure 6.8. In the model, the fibres give a direct line between the insertion points, so the actual strain is higher at the centre of the real ligament. The prediction of both the line of action of the ligament under load and the magnitude of the predicted strain contain errors as a result. This would add to the rotational effect described earlier.

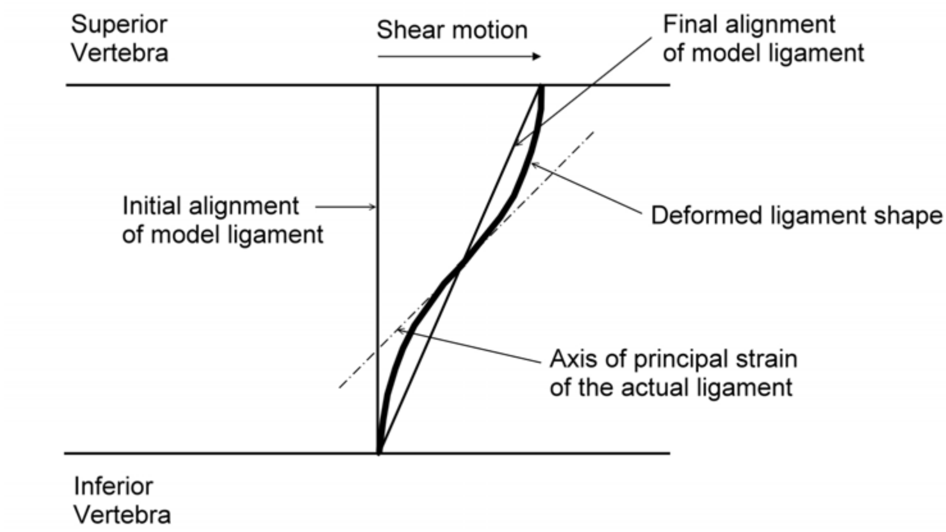


Figure 6.8 A comparison of the predicted uniaxial ligament strain of the model and the force line of action, with the deformed shape of an actual ligament for the facet capsular ligaments in shear.

For more complex combined loads, the model responses tended to underestimate the ligament strains in the facet capsule and did not lie within the corridors of the experimental results. The trends and magnitudes of the motion segment responses demonstrated are in reasonable agreement with the experimental values. Combined loading is difficult for a relatively simple multi-body model to accurately simulate, but acceptable results were achieved.

6.5 Summary

A mathematical multi-body model of a human C5/C6 motion segment has been developed. The aim of developing this model was to allow investigation of the causation of soft-tissue neck injury in rear impacts.

The model is based on the motion segment from the head and neck model developed by van der Horst et al. (2001). The new C5/C6 motion segment model includes the representation of the anatomical structures in the motion segment that improve the biofidelity of the model responses and allow the prediction of injury. The structure of the disc developed is based on recent developments in the anatomy described by Mercer and Bogduk (1999).

The small load responses of the C5/C6 motion segment model were developed to agree with the *in-vitro* experimental results obtained by Moroney et al. (1988) for all directions of direct loading, including the critical load directions for whiplash –negative x-axial displacement and negative y-axis rotation. The model responses are also in reasonable agreement for the coupled or indirect displacements and rotations.

The C5/C6 motion segment model responses were validated for large loads by comparison with further *in-vitro* tests to failure, on excised human motion segments. The combined loading tests simulated were representative of possible injury causing loads in real accidents and consisted of combined posterior shear and compression (Siegmund et al. 2000b), and axial torsion and flexion-extension bending (Winkelstein et al. 2000). The model motion segment results were in good agreement with the combined posterior shear and compression tests. The motion segment model demonstrated a high level of motion similitude in simpler combined load situations. In more complex combined loading situations it still maintains correct predictive trends, making the model suitable for the purpose intended in this study.

The next stage of the study is to include the C5/C6 motion segment model in a multi-body human neck model to allow its application to the study of whiplash-associated injury.

CHAPTER 7 THE HEAD AND NECK MODEL

7.1 Introduction

To use the C5/C6 motion segment model for the investigation of whiplash injury mechanisms, it was necessary to re-integrate it with a head and neck model to allow the application of realistic dynamic loading. Since a segment of the human head and neck model by van der Horst (2002) formed the basis for the motion segment, the integration of the enhanced motion segment was a relatively simple process. Validation of the dynamic responses of the new head and neck model to rear impact loading comprised several stages. The responses of the model and of the original van der Horst head and neck model were verified to have remained unchanged by comparing each of the model responses with the responses of volunteers in dynamic testing. The overall motion of the neck and the individual vertebrae, and the movement of the motion segment instantaneous axes of rotation (IARs) during the test were all assessed as part of the model validation process. The various stages in the head and neck model verification process are outlined in Figure 7.1.

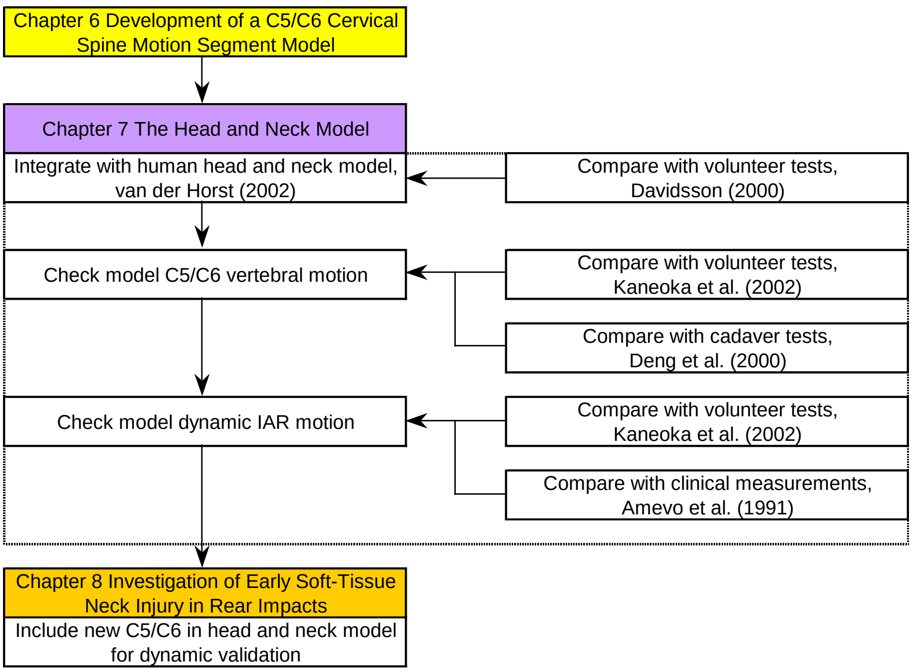


Figure 7.1 The human head and neck model integration and validation.

7.2 Method and Materials

The C5/C6 motion segment model was based on an existing, well-accepted multi-body neck model developed by van der Horst (2002). The motion segment was modified, as described in the previous chapter, Chapter 6, to improve the human dynamic similitude (biofidelity) as well as the capability of the joint soft-tissue components to sense injury to the disc, the neck ligaments and the facet capsule. To be able to apply realistic dynamic loading to the motion segment, the mathematical C5/C6 model was re-integrated into the van der Horst human head and neck model. The resulting model with the newly developed C5/C6 motion segment will be referred to hereafter as the head and neck model.

The original van der Horst (2002) head and neck model had been validated for rear impacts during its development. The testing used for this purpose, included cadaver testing by Bertholon et al. (2000) and volunteer testing by Davidsson (2000) and van der Kroonenberg et al. (1998). The model was found to give good to reasonable correlation with the test results for cadavers and volunteers in rigid seats (van der Horst 2000). The correlation was not as good in vehicle seats with headrests.

The van der Horst head and neck model was also validated by Stemper, Yoganandan and Pintar (2003b) for rear impacts, by comparison with testing of cadaver head and neck complexes. For this purpose up to 150 ms the model was found to give good correlation with the head and neck kinematics and the relative vertebral angular displacement.

Therefore the first stage was to verify that the dynamic responses of the new head and neck model were similar to those of the original van der Horst model under identical conditions. The test series chosen for this purpose was that reported by Davidsson (2000) and was selected for the following reasons:

- Ten instrumented volunteers were tested at $\Delta V = 9.3$ km/h;
- The rigid seat had the backrest at 20° and did not have a head restraint fitted;
- Comprehensive subject responses were available;
- Details of the vertebral motion were available from high-speed radiography of the head and neck motion of the volunteers during the tests.

The position of the head and neck model was reasonably well matched to the initial posture for volunteers in the Ono et al. (1997) series. The dimensions of the model neck were approximately those of a 50^{th}-percentile male (173 cm tall and weighing 80 kg) (van der Horst et al. 2000); similar in weight and stature to the volunteer (S6) – a 34 year-old male, 176 cm tall and weighing 72 kg.

Figure 7.2 shows the global T1 acceleration and angular rotations measured for the volunteer by Davidsson (2000), which were applied to the head and neck model at T1 (Figure 7.3). The head and neck model was run at the passive muscle setting and the dynamic response stiffness multiplier was selected. This increases the stiffness of the ligaments in the model by a factor of 2, to compensate for the velocity dependent characteristics exhibited by the ligaments, Yoganandan et al. (1998c).

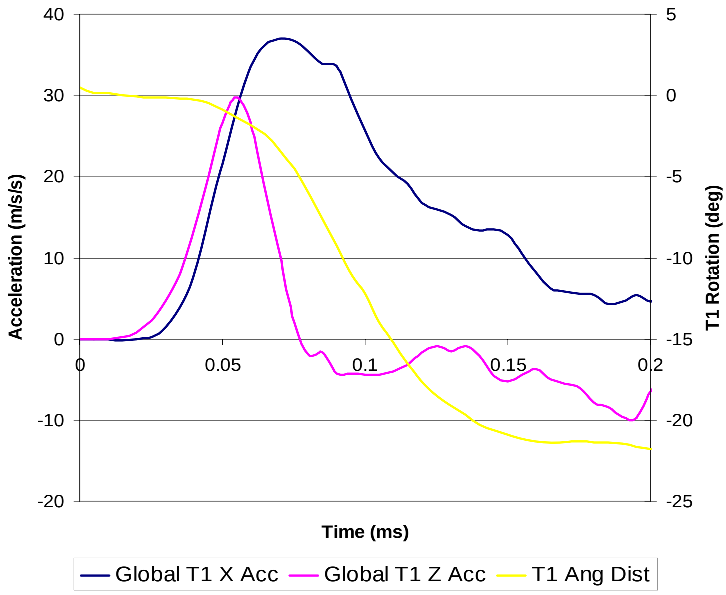


Figure 7.2 The volunteer global T1 x and z accelerations and angular rotation measured during the testing by Davidsson (2000).

For the purpose of verifying the head and neck model responses, the predicted responses due to rear impact loading were compared to the responses obtained in the JARI volunteer tests. The following parameters were included in the comparison:

- The neck link motion;
- The neck vertebral motion;
- The relative motion of the C5/C6 vertebra; and,

- The instantaneous axis of rotation (IAR).

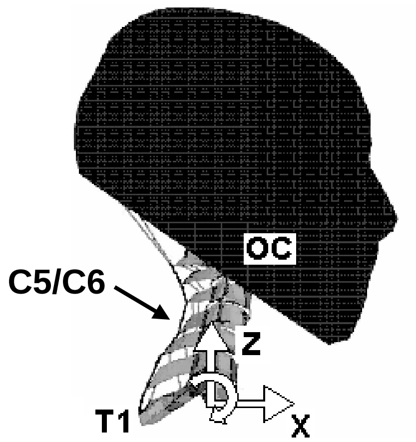


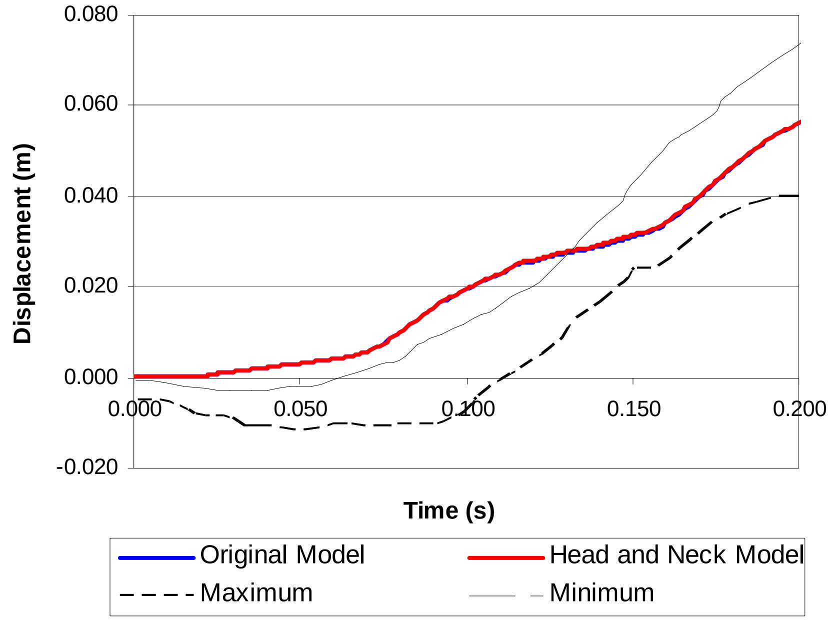
Figure 7.3 Volunteer global T1 x and z acceleration and angular rotation were applied to the head and neck model T1 motion segment.

7.3 Results

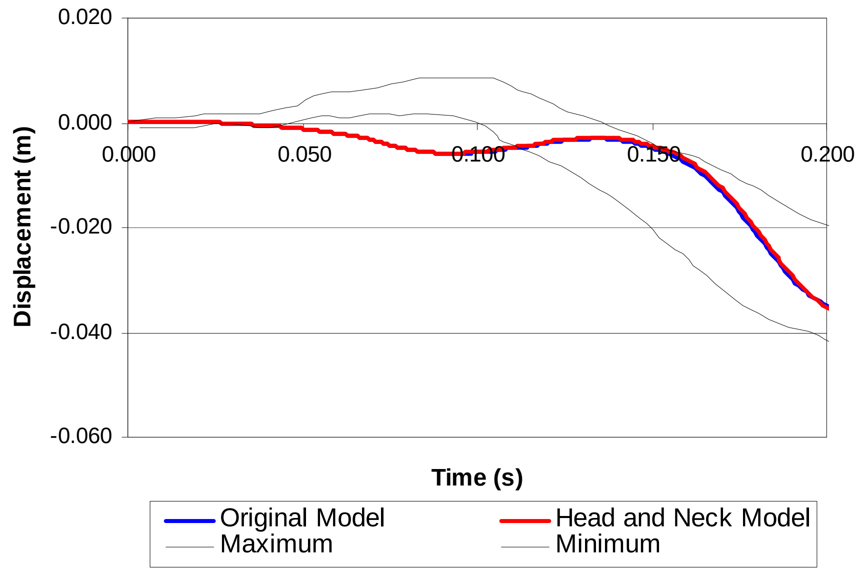
The first thoracic vertebra (T1) of the head and neck model was given the same motion as that measured during a volunteer test by Davidsson (2000). The volunteer corridors (maximum and minimum) of the resulting neck link motion are represented in Figure 7.4 along with the motion predicted by the head and neck model, and for comparison purposes the motion of the original van der Horst model under the same conditions. The head and neck model and the original van der Horst model are virtually identical in the predicted neck link responses. The mathematical models fail to predict the slight flexion motion of the neck that occurs in the initial 100 ms of the T1 motion. The effect of this motion is evident in the slightly flexed upper cervical spine in the high-speed radiograph taken at *t = 44 ms* for the volunteer in Figure 7.5. This motion results from the test methodology used for the volunteer tests, described in Section 3.7.2. The volunteers were placed in a seat that accelerated due to gravity down an inclined ramp. The rear impact occurred as the seat reached a buffer at the base of the ramp. The motion of a sled of this type causes the head of the subject to move slightly, to 'float' in its vertical position and move slightly into flexion. In these tests leading to an average displacement of the centre of gravity of the head approximately 6 mm forward in the and 2.5 mm upwards. These motions are not evident in the model, which has the measured local T1 accelerations from the volunteer test applied directly to the T1 motion segment. The snapshots of the head and neck motion predicted by the model are shown in Figure 7.5 along with the vertebral motion of the volunteer (S6) obtained from the high speed radiographs recorded during the test. It can be seen that the neck link moves from upright and neutral to full extension with similar timing.

Figure 7.4 (below) Neck link response predicted by the head and neck model and the original van der Horst model compared with the corridors (mean ± SD) from the volunteer sled test with a rigid seat (ΔV = 9.3 km/h) by Davidsson (2000).

(a) OC relative to T1 x-axial displacement



(b) OC relative to T1 z-axial displacement



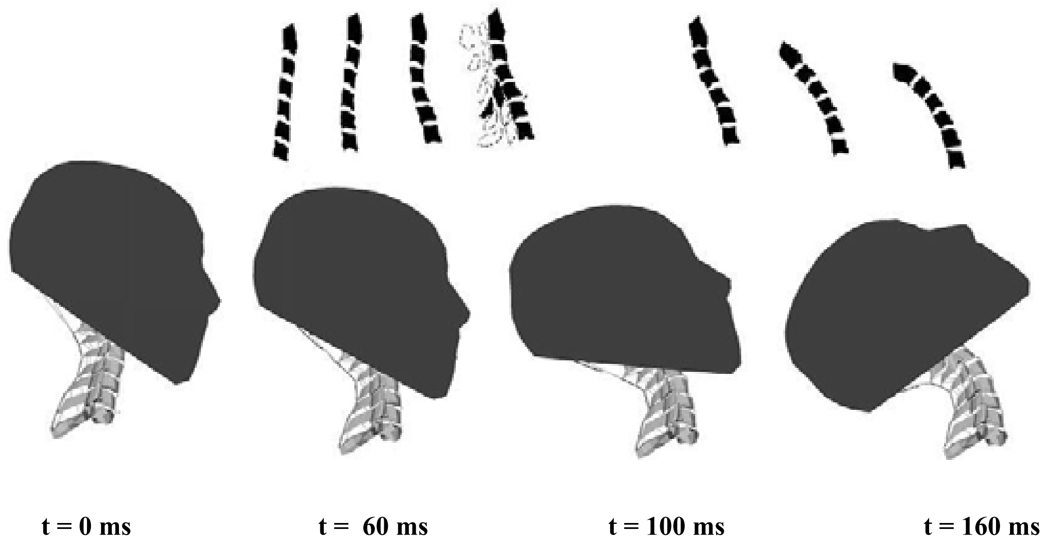
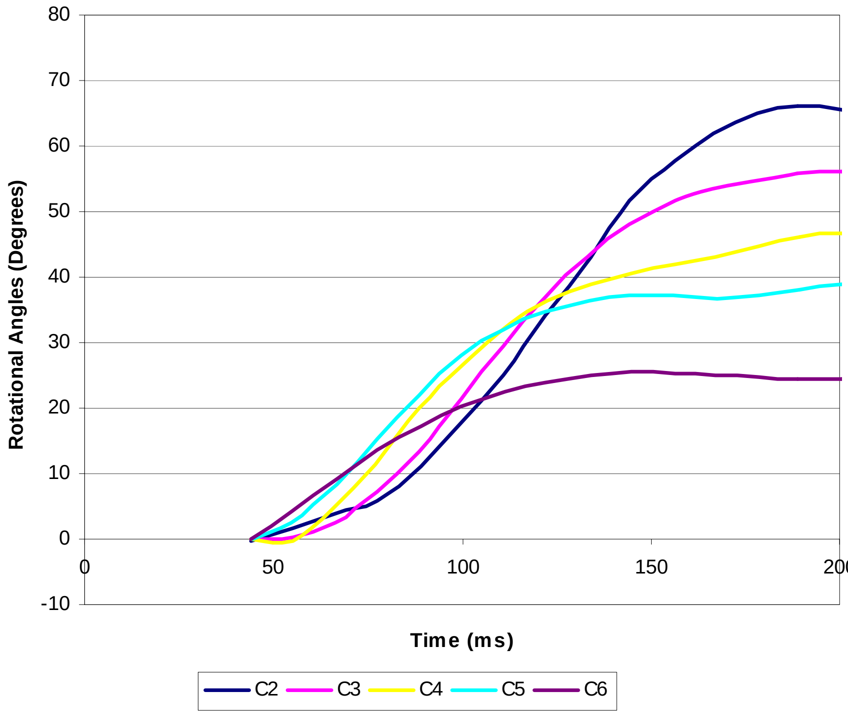


Figure 7.5 Time aligned motion of the neck of the volunteer (S6) from high-speed radiography (top row), and of the head and neck model (lower row) (up to t = 160 ms). The positions of the neck correspond with the phases of Ono et al. (1997).

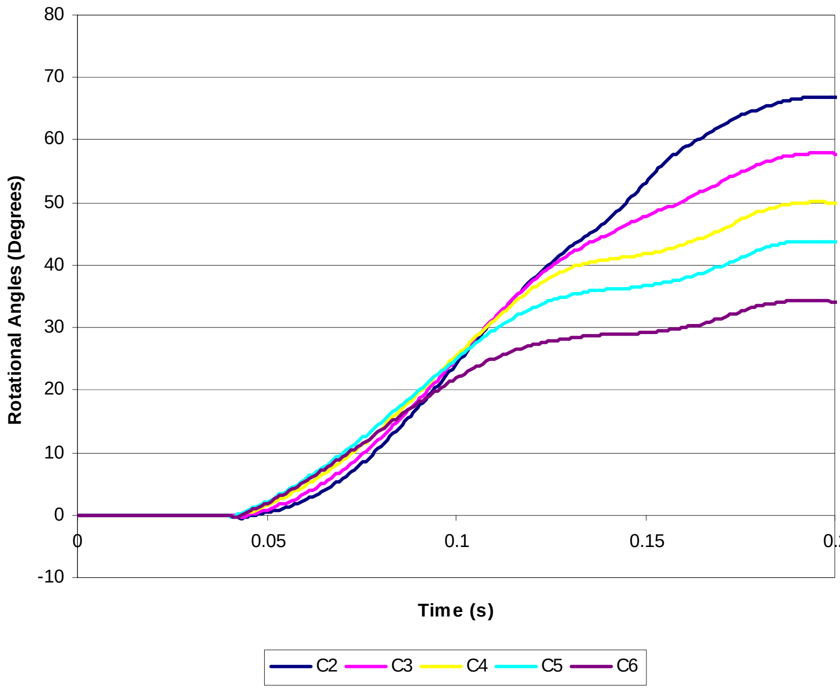
The predicted vertebral rotations for segments C2 to C6 with respect to the horizontal plane are compared with those of the volunteer (S6) from Kaneoka et al. (2002), for the same test series used by Davidsson (2000) in Figure 7.6. The volunteer S6 was used as the specific comparison for the head and neck model because more test response data was available in the published literature, including the high speed radiography. The model predictions of the vertebral rotation show good correspondence with those of the volunteer. Similarly, the relative vertebral rotations for C2/C3, C3/C4, C4/C5 and C5/C6, were predicted by the head and neck model and for all the test subjects (S3, S5, S6 S7 and S10) available in Kaneoka et al. (2002) were compared in Figure 7.7. The predictions by the head and neck model fall substantially within the subject corridors. For C5/C6, the relative rotation of the vertebra predicted by the model falls with in the volunteer motion during the test up until 125 ms, which is the period of maximum interest. At this point the motion diverts gradually below the relative vertebral rotation of the volunteers.

Figure 7.6 (below) The resulting vertebral rotations, in relation to the ground plane, for C2, C3, C4, C5 and C6 of the model are shown in comparison with the results from the volunteer test (S6) by Kaneoka et al. (2002).

(a) Subject S6



(b) Head and Neck Model



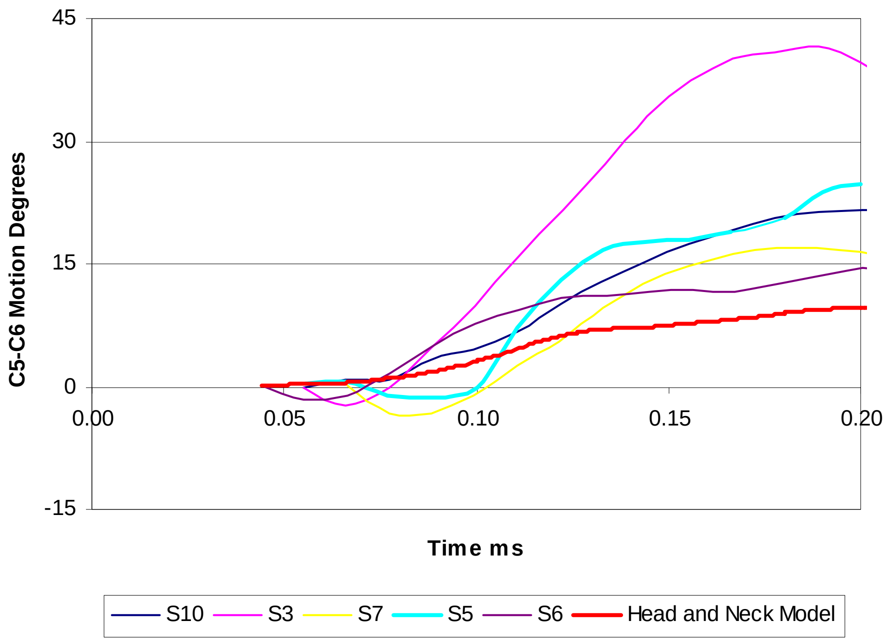


Figure 7.7 The relative vertebral rotations for the volunteers (S3, S5, S6, S7 and S10) tested by Kaneoka et al. (2002) with the response of the head and neck model for comparison.

For this relatively low-severity rear impact volunteer test with no pain outcome (or injury), the NIC_{max} was calculated to be 7.7 m/s at 70 ms, well below the level of 15 suggested for injury by Bostrom et al. (1996). The C5/C6 motion segment model predicts that for subject S6, C5 is displaced in the negative x direction (anterior/posterior shear) and rotates about the y-axis into extension. The resulting displacements of the bearing surfaces of the motion segment are given in Table 7.1. The displacements of the facet capsule (FC) in both the x and z directions reach a peak early in the motion, before 130 ms, justifying the range of interest.

Table 7.1 Table of peak predicted C5/C6 motion segment bearing surface displacements for the volunteer test, Davidsson (2000).

Component	Peak displacement (mm)	Time (ms)
FC		
X (front)	-1.03	100
Z (front)	1.33	130
AF		
X (front)	-1.14	100
Z (front)	2.70	160
NP		
X (back)	-1.17	100
Z (front)	2.04	160

The peak z-direction displacements for the anulus fibrosus (AF) and nucleus pulposus (NP) at 160 ms are due to the excessive extension motion of the head and neck model when run with passive muscles. The possibility of the facet surfaces impinging has been hypothesised between 50 ms and 75 ms, during the motion of the facet surfaces towards each other. It is this early motion that is of interest here.

The loading predicted by the model on the C5/C6 motion segment ligaments (ALL, PLL, LF, ISL and FC) for the volunteer test (S6) was investigated (Table 7.2). The relative elongations (strain) for these ligaments were calculated for comparison with the failure strains found by Yoganandan et al. (1998c). The posterior ligaments (LF and ISL) did not go into tension. The peak AF fibre elongations, which occurred in the lateral fibres, are also included in Table 7.2. The elongation at failure for the lower neck discs under tension comes from the five tests by Pintar et al. (1986) and was 10 mm (128% strain). This is based on an assumed disc height of 5.5 mm in the model and a 45° inclination of the AF fibres. The elongation at failure given for the major ligaments in Table 7.2 are all for catastrophic failure. Partial or sub-catastrophic failures were found to occur in the FC ligament at strains of 35% by Siegmund et al. (2000b). The S6 volunteer test was non injurious and the predicted values for the C5/C6 soft tissue components are in agreement with this result.

Table 7.2 Peak C5/C6 ligament elongations predicted by the head and neck model for volunteer (S6) during the test. The ligament elongation at failure data is from Yoganandan et al. (1998c) and the AF fibres from Pintar et al. (1986).

Component	Peak displacement (mm)	Time (ms)	Initial length (mm)	Predicted % elongation	% elongation at failure
ALL	3.01	160	18.3	16	30
PLL	0.70	130	17.9	3	18
FL	compression	-	8.5	-	88
ISL	compression	-	10.4	-	68
FC (front)	1.65	130	6.9	24	116
AF (lateral)	1.78	160	7.8	23	128

The head and neck model was used to predict C5/C6 dynamic instantaneous axis of rotation (IAR) for the volunteer test (S6). The IAR was calculated using the same technique described by Kaneoka et al. (2002) using the high-speed radiographs of the same volunteer test (S6). The technique followed a standard clinical approach (Amevo, Worth and Bogduk 1991) described in Chapter 3. The dynamic C5/C6 IAR predicted by the head and neck model for the volunteer test (S6), was very dependent on the loading on the neck. In a rear impact, the vertebrae of the neck move in a combination of anterior-posterior shear (-ve x-direction) and flexion-extension rotation (about the y-axis). The combination of these motions defines the IAR: the rotation moves the IAR high up near C5 and the shear translation shifts the IAR down, away from the motion segment centre. In a static measurement of IAR within the physiological limits of the joint, the shear motion component does not exist and its position is due to the combination of the flexion and extension motion capability of the individual. The dynamic IAR for the motion segment is dominated by the instantaneous shear motion of the neck segment, even for this low severity volunteer test.

7.4 Discussion

The overall motion predicted by the head and neck model is comparable to that of the volunteer (S6) as illustrated in Figure 7.5, where the phasing of the neck motion and the magnitude are similar. This comparison is taken a step further in Figure 7.4 where the neck link motion, OC relative to T1 in both the x and z directions, is compared with this same set of volunteer tests reported by Davidsson (2000). The original van der Horst model is shown with the motion of the head and neck model and there are only minor differences between the two.

The following discussion focuses on the performance of the head and neck model when simulating human motion, which is essentially the same as the original van der Horst model (2002). The International Standards Organisation, ISO (1997) has suggested a comprehensive approach for model validation. This method has been applied as the biofidelity rating system for a 50th percentile adult male surrogate (either dummy or mathematical model), for evaluating side impact crash occupant protection. A comparison is made between the model and the results of sets of tests covering a major body region, in this case the head and neck, which have been first normalised to the 50th percentile by dimensional analysis. The tests results are given a rating value of R, according to the following:

- R = 10 (excellent) if the response meets requirement;
- R = 5 (good) if the response is outside requirement but lies in one corridor width of the requirement;

- R = 0 (poor) if neither of the above is met.

In this case the head and neck model falls within one corridor width of the test requirements and therefore is rated as good.

The head and neck model does not exhibit the initial flexion of the upper neck apparent in the response of the volunteers between 0 and 75 ms. This flexion effect may be related to the compression in the neck due to the spine straightening caused by pressure from the seat back, for which the resulting T1 z-directional acceleration is shown in Figure 1.1Figure 7.2. This acceleration and the inertia of the head leads to compression loading on the neck: a 30 m/s^2 acceleration in the z direction, which for a head weighing 5 kg, will give an axial force in the neck of 150 N. The centre of gravity of the head is slightly in front of the OC. The set-up of these tests using acceleration down a 10° ramp may exaggerate this effect by allowing the head to float and elongating the neck as the volunteer accelerates down the ramp.

It is possible to modify the neck responses of the model by applying the active muscle loading capability of the head and neck model. The complexities inherent in trying to apply the multiple active muscles to control the dynamic head and neck motion of the volunteers was beyond the scope of the work presented here, but some comments are necessary to set the ground work for the next chapter. Most of the muscle system of the neck is arranged axially; therefore as the muscles tense they compress the neck (Siegmund et al. 2000b). Further, Ono et al. (1997) found that when the volunteers were instructed to tense pre-test, the resulting rotation of the head was reduced in magnitude and its onset was earlier.

For the head and neck model without active muscles, the neck link motion continued extending past a reasonable physiological position by about 200 ms (Figure 7.4). For the head, the excessive extension angle became apparent earlier by about 160 ms (Figure 7.5). These excessive rotations of the vertebra and head made the predicted motion segment ligament loadings likely to be unrealistic after about 160 ms into the test. For this reason, this timing is taken as the limit of realistic head motion for the head and neck model for use in this study.

Ono et al. (1997) tested subjects with the neck in different initial positions: extended, neutral and flexed. As the neck initial position moved to greater flexion, the vertebral rotations increased. Accordingly, the volunteer (S6) in this test appears to have the neck in a neutral position for the test. The head and neck model initial position was aligned with that of the S6 volunteer neck link. By the time the first radiograph of the neck vertebrae of the volunteer was taken (Figure 7.3), the vertebra formed an almost straight line with no lordosis. This loss of lordosis is a sign of the slight flexion of the upper spine by the onset of the impact, due to the motion down the inclined ramp. The model neck is representative of a typical human and exhibits more lordosis at the time of the onset of the impact, and hence the upper neck model was more extended at onset. Hence the straighter profile and more closely grouped individual vertebral rotations of the model in Figure 7.6(b) compared with that of the volunteer S6 in Figure 7.6(a).

The C5/C6 motion segment model did predict that impingement of the facet surfaces was possible in these volunteer tests (Kaneoka et al. 2002). The model also predicted a significant amount of shear motion at C5/C6 (Table 7.2), which was not evident in the volunteer test (see Figure 3.25). The intact cadaver head and neck tests by Pearson et al. (2004) (see Chapter 3) showed motion of the C5/C6 segment, which was very similar to that in the model. For a peak acceleration of 6.5g, Pearson et al. (2004) measured a peak facet capsule compression of 1.8 mm and a peak facet capsule sliding of 4.0 mm at C5/C6 with an average facet capsule ligament strain of 35.9%. The motion of the FC described by Pearson et al. (2004) for the intact cadaver head and neck, in both rotation and shear, where the posterior edge of the C6 facet surface is impinged by the C5 surface, agrees with the model predictions. This motion varies from that found by Ono et al. (1997) for a volunteer, where the impingement involves the posterior edge of C5 encroaching the C6 surface, due to the rotation of the facet capsule with no shear displacement (discussed in Chapter 3). This variation may well be due to some effect of the active muscle responses in the volunteer.

The dynamic IAR were difficult to interpret as they reflect the instantaneous response of the motion segment to the applied load. The motion is within the neutral zone of the motion segment and only approaches the physiological limits in severe impacts. The calculated dynamic IAR for the model appears to be dominated by a higher posterior shear component than that experienced in the volunteer test used by Kaneoka et al.

(2002) to demonstrate the facet surface impingement hypothesis. The results obtained by Kaneoka et al. (2002), indicated that at the C5/C6 level, the major motion of the volunteers neck in the test was in rotation rather than shear.

To clarify, comparison with the results of another test program using high-speed radiography was made. In this test series, 26 low speed, rear-end impacts were conducted on six human cadavers by Deng et al. (2000a). It is important to note the multiple testing on each specimen. The peak sled acceleration for test number HFH20 was 59.4 m/s^2 and the resulting motion of the C2/C3 and C5/C6 motion segments are shown in Figure 7.8. The impact began at t = 90 ms. No visible neck injury was found on the specimen after the experiment. As can be seen from the figure, the peak C5/C6 facet displacement in the shear component (x direction) was measured as approximately 4.25 mm at 135 ms after impact. The peak shear of the C2/C3 segment was only 1.8 mm with similar timing. This indicates that significant shear occurred in the neck of the specimen, and that it was higher at the C5/C6 level of the neck than at C2/C3. This is counter to the facet impingement hypothesis of Kaneoka et al. (2002), which suggested that the motion of the C5/C6 segment was mainly rotation (see Chapter 3).

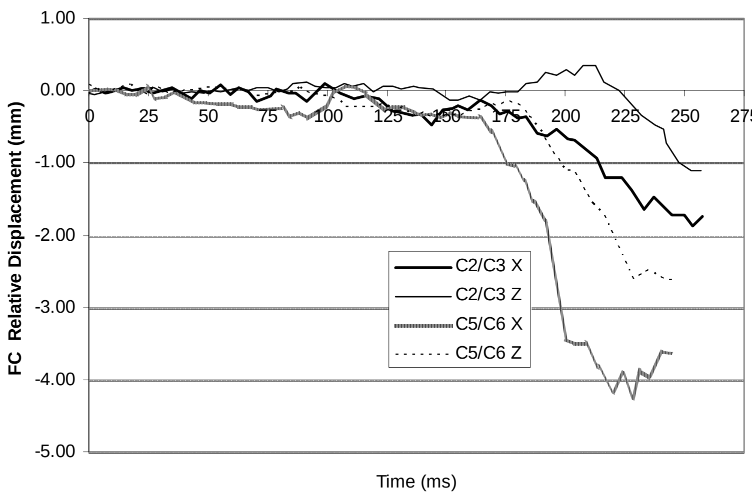


Figure 7.8 The relative displacement of the C2/C3 and C5/C6 motion segments in a 59.4m/s^2 rear impact test on a cadaver, Deng et al. (2000a).

7.5 Summary

To use the C5/C6 motion segment model for the investigation of whiplash injury mechanisms, it was necessary to re-integrate it with a head and neck model to allow the application of realistic dynamic loading. A segment of the human head and neck model by van der Horst (2002) formed the basis for the motion segment, and the integration of the enhanced motion segment into the head and neck model was a relatively simple process. Validation of the dynamic responses of this new head and neck model to rear impact loading required several stages:

- The overall motion predicted by the head and neck model was similar to that of the Ono et al. (1997) volunteer test, with similar phasing and magnitude of the neck link motion. The motion of the head and neck model until after 160 ms is consistent with the volunteer tests;
- The overall motion of the head and neck model was identical to the van der Horst model;
- When the head and neck model was run with passive muscle settings, the prediction of the vertebral motion was similar to that from high-speed radiography of the volunteer tests reported in Kaneoka et al. (2002), in terms of C5/C6 motion segment motion;
- When the head and neck model was run with passive muscle settings, the predictions of the relative vertebral motion at the C5/C6 level was within the response corridors obtained from high-speed radiography of the volunteer tests reported in Kaneoka et al. (2002), up to 125 ms. At this point it deviated slightly from the lower margin of the corridor in these non injury producing tests on volunteers. This extra stiff response of the model is appropriate for use in modelling injury; and,
- The investigation of the predicted movement of the motion segment instantaneous axes of rotation (IAR) during the test indicates that this is not suitable for dynamic validation of a neck model.

The head and neck model has been shown to have an adequate level of biofidelity in this Chapter by comparison with volunteer and cadaver tests and to be suitable for its application in Chapter 8. The van der Horst head and neck model was used as a means of applying realistic impact loads to the new motion segment. The limitations are associated particularly with the need for a better means of applying active muscle forces. This is outside the scope of this study, which is aimed at developing the motion segment capabilities. The final stage of the work in this study is described in Chapter 8, where the head and neck model was applied to the investigation of early soft-tissue neck injury in a sample of real crashes.

CHAPTER 8 INVESTIGATION OF EARLY SOFT-TISSUE NECK INJURY IN REAR IMPACTS

8.1 Introduction

The detailed C5/C6 motion segment model is applied to the investigation of longer-term pain outcomes in rear impacts. To do this the model is used to examine a group of real crashes by linking the results of crash reconstructions with the clinical pain outcomes of the vehicle occupants, over a 12-month period. The model responses are then used to investigate possible injury mechanism, which have been hypothesised by Barnsley, Lord and Bogduk (1998) to have pain outcomes.

The C5/C6 motion segment model was incorporated into the MADYMO-based human head and neck model developed by van der Horst et al. (2001). The head and neck model was validated in the previous Chapter by comparison with volunteer tests, using high-speed radiography of the neck motion by Kaneoka et al. (2002). This was achieved by driving the head and neck model by the T1 accelerations measured on the volunteers.

The T1 accelerations, used to drive the head and neck model, were derived from the reconstructions of real-life crashes using a MADYMO BioRID II dummy and seat model (Kullgren et al. 2003). The vehicles were part of a Swedish study, which used crash recorders to obtain the actual crash pulses of the vehicles. Whiplash associated outcomes to the vehicle occupants of the crashed vehicles was available in the form pain duration following the impact. This allowed the comparison of early C5/C6 motion segment component strains resulting from the crashes to be correlated with real-life AIS1 neck injury to the vehicle occupants.

The main aim in this section of the study was to demonstrate the use and effectiveness of the C5/C6 motion segment model. The complexity of achieving this, and the limited resources available for its implementation, made it necessary for four simplifying assumptions to be made.

- The first assumption was that the C5/C6 level of the neck was a significant source of injury in rear impacts. This is supported by accident data (Gibson et al. 2000), clinical investigation (Aprill & Bogduk 1992) and volunteer and cadaver test data (Kaneoka & Ono 1998; Deng et al. 2000b), and is discussed in Chapters 2 and 3.

- Second, it was assumed that a major injury-causing event in a rear impact was due to the head retraction motion (the 'S' shape) resulting from a combination of the head's inertia and the initial forward motion of the thorax at the onset of the crash. This early neck motion takes place before the head contacts the head restraint. This assumption is based on the timing of the phases of head motion from cadaver and volunteer studies (Deng et al. 2000b; Kaneoka & Ono 1998) and recent modelling studies (Stemper, Pintar & Yoganandan 2005). These are discussed in Chapters 3 and 7.

- Third, it was accepted that the crash data collected in the Swedish study was suitable for the purpose. The data, which forms the basis for this chapter, has been analysed in a number other published studies discussed in Chapter 2 (Krafft 1998; Krafft et al. 2000; Kullgren et al. 2003; and, Eriksson & Kullgren 2003). It is the only data available which has a direct measurement of in-service vehicle crash pulses in actual rear impacts. The vehicle models used are small to medium in size, have distinctly different seat designs, which have dynamic stiffness typical of the class. The pain outcomes to the occupants have been followed for a period of 6 months post impact. Finally the numbers in the study are of sufficient magnitude to allow statistical analysis of the results. Further investigation of this mass data source was beyond the scope of this project.

- Fourth, it was assumed that there was a direct link between the persistence of pain following a rear impact and the type of injury to soft tissues of the neck. This hypothesis has been supported by the comparison of animal and clinical studies regarding the time dependent progression of symptoms (Spitzer et al. 1995 & Radanov, Sturzenegger & Di Stefano 1995); by clinical studies of location of facet joint related pain (Barnsley et al. 1995); studies of the persistence of pain (Cavanaugh 2000); studies of neck pain based on volunteers in rear impacts (Szabo 1997); and actual injury to the neck in rear impacts based on cadaver test studies (Stemper, Pintar & Yoganandan 2004; Pearson et al. 2004; and, Deng et al. 2000a) and autopsy studies of neck injury (Taylor & Taylor 1996).

The study was pursued as it has merit, even with the limitations, and provides a useful step along the path that future research will adopt to develop to validate better human-body models and to define real injury-based criteria.

8.2 Methods and Materials

8.2.1 Introduction

A flow chart of the study is shown in Figure 8.1 and a brief description of each of these sections follows. The MADYMO based mathematical models used at each stage of the study were accepted models, each of which had been individually validated during development by comparing the model responses with appropriate test data.

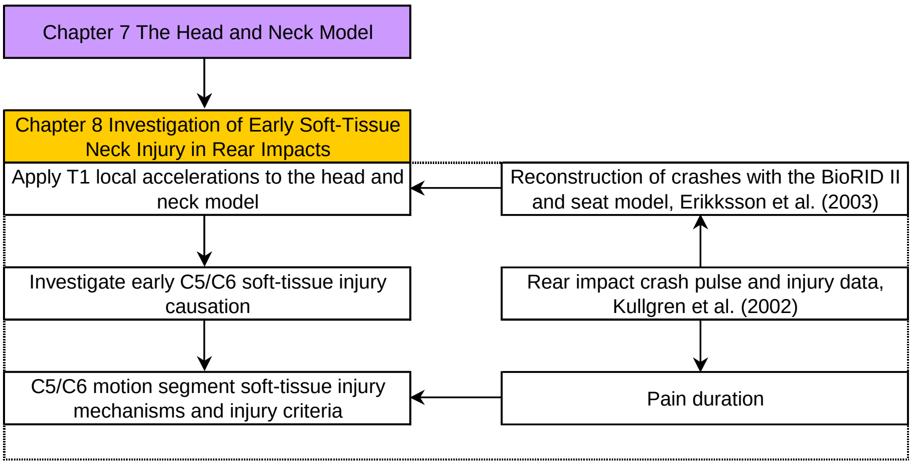


Figure 8.1 Diagram showing the linkage between the various sections of the study.

8.2.2 Rear Impact Crash Data

In the years following 1996, the Folksam Insurance Company, Sweden fitted more than 40,000 cars with crash-pulse recorders aimed at measuring acceleration-time history in rear-end impacts (Kullgren et al. 2003). These vehicles consisted of 7 models of the same make. All crashes involving these cars, irrespective of repair cost and injury outcome, have been reported. In this study, the three most highly represented car models were selected for inclusion if the vehicle was involved in a single rear-end crash with a recorded crash pulse and with front seat occupants who had no previous history of long-term AIS1 neck injury. These criteria gave a group of 79 crashes with 110 front-seat occupants for analysis in this study. The average age of the occupants was 45 years and the gender distribution was 47% male and 53% female.

The crash-pulse recorder records the acceleration-time history with a sampling frequency of 1000 Hz in the impact phase of a crash. Acceleration was measured in the principle direction of force within ±30 degrees. Crash pulses were filtered at approximately 60 Hz. Change of velocity, mean, and peak accelerations were calculated from the recorded crash pulses. Mean acceleration was calculated during the main part of the pulse until the acceleration approached zero. The threshold of the recorder is approximately 3 g.

Whiplash injury to the vehicle occupants was reported as whiplash symptoms and their duration. Examples of the whiplash symptoms reported are neck pain, headache, dizziness, and neck stiffness. The occupant injury status was divided into four categories according to the duration of symptoms: no symptoms; symptoms initially; symptoms remaining for more than one month; and, symptoms remaining for more than six months. Injury status was established from telephone interviews. For those occupants reporting a whiplash injury, follow-ups of medical symptoms were made on several occasions, with at least one after 6 months. The numbers of occupants in the three car models are presented in Table 8.1 with the various injuries.

Table 8.1 The number of occupants with their seating position and duration of symptoms in the three car models used in the study, from the Swedish crash data (Kullgren et al 2003).

Symptoms	Total			Car Model 1		Car Model 2		Car Model 3	
	Total	D	FSP	D	FSP	D	FSP	D	FSP
None	68	49	19	15	4	20	11	14	4
Initial symptoms	28	19	9	2	3	13	6	4	0
Symptoms > 1 month	7	6	1	2	0	2	1	2	0
Symptoms > 6 months	7	4	3	2	1	1	1	1	1
Total	110	78	32	21	8	36	19	21	5

D = Driver, FSP = Front Seat Passenger

In the crash data used here, the average velocity change of the vehicle was 10.0 km/h and the average mean acceleration was 3.5 g. The maximum change in velocity was 33.2 km/h and the maximum mean acceleration was 10.2 g. The data from this study has been analysed elsewhere by Eriksson and Kullgren (2003) and Kullgren et al. (2003).

8.2.3 Crash Reconstruction with the BioRID II and Seat Model

The seats in the three car models selected for further reconstruction differed in geometry and stiffness characteristics (Kullgren et al. 2003). The geometries of the seats were measured and the cushion, seat back, and head restraint contours were implemented in MADYMO in order to achieve correct contact areas between the seats and the dummy. Additionally, parts of the seat structures that may influence the dummy kinematics during the crashes were implemented. The BioRID II simulated was the first version, which was manufactured by Chalmers University and Autoliv Sweden, not by Denton.

For each seat model two sled tests with the BioRID II were carried out at ΔV = 23 km/h and mean acceleration of 4.5 g. These crash parameters were selected to be at the high end of the crash data. The BioRID II was seated in a normal posture, with no seatbelt used. The initial seat back inclination in the mechanical tests corresponded to a torso angled at 25° on an H-point mannequin and the head restraints were placed in their lowest positions.

The spread in the dummy responses within similar seats were used to establish corridors (Figure 8.2), for the following:

- The x and z accelerations in the dummy head, C4, T1, T8, L1 and pelvis;
- The y-rotations of the dummy head, T1, and pelvis;
- The seat inclinations and deformations; and,
- The dummy upper neck loads.

The stiffness characteristics of the MADYMO seat models were then tuned with the aim of fitting the responses of the MADYMO models into the sled test response corridors. A summary of the model responses (Kullgren et al. 2003), compared with the corridors obtained from BioRID II dummy sled tests in seats from the three car models, is presented in Figure 8.2. Of particular interest for the purposes of this study is the matching of the T1 responses from the BioRID II dummy and seat model with the dummy sled test results. The BioRID II dummy was designed for human like responses under such test conditions (Davidsson 1999a).

All of the selected crashes were reconstructed by exposing the MADYMO seat models and a MADYMO model of the BioRID II to the recorded crash pulses. The MADYMO BioRID II was placed in a normal seated posture, unrestrained, for the simulations and the head restraints were placed in the lowest position (worst case), as the position of the head restraint in the crash was unknown. The BioRID II dummy is a 50^{th}-percentile male in stature.

8.2.4 The Human Head and Neck Model

The development and validation of the human head and neck model used is described in Chapter 7. In brief, the original head and neck model developed by van der Horst (2002) was modified with the addition of a C5/C6 motion segment with improved anatomic representation, to allow the investigation of soft-tissue injury. The development of the C5/C6 motion segment model is described in Chapter 6. The head and neck model is a detailed multi-body model of a fiftieth-percentile human head and neck consisting of: a rigid head; rigid vertebrae (C1, C2, C3, C4, C5, C6, C7 and T1); non-linear, visco-elastic discs; frictionless facet joints; non-linear, visco-elastic ligaments; and, controllable, segmented contractile muscles (van der Horst 2002). The model vertebra shapes are based on the scanned vertebra of an individual's neck, but are represented as lumped masses. The muscles follow the curvature of the neck, with realistic lines of action for the muscle forces, and can be actuated as active muscles. The head and neck model has been validated quasi-statically with respect to *in-vitro* test data as well as dynamically using volunteer test data.

For simplicity, the head and neck model was kept in its standard posture, of an erect, standing volunteer (van der Horst 2002). Only the passive responses of the active muscle capability of the model were used. The accident reconstructions made by Kullgren et al. (2003), included the effect of the standard head restraint fitted to the seat in the BioRID II and seat model, but a head restraint was not included in the present head and neck model simulations. To realistically model the influence of the head restraint was beyond the scope of this study, which was aimed at developing and validating the C5/C6 motion segment model. These limitations in the simulation restricted the applicable time for the simulation of the internal C5/C6 motion segment loads within the neck to the first 150 ms of a typical rear impact event. Injury hypotheses based on volunteer testing (Ono et al. 1997; Deng et al. 2000b) and cadaver testing (Yoganandan et al. 1998a; Pierson et al. 2004; and, Stemper, Pintar & Yoganandan 2004) have emphasised the early onset injury mechanism.

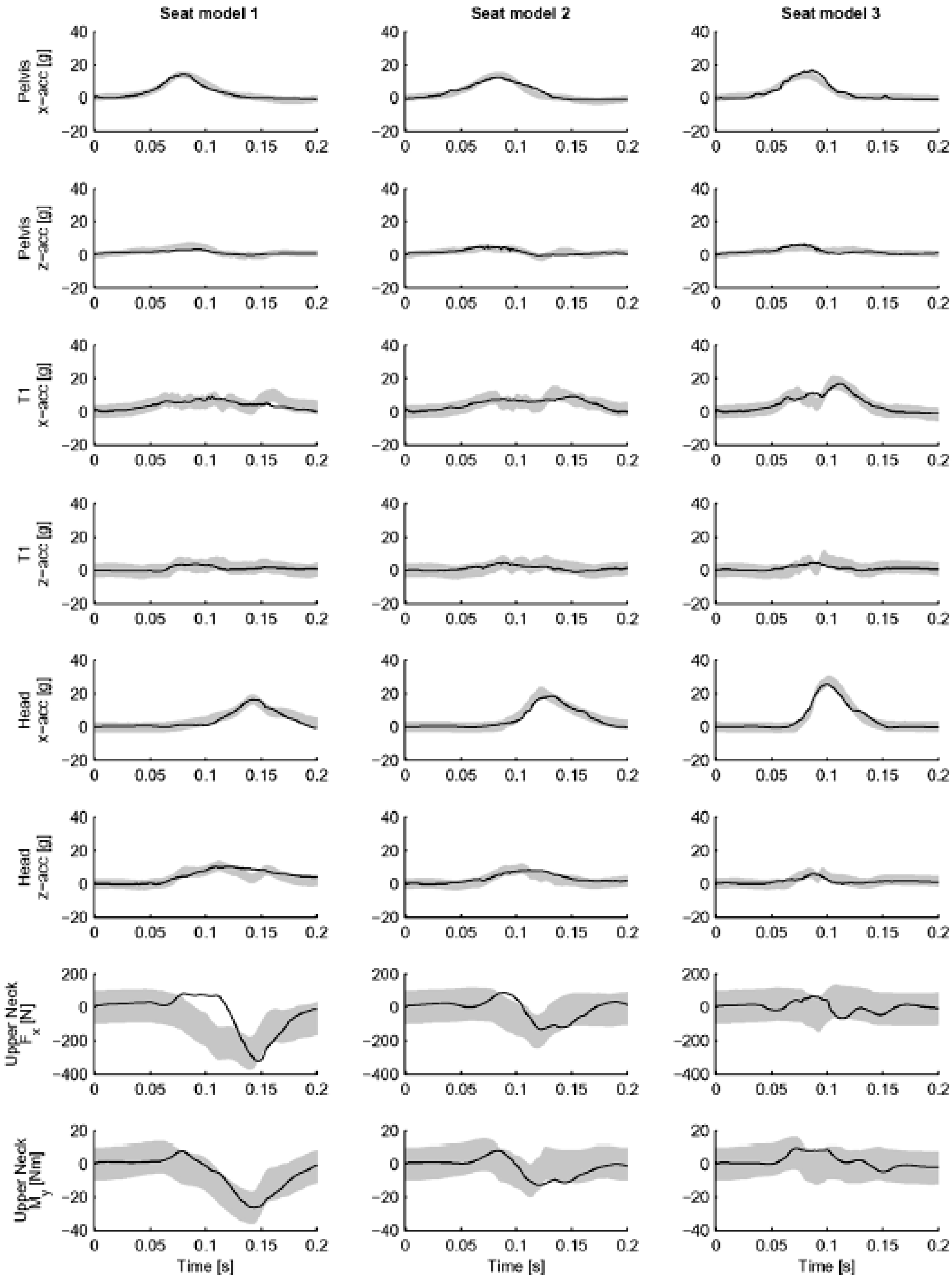


Figure 8.2 A summary of the BioRID II dummy and seat model responses compared to the corridors obtained from the results of sled tests with the BioRID II dummy in the seats for the three car models, from Kullgren et al. (2003).

8.2.5 The Study

The local T1 accelerations predicted by the BioRID II dummy and seat model were supplied for the 78 crash reconstructions of the Swedish data by Kullgren et al. (2003).

The local T1 accelerations from Kullgren et al. (2003) were used to drive the head and neck model by the T1 segment. Each crash scenario was run only once, even if there were multiple occupants in the vehicle: thus 78 runs were made. Where a front-seat passenger was in the crash, the simulation of the driver (n = 78) was the same as that of the non-driver (n = 32). The whiplash associated pain outcome, which usually differed for the two occupants, was included in the later analysis.

The NIC_{max} was calculated for each case for comparing the similitude of the BioSIDII seat and dummy model (Kullgren et al 2003) and the human head and neck model. It was outside the scope of the study presented here to make a general assessment of the various soft tissue neck injury criteria. NIC_{max} was chosen as it is a measure of the retraction motion of the neck, which leads to the whiplash associated 'S' of the neck, see Section 3.11.2. At the time of the modelling work for this thesis, it was the only one of these criteria, which could claim validation, see discussion in Chapter 3. Further, NIC_{max} was formulated to work in this early onset neck injury area (Bostrom et al. 1996). An alternative soft tissue neck injury criterion, which gives acceptable results, N_{km} was also investigated. This criterion is applicable through the entire whiplash event, but the lack of similitude of the model extension motion later in the impact event required selective interpretation of the results and this was thought to be inappropriate.

The motions and forces within the C5/C6 motion segment resulting from the T1 accelerations were analysed and compared to the pain outcomes for the vehicle occupants. The pain duration following a rear impact was split into four groups dependent on likely injury type: Injury Group 0 (no pain); Injury Group 1 (pain persisted for one week); Injury Group 2 (pain persisted for one month); and, Injury Group 3 (pain for six months). The possible injury mechanisms, leading to the longer term pain outcomes investigated were based on the WAD injuries discussed in Chapter 3. The components of the C5/C6 motion segment selected for further investigation are summarised in Table 8.2 (the axes are indicated in Figure 8.3).

Table 8.2 The components of the C5/C6 motion segment investigated for possible early whiplash associated injury.

Symbol	Component	Units
ALL	Anterior longitudinal ligament	Strain, %
FC	Facet capsule ligament	Strain, %
AF	Anulus fibrosus ligament fibre	Strain, %
BFz	Back of facet surface	Force, N
Fx	Facet capsule	Motion in x direction, m
Fz	Facet capsule	Motion in z direction, m
AFx	Anulus fibrosus	Motion in x direction, m
AFz	Anulus fibrosus	Motion in z direction, m
NPx	Nucleus pulposus	Motion in x direction, m
NPz	Nucleus pulposus	Motion in z direction, m
FPP	Facet surface	Displacement from median, m

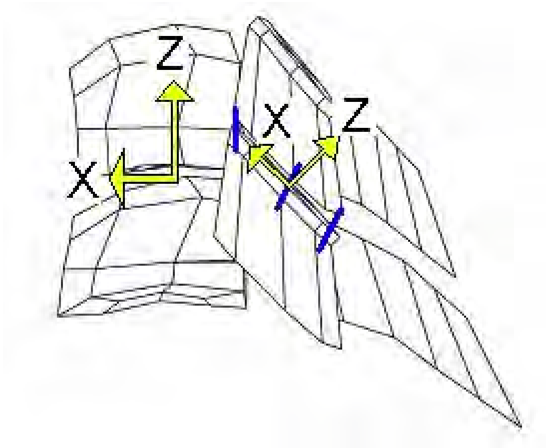


Figure 8.3 The isolated C5/C6 motion segment including representation of the intervertebral discs by large yellow X-Z axes and facet capsule sliding surface by small inclined yellow X-Z axes.

Elongations of the ligaments (ALL, FC and AF) are derived from the model, but it is easier to interpret the relative elongation or simple strain of the ligament. This was derived by the formula (Strain $\varepsilon = \Delta l/l$, where Δl is the change in length of the ligament or fibre as a result of the load, and l is the initial length). The stiffness and lengths used for the ligaments in the motion segment were given in Chapters 3, 5 and 6. The other loads chosen for investigation included the displacements and forces on the posterior bearing surfaces in the motion segment.

8.2.6 Prediction of Symptom Duration

The capability of using the C5/C6 motion segment model loading as a predictor for the persisting (greater than 1 month) whiplash symptoms of vehicle occupants was investigated by means of diagnostic tests based on the concept of positive and negative predictive values (Altman et al. 2000). The pain duration period of 1 month was selected because it gave a sufficient number of occupants for the analysis. A criterion value was selected based on a threshold (or cut-off point) for the component loading of interest. If the loading was greater than the threshold, then it was predicted that the subject was injured. The thresholds were sought to yield a sensitivity of 85%, the actual low cut off selected giving 86% sensitivity and the high 71%. The 2-by-2 contingency tables (Table 8.3) were constructed for each criterion, allowing the sensitivity, specificity, positive predictive value, negative predictive value and the 95% confidence intervals to be calculated.

Table 8.3 The 2x2 contingency table

Neck Injury Symptoms		**Long-term**	**No or initial**	**Total**
Criterion Value	Above threshold	a	b	a + b
	Below threshold	c	d	c + d
	Total	a + c	b + d	a + b + c + d

Based on the contingency table above, the following indexes are defined:

- Sensitivity = a/(a + c)

  The proportion of occupants with long-term symptoms that have estimated criterion values above a fixed level.

- Specificity = d/(b + d)

  The proportion of occupants with no symptoms or initial symptoms that have estimated criterion values below a fixed level.

- Positive predictive value = a/(a + b)

  The proportion of occupants with estimated criterion values above a fixed level that have long-term symptoms.

- Negative predictive value = d/(c + d)

  The proportion of occupants with estimated criterion values below a fixed level that have no symptoms or initial symptoms.

For each cut-off these four indices and the 95% confidence intervals were calculated for the selected possible predictors of injury. The 100(1–α)% confidence interval for the proportion in the population was calculated by the traditional method: $CI_{100(1-\alpha)\%} = p \pm \lfloor z_{1-\alpha/2} x SE(p) \rfloor$, where the standard error, $SE(p) = \sqrt{(pq/n)}$, p is the index of interest, $q = 1 - p$, and n is the numerator of the proportion (Altman et al. 2000). Altman et al. (2000) suggest that this method may give some problems when dealing with situations where no true positive is observed when calculating the sensitivity, but this is not true in this case. The analysis was repeated for both drivers-only and drivers and passengers combined.

8.2.7 Limitations of the Study

This study uses a series of linked simulations to reconstruct possible early injury causation in the C5/C6 motion segment of the neck, in real accidents. Each area of the study has its own set of limitations with respect to the data used and the capability of the simulation in faithfully reconstructing the original crash circumstances. The use of the in-vehicle crash pulse recording is a significant step forward with regard to the vehicle crash parameters such as vehicle velocity and impact direction, but some limitations remain with the real crash data.

The occupant factors, which are unknown, have an effect on the cervical spine kinematics during the crash. These factors include:

- The position and stature of the occupant;
- The inclination of the seat back;
- The position of the adjustable head-restraint;
- The occupant awareness of the crash; and,
- The flexibility and degenerative changes of the occupant's neck.

The use of the BioRID II dummy and seat model for the crash reconstruction (Eriksson and Kullgren 2003) has limitations in the following areas:

- The biofidelity of the BioRID II and seat model;
- A uniform position was assumed for the model of the BioRID II dummy and seat for all the case simulations, with no seatbelt and the head restraint in its lowest position; and,
- The BioRID II dummy stature is 50^{th}-percentile male and hence does not reflect the variation in anthropometry of the injured population.

The reconstruction of the neck motion with the head and neck modelling, Chapter 7, has limitations in the following areas:

- The neck was in the standard posture of an erect standing volunteer;
- The head and neck model represents a 50th-percentile male in size;
- The neck was using the passive muscles; and,
- The head restraint was not included.

Three factors limit the useable time duration of the modelling to the first 150 ms of the impact, all these factors have been discussed in Chapters 2 and 3:

- The lack of a seatbelt restraint in the BioRID II dummy simulation limits the dynamic similitude to the time before the dummy ramps up the seat back;
- The lack of head contact with the head restraint; and,
- The lack of active muscle forces in the neck restricts the similitude of the head and neck motion to before the model moves into hyperextension.

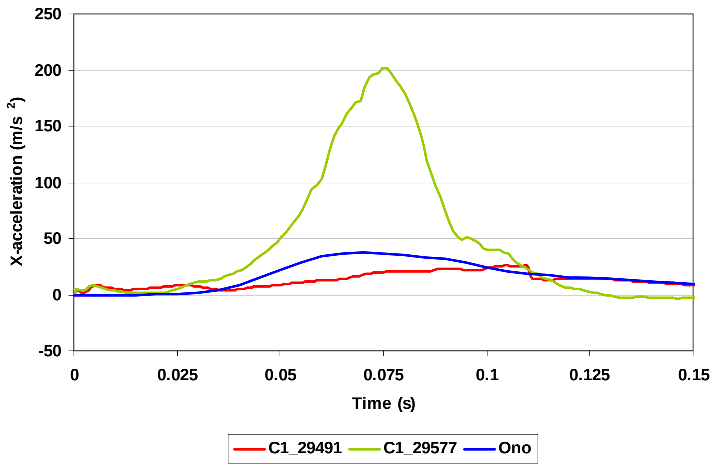
8.3 Results

8.3.1 Comparison of Real Crash Cases

Two example cases were selected and, with comparison to the Ono et al. (1997) volunteer test, are presented in detail to show the responses predicted by the C5/C6 model. The local T1 accelerations predicted by the BioRID II and seat reconstruction of the crash for two example cases from the accident data, C1_29491 and C1_29577, together with the Ono et al. (1997) volunteer test-pulse are presented in Figure 8.4.

The high-severity rear impact case, C1_29577, is one of the more severe crashes in the crash data, with a T1 x-acceleration (from the BioRID II reconstruction) of 20 g. The driver had persisting pain for longer than 6 months (Group 3) as a result of the impact. C1_29491 is a low-severity rear impact of 2.5 g (from the BioRID II reconstruction). The driver was uninjured and the passenger had pain for 1-month duration (Group 1). The sled pulse used for the Ono et al. (1997) 9.6 km/h volunteer test series is included for comparison. The Ono volunteer test series included no head restraint, no recorded injuries and a measured T1 x-acceleration of 3.5 g. The T1 z-acceleration for the volunteer is almost the same as that for the x-direction, which was not the case for the crash reconstructions where on average the z-acceleration is 40% of the x-acceleration.

(a) Local T1 x-direction acceleration



(b) Local T1 z-direction acceleration

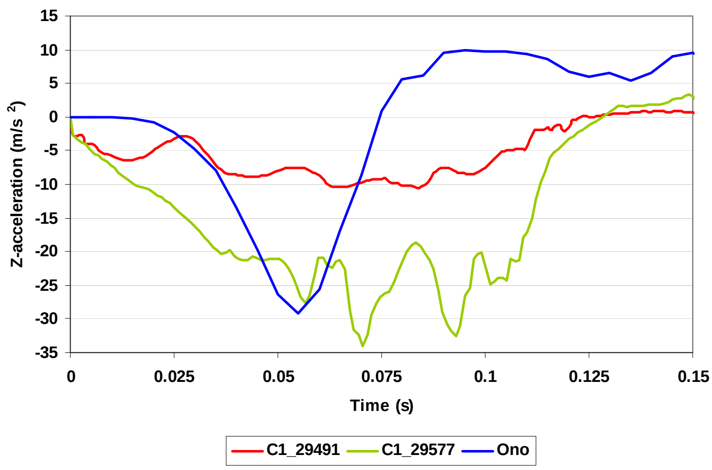


Figure 8.4 Local T1 accelerations predicted by the BioRID II and seat model for two cases, C1_29491 and C1_29577, were used to drive the head and neck model T1. Also included is the pulse shape for the Ono et al. (1997) 8.5 km/h volunteer test series for comparison.

The NIC_{max} predicted by the head and neck model for these two cases and the Ono volunteer test is shown in Figure 8.5. The initial peaks (at t = 0 ms) are due to the calculation method. The severe impact, C1_29577, resulted in a NIC_{max} of 31 (at t = 75 ms), the low-severity case, C1_29491, had a NIC_{max} of 4 (at t = 110 ms) and the Ono volunteer test had a NIC_{max} of 8 (at t = 70 ms). The injury criterion specifies that the impact is likely to result in longer-term injury if the positive NIC_{max} is greater than 15

(Bostrom et al. 1996). The high negative NIC_{max} peak of 37 (at t = 137 ms) for C1_29577 is possibly due to the BioRID II head impacting the head restraint in the initial crash reconstruction.

As previously discussed in Section 8.2.7, the head and neck simulation is only valid until approximately 150 ms, before the neck motion reaches hyperextension. Usually the peak values occur early in the impact before 150 ms, but the addition of a poorly designed head restraint and seatbelt has the potential to generate neck loads later in the impact.

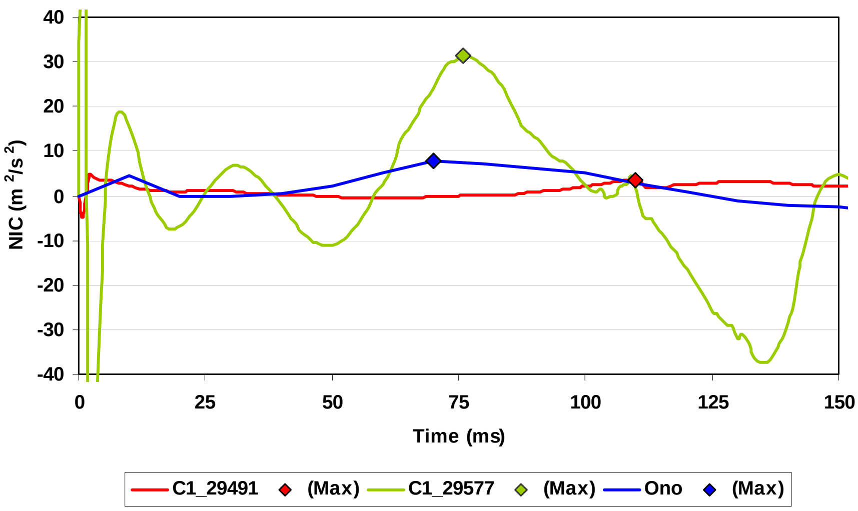


Figure 8.5 NIC_{max} predicted by the head and neck model for the two cases, C1_29491 and C1_29577, and the Ono et al. (1997) volunteer test. The small diamonds indicate the maxima.

The ligament elongations predicted by the head and neck model in the C5/C6 motion segment were then analysed for the two cases, C1_29491 and C1_29577, and the Ono et al. (1997) volunteer test. The ligaments in the anterior region of the motion segment had the maximum elongation: the anterior longitudinal ligament (ALL) elongation, Figure 8.6; the anterior FC ligament elongation, Figure 8.7; and the lateral anulus fibrosus (AF) fibre elongation, Figure 8.8. The ligament elongations resulting from the severe impact, case C1_28577, were significantly higher than for the other examples.

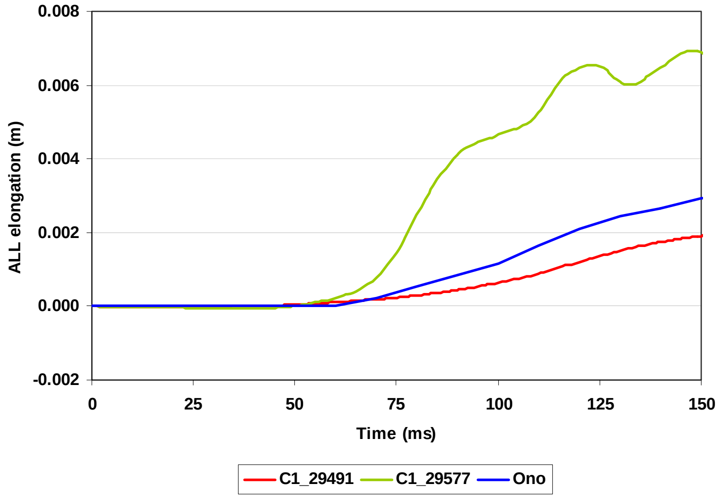


Figure 8.6 The C5/C6 anterior longitudinal ligament elongation predicted for C1_29491, C1_29577, and the Ono et al. (1997) volunteer test.

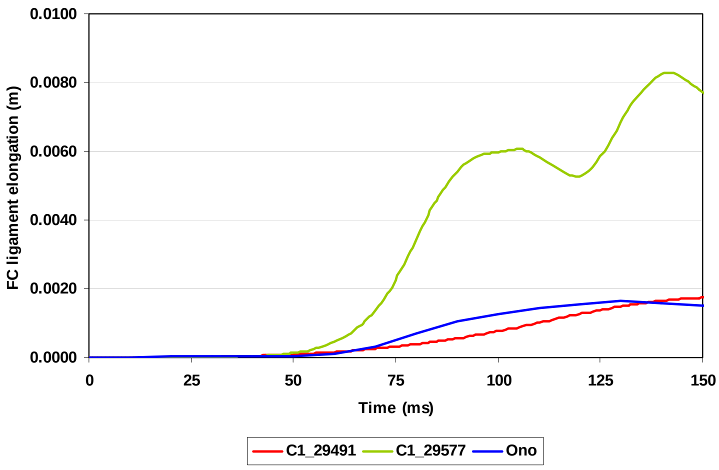


Figure 8.7 The C5/C6 anterior FC ligament elongation predicted for C1_29491, C1_29577, and the Ono et al. (1997) volunteer test.

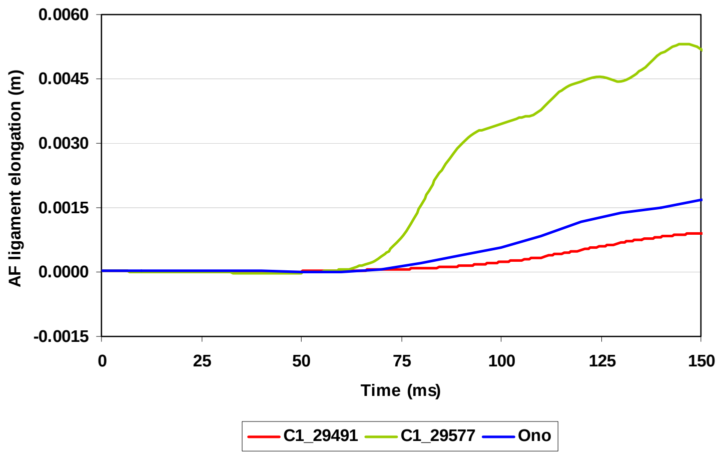


Figure 8.8 The C5/C6 anulus fibrosus lateral fibre elongation predicted for C1_29491, C1_29577, and the Ono et al. (1997) volunteer test.

The critical motion segment bearing-surface loads predicted by the C5/C6 model were also analysed for the three cases. These are presented as displacements, as the compression stiffness for the facet surfaces used in the motion segment model were assumed. The rear facet surface point restraint z-displacement with respect to time is plotted in Figure 8.9. A point restraint (PR) is the method used to represent the bearing surfaces of the motion segment in the model; a negative value means that the surfaces are moving closer and is an indication of possible facet surface impingement. For the dominant extension motion of the neck, the motion of the rear facet surfaces is distinctly different in the three examples. In all three cases, the posterior of the facet surfaces initially move together very slightly at about 50 ms. For C1_29491 the facet capsule continues further into compression while for the severe impact case C1_29577, the facet surfaces separate, as does the volunteer (to a lesser extent). The rear nucleus pulposa PR z-displacement with respect to time is plotted in Figure 8.10. A negative value is an indicator of compression on the facet surface. The displacement from median distance separating the facet planes (FPP) was also derived from the model and is measured at the rear edge of the facet surface using C6 as the reference. This is plotted in Figure 8.11; again a negative value is an indication of the surfaces narrowing the gap and possible facet surface impingement.

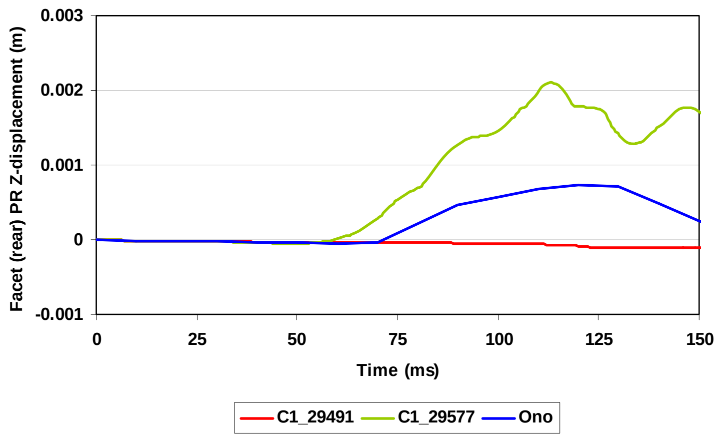


Figure 8.9 Facet surface rear point restraint z-displacement for C1_29491, C1_29577 and the Ono et al. (1997) volunteer test. A negative value is an indicator of possible facet surface impingement.

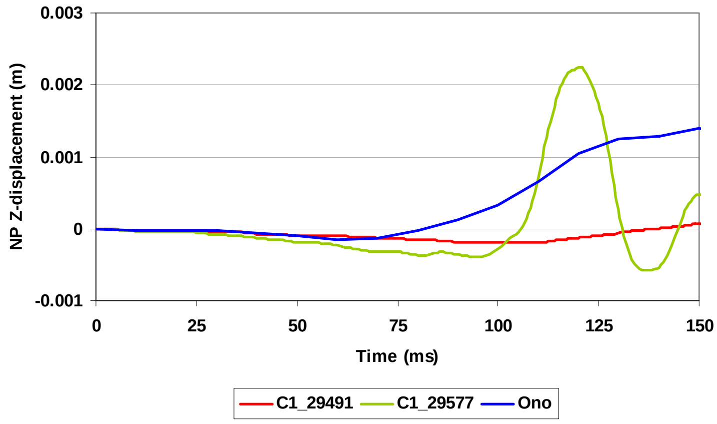


Figure 8.10 The NP back PR z displacement for C1_29491, C1_29577 and the Ono et al. (1997) volunteer test. A negative value is an indicator of compression.

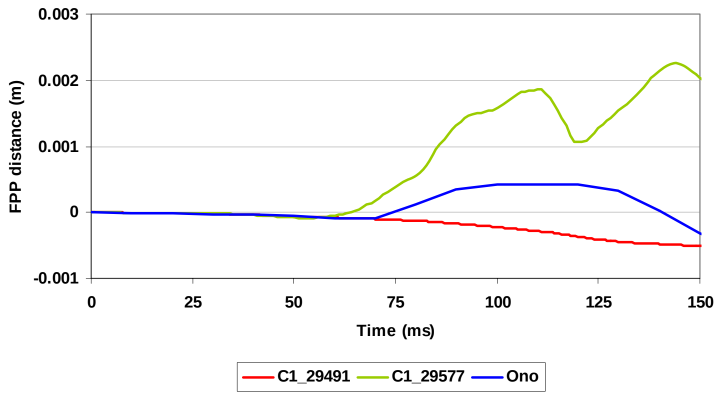


Figure 8.11 The minimum distance separating the facet planes (FPP) using C6 as the reference for C1_29491, C1_29577 and the Ono et al. (1997) volunteer test. A negative value is an indication of possible facet surface impingement.

8.3.2 The Crash Reconstructions

The simulations of all the crash cases (n = 78) were completed. Four of the more severe impact cases could not be used due to instability leading to asymmetrical responses. In retrospect, this may have been due to the use of an integration time-step in MADYMO that was too large.

The 74 remaining rear impact crashes with associated whiplash injury outcomes were classified into 4 neck injury groups by duration of the resultant pain: no pain (Group 0); pain duration for less than one month (Group 1); pain for more than one month but less than 6 months (Group 2); and, pain for greater than 6 months (Group 3). The separation distance of the posterior facet surfaces (FPP) for the C5/C6 motion segment were plotted against time for the above injury groups (Figure 8.12). The posterior edge of the C6 superior facet surface was the reference. Negative displacements indicate a decrease in distance between the posterior edges of the facets while positive displacements indicate an increased separation.

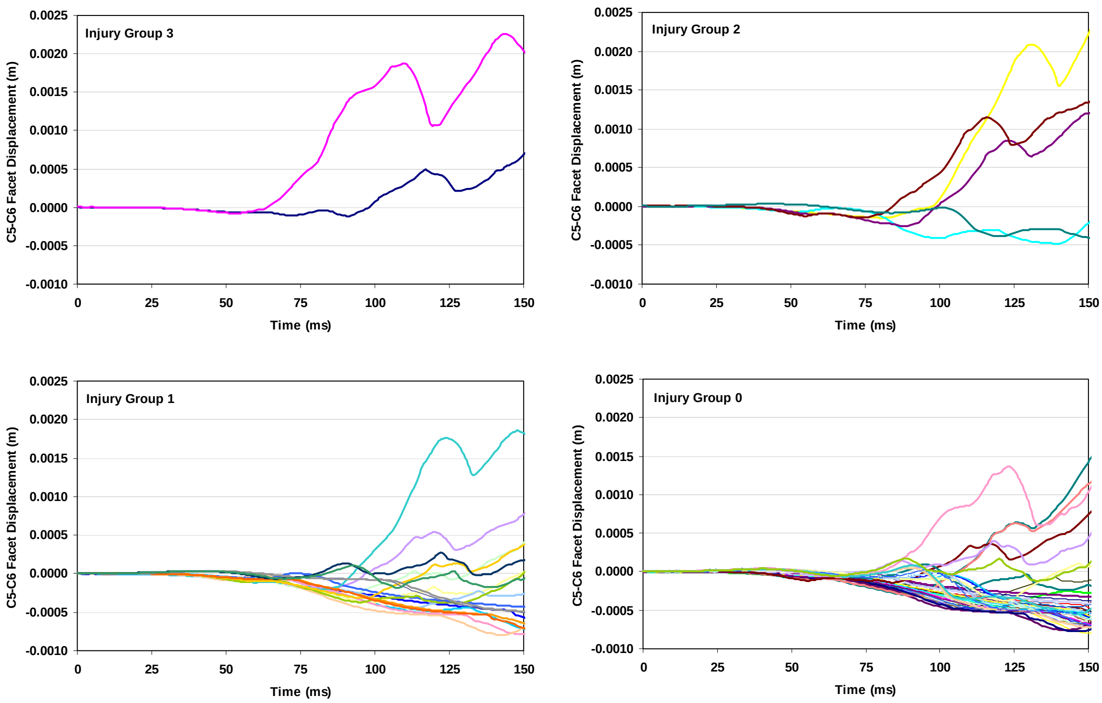


Figure 8.12 The separation of the facet surfaces (FPP) predicted by the C5/C6 motion segment model with time during the rear impact crash reconstructions (n = 74). The four neck injury groups are: No Injury (Group 0); Pain duration < 1 month (Group 1); 1 month < Pain duration < 6 month (Group 2); and Pain duration > 6 month (Group 3).

The cases appear to fall into one of two types of C5/C6 motion segment trajectory. The first type of motion is connected with the more severe impacts and is represented by the severe example C1_29577. The motion is dominated by the shear response, which separates the facet surfaces with little chance of impingement. The second group, represented by the low severity example C1_29491, has the rear facet surfaces moving together into compression for the early motion (up to *t = 150 ms*) and hence the possibility of some form of impingement. Group 3, with the highest injury, has 2-shear/0-impingement impacts, the next less severe injury group, Group 2, has 3-shear/2-impingement, Group 1 has 3-shear/16-impingement and Group 0 has 6-shear/52-impingement.

The predicted maxima of the C5/C6 motion segment components were plotted against the NIC_{max} along with the duration of symptoms in the following figures. The use of NIC_{max} gave higher correlations than other possible indicators, such as the pulse acceleration characteristics. The AF ligament relative elongation (strain) predicted for the C5/C6 motion segment (n = 74, correlation 0.840) is shown in Figure 8.13. A NIC_{max} of 15 is the accepted injury threshold. Comparison of the predicted anterior FC ligament strain versus NIC_{max} for the C5/C6 motion model showing the duration of symptoms (n = 74, correlation 0.903) is shown in Figure 8.14.

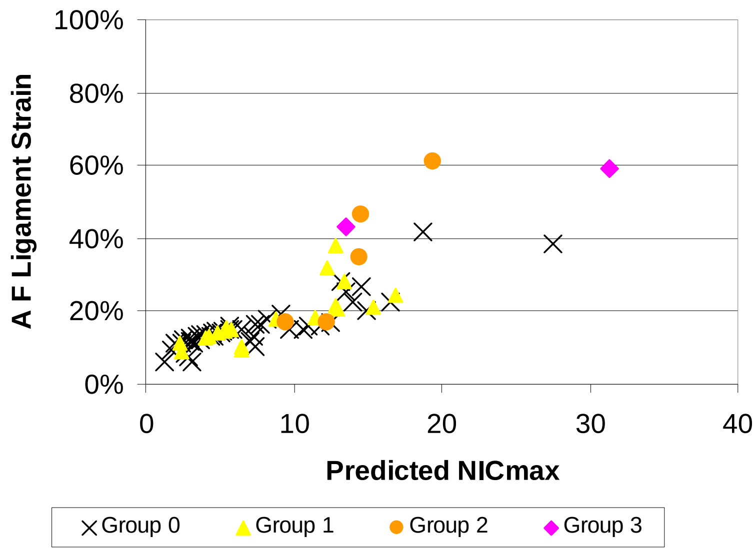


Figure 8.13 The predicted AF ligament relative elongation (strain) for the C5/C6 motion segment in the crash reconstructions, showing the duration of symptoms (n = 74, correlation 0.840). The injury threshold NIC_{max} = 15.

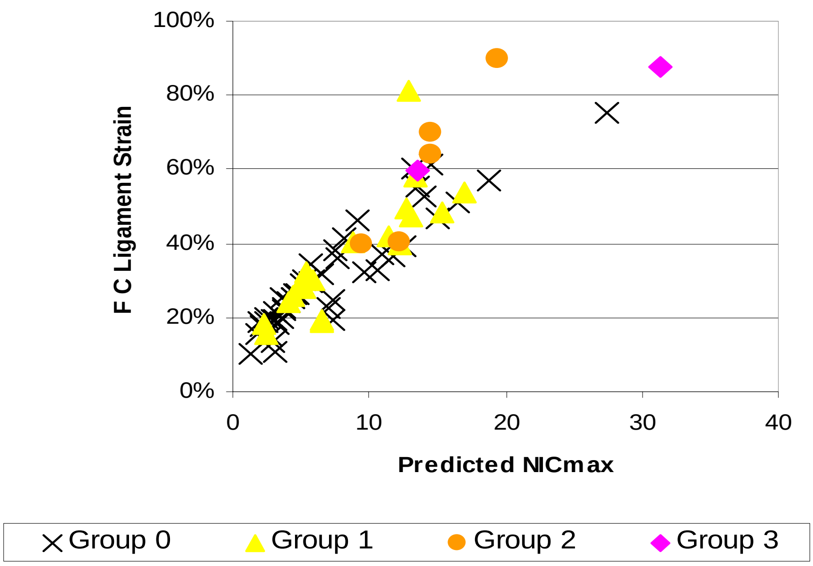


Figure 8.14 Comparison of the anterior FC ligament strain and NIC_{max} predicted by the head and neck model in the crash reconstructions, showing the duration of symptoms (n = 74, correlation 0.903). The injury threshold NIC_{max} = 15.

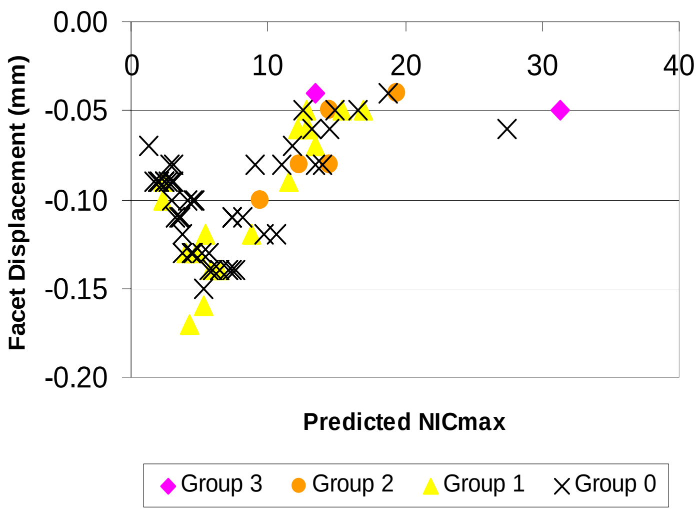


Figure 8.15 Comparison of the maximum rear facet surface compression and NIC_{max} predicted by the head and neck model in the crash reconstructions, showing the duration of symptoms (n = 74, correlation 0.612). The injury threshold NIC_{max} = 15.

The maximum rear facet surface compression displacement is plotted against predicted NIC_{max} for the C5/C6 motion segment (n = 74, correlation 0.612) is shown in Figure 8.15. The strains of the segment components, especially the anterior FC ligaments, show good correlations with NIC_{max}, while the correlation with facet surface separation was found to be poor.

8.3.3 Prediction of Symptom Duration

The ability of the C5/C6 peak loading data for the 74 drivers and 29 passengers remaining to predict the longer duration whiplash symptoms was analysed. Seven of the drivers and 3 passengers had longer duration whiplash symptoms (defined as greater than 1 month). The predictive capability of each load type from Table 8.2 was analysed for high and low cut-offs for drivers only and for drivers and front seat passengers combined. The thresholds were selected to give a sensitivity of 85%, as this has been found to give acceptable results for injury criteria (Kullgren et al. 2003). The actual low cut-off selected giving 86% sensitivity and the high 71%. The contingency tables were constructed for each criterion, allowing the sensitivity, specificity, positive predictive value, negative predictive value and the 95% confidence intervals to be calculated. The results for both the high and low cut-off values are given in Table 8.4 for the drivers only and Table 8.5 for the drivers and passengers combined.

The strains of the anterior aspects of the major C5/C6 motion segment ligaments, ALL, FC and AF were all good predictors of longer-duration pain outcomes. The x-displacements of the bearing components of the motion segment, AFx, NPx and the facet surfaces, Fx, were also good predictors of the occurrence of longer duration pain.

An injury criterion for use in designing safety equipment needs to predict a non-injurious event with absolute accuracy (100%), but only requires reasonable accuracy (10-50%) when predicting an injurious event. For the 85% sensitivity (the proportion of injured occupants above the chosen cut-off/threshold) used here, the negative predictive values for the good predictive parameters (ALL, FC and AF ligament strains; AFx, NPx, Fx and NPz displacements) were close to 100%. The proportion of all occupants below the threshold that were uninjured was close to 100%, or stated in a different way, this was the probability of correctly predicting a non-injurious event. The positive predictive value is approximately 30%. This is the proportion of all occupants above the threshold that were uninjured and is the probability of correctly predicting an injurious event. The corresponding thresholds were found to be 17% strain of the ALL, 50% strain of the FC, and 23% strain of AF – considering drivers only and the high cut-off point. The predictive capabilities were slightly improved when the both drivers and passengers were considered.

Table 8.4 The loading of the C5/C6 motion segment components

(a) For the low cut-off.

Component	Drivers Only				
	Cut-off	Sensitivity	Specificity	Positive Predictive Value (±Confidence Interval)	Negative Predictive Value (±Confidence Interval)
ALL	0.13	0.86	0.72	0.24 (±0.17)	0.98 (±0.04)
FC	0.40	0.86	0.74	0.25 (±0.17)	0.98 (±0.04)
AF	0.33	0.86	0.74	0.25 (±0.17)	0.98 (±0.04)
BF z	-36.3	0.86	0.00	0.08 (±0.06)	0.00 (±0.00)
F x	-0.003	0.86	0.75	0.26 (±0.18)	0.98 (±0.04)
F z	-.00004	0.86	0.00	0.08 (±0.06)	0.00 (±0.00)
AF x	-0.002	0.86	0.75	0.26 (±0.18)	0.98 (±0.04)
AF z	-.00006	0.86	0.13	0.09 (±0.07)	0.90 (±0.19)
NP x	-0.0025	0.86	0.75	0.26 (±0.18)	0.98 (±0.04)
NP z	-.00034	0.86	0.72	0.24 (±0.17)	0.98 (±0.04)
FPP	0.00044	0.86	0.04	0.09 (±0.07)	0.75 (±0.42)
NICmax	10.86	0.86	0.72	0.24 (±0.17)	0.98 (±0.04)

(b) For the high cut-off.

Component	Drivers Only				
	Cut-off	Sensitivity	Specificity	Positive Predictive Value (±Confidence Interval)	Negative Predictive Value (±Confidence Interval)
ALL	0.17	0.71	0.90	0.42 (±0.28)	0.97 (±0.04)
FC	0.50	0.71	0.85	0.33 (±0.24)	0.97 (±0.05)
AF	0.23	0.71	0.88	0.38 (±0.26)	0.97 (±0.04)
BF z	-41.8	0.71	0.06	0.07 (±0.06)	0.67 (±0.38)
F x	-0.004	0.71	0.85	0.33 (±0.24)	0.97 (±0.05)
F z	-0.00005	0.71	0.01	0.07 (±0.06)	0.33 (±0.53)
AF x	-0.003	0.71	0.85	0.33 (±0.24)	0.97 (±0.05)
AF z	-0.00007	0.71	0.40	0.11 (±0.09)	0.93 (±0.09)
NP x	-0.003	0.71	0.85	0.33 (±0.24)	0.97 (±0.05)
NP z	-0.0004	0.71	0.75	0.23 (±0.18)	0.96 (±0.05)
FPP	0.0004	0.71	0.06	0.07 (±0.06)	0.67 (±0.38)
NICmax	12.88	0.71	0.82	0.29 (±0.22)	0.96 (±0.05)

Table 8.5 The loading of the C5/C6 motion segment components for the drivers and passengers combined.

(a) For the low cut-off.

Component	Drivers and Passengers				
	Cut-off	Sensitivity	Specificity	Positive Predictive Value (±Confidence Interval)	Negative Predictive Value (±Confidence Interval)
ALL	0.13	0.89	0.73	0.24 (±0.15)	0.99 (±0.030)
FC	0.40	0.89	0.76	0.26 (±0.15)	0.99 (±0.03)
AF	0.33	0.89	0.76	0.26 (±0.15)	0.99 (±0.03)
BF z	-36.3	0.89	0.00	0.08 (±0.05)	0.00 (±0.00)
F x	-0.003	0.89	0.77	0.27 (±0.16)	0.99 (±0.03)
F z	-0.00004	0.89	0.00	0.08 (±0.05)	0.00 (±0.00)
AF x	-0.002	0.89	0.77	0.27 (±0.16)	0.99 (±0.03)
AF z	-0.00004	0.89	0.04	0.08 (±0.05)	0.80 (±0.35)
NP x	-0.002	0.89	0.77	0.27 (±0.16)	0.99 (±0.03)
NP z	-0.0003	0.89	0.72	0.24 (±0.14)	0.99 (±0.03)
FPP	0.0004	0.89	0.03	0.08 (±0.05)	0.75 (±0.42)
NICmax	10.85	0.89	0.72	0.24 (±0.14)	0.99 (±0.03)

(b) For the high cut-off.

Component	Drivers and Passengers				
	Cut-off	Sensitivity	Specificity	Positive Predictive Value (±Confidence Interval)	Negative Predictive Value (±Confidence Interval)
ALL	0.17	0.78	0.87	0.37 (±0.22)	0.98 (±0.03)
FC	0.50	0.78	0.85	0.33 (±0.20)	0.98 (±0.03)
AF	0.46	0.78	0.87	0.37 (±0.22)	0.98 (±0.03)
BF z	-39.90	0.78	0.02	0.07 (±0.05)	0.50 (±0.49)
F x	-0.004	0.78	0.85	0.33 (±0.20)	0.98 (±0.03)
F z	-0.00005	0.78	0.01	0.06 (±0.05)	0.25 (±0.42)
AF x	-0.003	0.78	0.85	0.33 (±0.20)	0.98 (±0.03)
AF z	-0.00006	0.78	0.14	0.08 (±0.06)	0.87 (±0.17)
NP x	-0.003	0.78	0.85	0.33 (±0.20)	0.98 (±0.03)
NP z	-0.0004	0.78	0.76	0.23 (±0.15)	0.97 (±0.03)
FPP	0.00041	0.78	0.043	0.07 (±0.05)	0.67 (±0.38)
NICmax	12.88	0.78	0.84	0.32 (±0.19)	0.98 (±0.03)

8.4 Discussion

Two cases and the comparison with the Ono et al. (1997) volunteer test are presented in detail to show the responses of the C5/C6 model. The high-severity rear impact case, C1_29577, is one of the more severe crashes in the data with a T1 x-acceleration of 20 g (from the BioRID II reconstruction). As a result of the impact the driver had pain of longer than 6 months duration (Group 3). In comparison, C1_29491 is a low-severity rear impact of 2.5 g (from the BioRID II reconstruction). The driver was uninjured and the passenger had pain for 1-month duration (Group 1). The Ono volunteer test was non-injurious and had a measured T1 x-acceleration of 3.5 g. Later in the time period (after 140 ms) the neck extension simulated becomes excessive in comparison to the volunteer, as active muscles do not control it.

In the more severe case C1_29577, the acceleration pulse causes significantly higher elongations of all the ligaments. For the ALL the maximum ligament strain was 36% (Figure 8.6) and for the FC it was 90% (Figure 8.7). Both are likely to produce catastrophic ligament failure. Failure strains of 31% (±5.9%) for the ALL and 116% (±19.6%) for the FC were measured quasi-statically by Yoganandan, Kumaresan and Pintar (2000). As allowable dynamic ligament failure loads have been found to be rate dependent Yoganandan et al. (1998a), these predicted ligament strains fall in the area of sub-catastrophic failure. The strain levels for C1_29491 and the Ono volunteer test are significantly lower and are unlikely to be injurious (Table 8.6). The NIC_{max} predictions in Figure 8.5 are in agreement with this interpretation, when the possibility of facet impingement for the three cases is reviewed (Figure 8.11). The predicted dynamic motion of C5/C6 for C1_29491 and the Ono volunteer appear to make such an event possible, with a decrease in the facet surface separation occurring between 40 and 75 ms. This prediction of the dynamic facet motion agrees with the Ono et al. (1997) facet impingement hypothesis. The shear loading to C1_29577 appears to be so rapid that the motion segment is rapidly stretched in shear and no significant decrease occurs. This is in agreement with the shear hypothesis suggested by Deng et al. (2000b).

The hierarchy of injury suggested by Taylor and Taylor (1995) for the injury to the soft tissue of the neck is supported by these results. For the neck ligaments, the ALL has the lowest predicted strain in a rear impact and is also the longest, followed by the AF and the FC, which are each progressively shorter and have increasingly higher strain (Table 8.6). In the severe impact example C1_29577, the maximum strain in the C5/C6 FC anterior ligaments predicted by the model is in the sub-catastrophic failure area found by Siegmund et al. (2000b). When all the crash data is investigated, the predicted average strains for the (n = 7) cases with pain duration of greater than 1 month, the average peak strain is 64% (±20%). For the (n = 19) cases with pain duration less than 1 month, the predicted average strain is 37% (±17%) and for the remainder of cases with no pain (n = 48) the predicted strain is 31% (±15%). All this indicates a high level of correlation between the predicted anterior facet ligament strain and the duration of pain in this sample of crashes, which supports the shear mechanism of whiplash injury as suggested by Deng et al. (2000b) and others.

Table 8.6 The predicted C5/C6 motion segment maximum strain for the two example crashes, C1_29577 and C1_29491, and the Ono volunteer test with the failure strains.

Ligament	C1_29577		C1_29491		Ono Volunteer		Failure Strain % (±SD)
	Strain %	Injury	Strain %	Injury	Strain %	Injury	
ALL	36	+	10	–	18	–	31 (±5.9) [a.]
AF fibre	60	+	13	–	23	–	60 [b.]
FC	90	+	25	–	26	–	116 (±19.6) [a.] 94 (±85) [c.] 35 (±21) [d.]

Source: a. Yoganandan et al. (2000)
b. Pintar et al. (1986)
c. Siegmund et al. (2000b).
d. Sub-catastrophic failure, Siegmund et al. (2000b).

The shear mechanism is further supported by the NIC_{max} criterion. NIC_{max} has very good correlation with predicted elongation of the motion segment ligaments (above 0.90) (Figure 8.13 and Figure 8.14), but only poor correlation (0.61) with the predicted facet separation (Figure 8.15). The spread of the individual data points in this figure seems to indicate that there may be another factor or factors involved in the injury causation. The facet surface separation has a possible connection with the other hypothesised mechanism of injury of the impingement of the facet surfaces. This was also investigated further with the head and neck model predictions. The predicted C5/C6 facet surface separation with time was plotted in Figure 8.12. The plots are separated into the 4 injury groups and there appears to be two distinctive impact motions:

1. A shear dominated C5/C6 vertebral motion which causes positive separation of the facet surfaces; and
2. A rotation dominated C5/C6 vertebral motion, which causes negative separation of the facet surfaces and has the possibility of facet impingement.

When the predicted FC ligament strain is plotted against the rear facet surface compression force, an interesting pattern arises (Figure 8.16). The two Group 3 injury cases, the most severely injured vehicle occupants in the study, have shear-dominated motion, which includes positive separation of the facet surfaces.

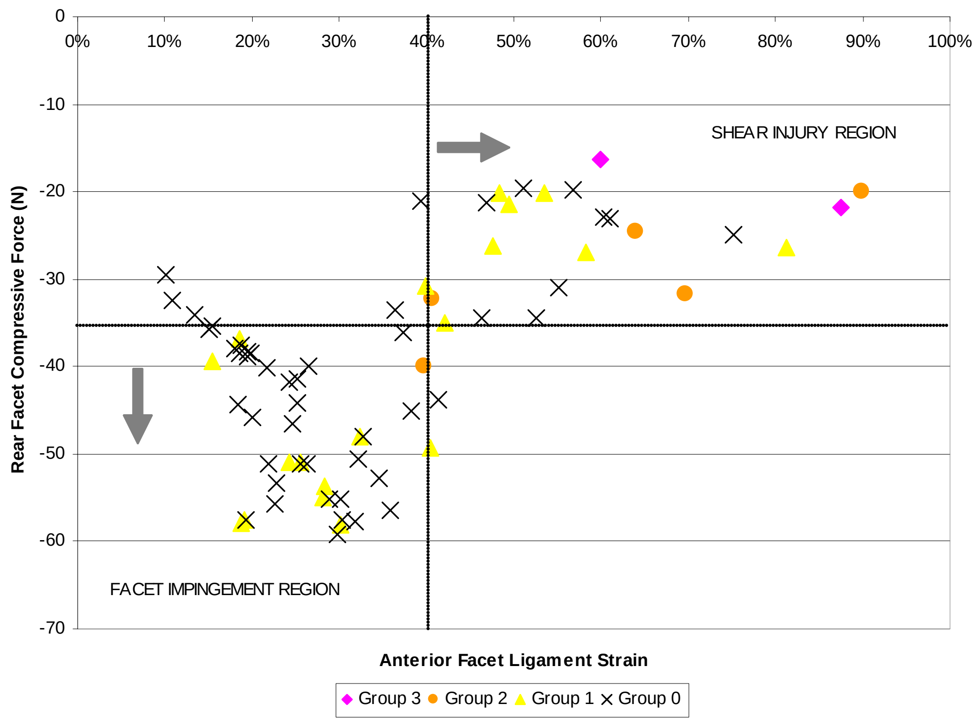


Figure 8.16 Predicted C5/C6 facet capsular ligament strain versus twice the rear facet surface compression force for the drivers only (n = 74) in the crash reconstructions with the different injury outcomes. The dotted lines and arrows represent the greater than 40% strain and 35N compression regions.

Three of the five cases from Group 2 also have separation of the facet surfaces. The remaining two cases have contact of the facet surfaces predicted. A similar split in the motion types is found in the lower-severity injury groups with initial pain, and no injury. It is hypothesised based on the C5/6 model predictions, that at the higher impact severities the head and neck the shear displacement velocity of C5 with respect to C6 becomes great enough that the facet capsular surfaces separate before the extension rotation of C5 with respect to C6 occurs (shear injury to the facet capsule). It is the timing and magnitude of the extension rotation of C5 with respect to C6, which causes the facet surfaces to impinge.

Figure 8.16 shows the shear injury region due to facet ligament strain greater than 40% in the upper right. The figure of 40% shear is derived from the statistical analysis. The diagram also shows where possible facet impingement could occur with a predicted force on the rear facet surface of greater than 35 N. In the sample of rear impacts investigated there was only one of the longer-term pain cases from Groups 2 and 3, which may have been influenced by the compression on the facets.

8.5 Summary

The detailed C5/C6 motion segment model was applied to the investigation of early AIS1 soft-tissue neck injury in rear impacts. It was used to investigate early soft tissue neck injury in a group of real crashes by linking the results from crash reconstructions of field accident data (based on the use of in-vehicle crash recorders), with the clinical pain outcomes of the vehicle occupants over a 12-month period.

The C5/C6 motion segment model was included in the MADYMO based human head and neck model developed by van der Horst et al. (2001). The head and neck model was driven by T1 accelerations derived from the reconstruction of a group of real life crashes using a MADYMO BioRID II dummy and seat model, where the injury outcome was known (Kullgren et al. 2003). Injuries to the occupants of the crashed vehicles were available in the form of pain duration following the impact. This allowed the comparison of the early C5/C6 motion segment component strains resulting from the crashes to be correlated with the real life AIS1 neck injuries to the vehicle occupants. Good correlations were obtained with predicted shear-related elongation of the model C5/C6 motion segment joint ligaments, with critical levels of shear strain occurring in the facet capsular ligaments. The values predicted for shear strain in the anterior facet ligaments were equivalent to sub-catastrophic failure strains measured by Siegmund et al. (2000b) *in-vitro* laboratory tests. The threshold for long-term pain duration as a result of a rear impact is at 40 % strain of the FC ligaments.

For the crash reconstructions, the C5/C6 model predicted two distinct impact motions: a shear-related motion and a rotation-related motion with possible facet impingement. This supports two of the main hypotheses of early whiplash-associated injury causation suggested by Deng et al. (2000b) and Ono et al. (1997).

CHAPTER 9 DISCUSSION, CONCLUSION AND RECOMMENDATIONS

9.1 Discussion

Field accident studies have shown that the typical sufferer of chronic neck pain resulting from a motor vehicle accident is female, mid-thirties in age, who was involved in a rear impact while stationary, with symptomatic intervertebral joints at the C2/3 and C5/6 levels of the neck and with probable involvement of the facet capsule. Whiplash injury not only occurs in rear impacts, but may also result from all directions of impact in motor vehicles.

The field accident studies also showed that when the seatback was deformed in a rear impact, there was less likelihood of the occupant sustaining whiplash-related injuries. Head restraints have been proven only partially effective, while the use of seatbelts seems to increase the risk of injury.

Some investigators have linked the more severe injuries to a lack of awareness of the impending impact and to having the head out of position, for example with the neck turned to observe the rear-view mirror. The awareness issue has been linked to the possibility of pre-impact tensing reducing the possible injury, and this is consistent with the results of volunteer studies, which show reductions in head motion with awareness.

The vehicle factors implicated by the field studies have included the stiffness of the vehicle rear structure, the dynamic stiffness of the seat back and head restraint, and the detail design of the seatbelt. It has been shown however, that a well-designed seat and head restraint is able to overcome the other factors.

The prevention of injury requires the injury mechanism to be understood: the lack of understanding in this area has been a handicap in the implementation of engineering solutions. Attempts at regulating the dimensions and stiffness of head restraints have been only moderately effective, as has the availability of consumer information based on static testing of the head-restraint position. As a result, industry is developing a dynamic test requirement based on the BioRID II dummy in the vehicle seat.

The factors influencing the biomechanics of cervical spine soft-tissue injury include aspects of the neck anatomy, clinical data, autopsy data, and the results of many experimental studies using animal, human cadaver and human volunteer models to investigate these types of injury. The convergence noted by Barnsley, Lord and Bogduk (1998) of many of these factors, along with the findings of the field accident studies has become more apparent. Injuries to the facet capsule of the neck have been shown to be a major source of post-crash pain and there are many hypotheses on how such injuries may occur. To investigate these hypotheses, it is necessary to develop an investigative tool to establish the causal links connecting the mechanical load to the neck in a motor vehicle crash and the symptoms of WAD sufferers. Two of these hypotheses are related to events early in the impact and formed the basis for the investigation here.

Preliminary investigation of the accident characteristics associated with the 88 drivers suffering chronic neck pain confirmed the more general field accident studies in regard to gender bias and rear impact susceptibility. The impact severity was found to be at the high end of the usual range. The most common confirmed pain sites were to the facet capsule at C2/C3 (34%) and C5/C6 (32%). By combining this information with other accident and clinically based data separate from the insurance claims data, the incidence of whiplash injury in Australia (in 1998) was estimated to be 170 cases per 100,000/year. The definition and treatment of whiplash claims varies between states in Australia. The cost of these whiplash injuries to the Australian community (in 1998) was estimated to be $540 million.

The most appropriate model for investigating the soft-tissue neck injury was a mathematical model of the head and neck. For the combination of the human body responses, the head and neck kinematics during a rear impact, and the required strains in the soft tissues of the neck, a multi-body mathematical model with detailed soft tissue for the motion segment was chosen as the best option. Even with the limitations imposed by the multi-body format, such a model could be developed to investigate soft-tissue neck injury causation at the motion segment level in dynamic loading situations. The available information regarding both the anatomy and the properties of the soft tissue components supported this approach. This detailed motion segment was then to be integrated into a human head and neck model for the purposes of dynamic validation and application in realistic dynamic loading situations. It was also necessary for the head and neck model to be compatible with a full human body model, to allow integration again for the purposes of validation with volunteer tests and for the application of the model in investigating of soft-tissue neck injury in rear vehicle impacts. The model that was found to best fulfil these requirements was the van der Horst (2002) head and neck model.

A new C5/C6 motion segment model was developed for the van der Horst (2002) head and neck model with representation of the anatomical soft-tissue structures, giving improved biofidelity and the ability to predict soft tissue injury resulting from dynamic loading. The structure of the disc developed followed the anatomy described by Mercer and Bogduk (1999).

The small load responses of the C5/C6 motion segment model were developed to agree with the *in-vitro* experimental results obtained by Moroney et al. (1988) for all directions of direct loading. The C5/C6 motion segment model responses were validated for large loads by comparison with further *in-vitro* tests up to failure. The model motion segment results were in good agreement with the combined posterior shear and compression tests by Siegmund et al. (2000b). In more complex combined loading situations (Winkelstein et al. 2000), it still maintains the correct predictive trends and this makes the model suitable for the purpose intended in this study of whiplash-associated loading. The whiplash injuries are sub-catastrophic injury of the neck soft tissue and occur close to the small load range for which the model is well validated. Some care should be taken if the C5/C6 model is used in the catastrophic failure area.

It was necessary to re-integrate the C5/C6 motion segment model with a head and neck model to allow the application of realistic dynamic loading for the investigation of whiplash injury mechanisms. The integration of the new segment with the human head and neck model by van der Horst (2002) was a simple process. The dynamic responses of this new model to rear impact loading were validated in several stages with the Ono et al. (1997) volunteer test data series and against the original van der Horst model. The major limitations of the head and neck model, as used in the study, were found to be with the capabilities in the original van der Horst (2002) model itself. This was outside of the scope of this study, which was aimed at developing the motion segment capabilities. The van der Horst model was used only as a means of applying realistic impact loads to the motion segment. This study does give some indications of where future developments should be aimed to improve the model responses. The application of active muscles is of great importance in this regard. The van der Horst model needs to be developed to more accurately simulate the individual motion segment rotation and shear motions when compared with volunteer testing. To enhance the capability of the van der Horst model in these areas will most likely require a solution to the active muscle problem.

The detailed C5/C6 motion segment model was applied to the investigation of long duration pain outcomes to vehicle occupants in rear impacts. It was used to investigate early soft-tissue neck injury in a set of real crashes by linking the results obtained from crash reconstructions of field accident data, based on the use of in-vehicle crash recorders, with the clinical pain outcomes of the vehicle occupants over a 12-month period. The C5/C6 motion segment model was driven by T1 accelerations derived from the reconstruction of real-life crashes, where the injury outcome was known, using a MADYMO BioRID II dummy and seat model (Kullgren et al. 2003). WAD outcomes to the occupants of the crashed vehicles were available in the form of pain duration following the impact. This allowed the early C5/C6 motion segment component strains resulting from the crashes to be correlated with real-life AIS 1 neck injuries to the vehicle occupants. Good correlations were obtained with predicted shear-related elongations of the model C5/C6 motion segment joint ligaments, with critical levels of shear strain occurring in the facet capsular ligaments. The values predicted for shear strain in the anterior facet ligaments were equivalent to sub-catastrophic failure strains measured by the Siegmund et al. (2000b) *in-vitro* laboratory tests. The threshold for the injury resulting in longer-term pain duration as a result of a rear impact was at 45% strain of the facet capsule ligament.

Of equal significance, the C5/C6 motion segment model predicted that two distinct impact motions due to rear impacts in vehicles could occur at this level of the neck. One was directly shear-related and the other more rotation-related with facet impingement. The specific motion occurring in an impact was due to the shape of the pulse applied at T1. This supports two of the main hypotheses of early whiplash associated injury: that suggested by Deng et al. (2000b) and Ono et al. (1997). It also strongly supports the work in progress to derive an appropriate dynamic rear-impact test methodology for the design of seat systems for motor vehicles.

9.2 Limitations of the Study

This study uses a series of linked simulations to reconstruct possible early injury causation in the C5/C6 motion segment of the neck in real accidents. Each area of the study has its own set of limitations with respect to the data used and the capability of the simulation in faithfully reconstructing the original crash circumstances. The use of the in-vehicle crash pulse recording is a significant step forward with regard to the vehicle crash parameters including vehicle velocity and impact direction, but some limitations remain with the real crash data.

The occupant factors, which are unknown, have an effect on the cervical spine kinematics during the crash. These factors include:

- The position and stature of the occupant;
- The inclination of the seat back;
- The position of the adjustable head-restraint;
- The awareness of the crash; and
- The flexibility and degenerative changes of the occupant's neck.

The use of the BioRID II dummy and seat model for the crash reconstruction (Eriksson and Kullgren 2003) has limitations in the following areas:

- The limits to the biofidelity of the BioRID II and seat model;
- A uniform position was assumed for the model of the BioRID II dummy and seat for all the case simulations, with no seatbelt and the head restraint in its lowest position; and,
- The BioRID II dummy stature is 50th-percentile male and hence does not reflect the variation in anthropometry of the injured population.

A major limitation with the use of the head and neck model was found to be in the capabilities of the van der Horst (2002) human head and neck model. The head and neck model was used as a means of applying realistic impact loads to allow the motion segment capabilities to be used. The specific reconstruction of the neck motion with the head and neck modelling, described in Chapter 7, has limitations in the following areas:

- The neck was in the standard posture of an erect standing volunteer;
- The head and neck model represents a 50^{th}-percentile male in size;
- The neck was using the passive muscles; and,
- The head restraint was not included.

Three factors limited the useable time duration of the modelling to the first 150 ms of the impact:

- The lack of a seat belt restraint in the BioRID II dummy simulation limits the dynamic similitude to the time before the dummy ramps up the seat back;
- The lack of head contact with the head restraint; and
- The lack of active muscle forces in the neck restricts the similitude of the head and neck motion to the period before the model moves into hyperextension.

9.3 Conclusions

A mathematical C5/C6 motion segment model able to investigate the causation of soft-tissue neck injury in rear impacts has been developed. The predicted motions and injuries of the C5/C6 model have been validated with available static *in-vitro* experimental data.

The C5/C6 model has been integrated into a human neck model to allow the application of dynamic loads and the predicted motions and injuries of the C5/C6 model validated by comparison with available dynamic experimental data for volunteers and cadavers in rear impacts.

The injury sensing capabilities of the C5/C6 model have been demonstrated by applying it to the investigation of early soft-tissue neck injury in a group of real crashes by linking the results obtained from crash reconstructions of field accident data, based on the use of in-vehicle crash recorders, with the clinical pain outcomes of the vehicle occupants over a 12-month period.

Based on these reconstructions the correlation of early soft-tissue injuries to the neck with the available neck injury criteria was made. Two of the hypotheses on early injury causation to the facet capsule, the shear hypothesis (Deng et al. 2000b) and the facet impingement hypothesis (Ono et al. 1997), have been further confirmed and a possible early injury tolerance criterion was suggested.

Despite the limitations of the modelling process as a result of the simplifications and assumptions, the results obtained suggest that the detailed C5/C6 model possesses adequate human biofidelity and has been proved to be a useful tool in investigating soft-tissue neck injury.

9.4 Recommendations for Future Work

The recommendations for future work cover several areas: specific requirements for test data to make improvements in the motion segment model; improvements to the human neck model; and a more comprehensive reconstruction of the Folksam crash data using a human body model rather than a dummy.

The C5/C6 motion segment model would gain improvements with the availability of additional *in-vitro* test data. This test data is required in the areas of the tension capability of the anulus fibrosus fibres and the compression capability of the disc and the facet capsule. The concepts developed in the C5/C6 model should also be extended to a C2/C3 motion segment model.

The head and neck model was used as a means of applying realistic impact loads to allow the motion segment capabilities to be used. Some of the limitations of this part of the project were due to the limitations of the van der Horst (2002) human head and neck model. The dynamic modelling of the human neck is currently restricted by the difficulties inherent in the timing and magnitude of the application of active muscles. The variations in the rotation and shear motions of individual motion segment levels predicted by the head and neck model when compared with volunteer tests need to be addressed. It is possible that these areas are related to a variety of simulation errors such as postural variations with the volunteers, being prepared for the impact and the early activation of the muscles in the volunteers. The human head and neck model also needs an improved C2/C3 motion segment to address the facet capsule injury in the upper neck.

Finally, if these improvements were available for the motion segment and for the head and neck, there is a case for repeating the study using an integrated human body model in a seatbelt and with a realistic vehicle seat and headrest. At the current stage of development, this extended study would be useful as it would be able to address most of the remaining questions regarding the roles of the head restraint, the rebound from the seat back, and the seatbelt in the causation of injury to the soft tissues of the neck.

Such work could also be usefully extended into other directions of impact and the related soft-tissue neck injury causation.

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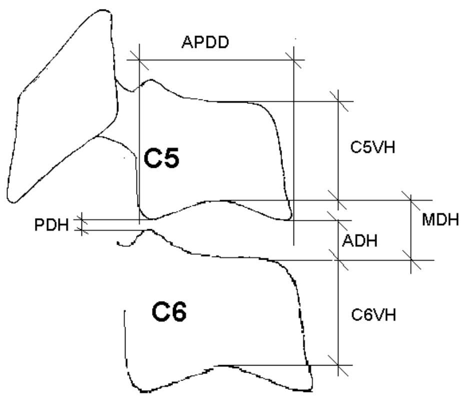
APPENDIX A1 CERVICAL SPINE RESEARCH UNIT DIAGNOSTIC AND CRASH DATA

This Appendix is not available for distribution due to the requirements of the NSW Privacy Committee.

APPENDIX A2 C5/C6 MOTION SEGMENT MEASUREMENT DATA

Table A2.1 C5/C6 discs were measured from radiographs of a group of 20 non-symptomatic patients from CSRU, Newcastle.

Number	ADH (mm)	MDH (mm)	PDH (mm)	APDD (mm)	C5VH (mm)	C6VH (mm)
1	3	6	4	20	16	15
2	4	8	6	20	12	11
3	2	3	3	28	15	15
4	4	7	4	26	16	17
5	3	6	1	26	21	17
6	5	7	3	22	15	14
7	2	5	2	20	15	12
8	5	7	5	22	15	14
9	2	4	1	20	14	14
10	7	7	2	25	15	14
11	4	6	3	25	18	18
12	6	6	3	20	15	15
13	5	6	2	18	14	13
14	3	5	2	20	15	15
15	3	6	2	22	16	15
16	6	7	2	24	15	15
17	4	8	5	21	16	16
18	4	7	4	24	17	14
19	5	5	2	19	13	13
20	6	10	3	24	17	18
Sum	83	126	59	446	310	295
Average	4.15	6.30	2.95	22.30	15.50	14.75
SD	1.46	1.53	1.36	2.77	1.88	1.83



ADH	Anterior Disc Height
MDH	Mid Disc Height
PDH	Posterior Disc Height
APDD	Anterior /Posterior Disc Diameter
C5VH	C5 Vertebral Height

Figure A2.1 Motion Segment Dimensions

APPENDIX A3 LABORATORY QUASI-STATIC IN VITRO NECK MOTION SEGMENT TESTING

A3.1 Moroney et al (1988)

A3.1.1 Test Method

The load-displacement behaviour of 35 fresh adult cervical spine motion segments was measured in compression, shear, flexion, extension, lateral bending and axial torsion tests. Motion segments were tested both intact and with posterior elements removed. Applied forces ranged to 73.6 N in compression and to 39 N in shear, while applied moments ranged to 2.16 Nm. For each mode of loading, principal and coupled motions were measured and the stiffness were calculated. The effect of disc degeneration on motion segment stiffness and the moments required for motion segment failure were also measured.

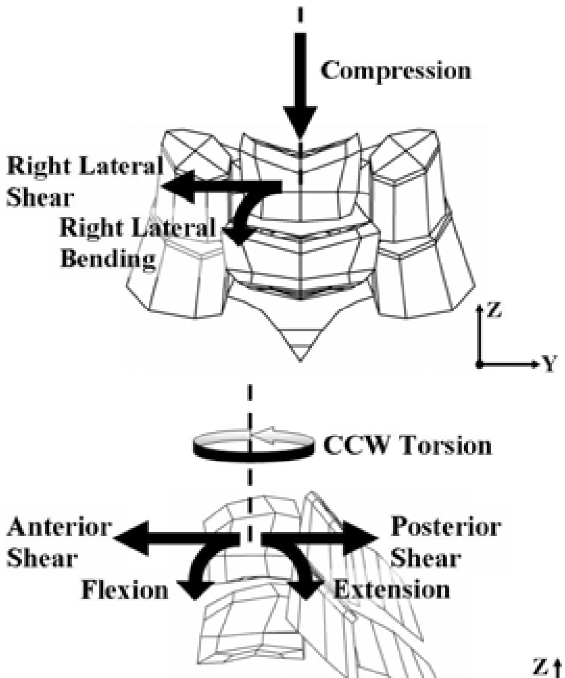


Figure A3.1 Loading of the C5/C6 motion segment used by Moroney

A3.1.2 Results

Motion coupling patterns of cervical motion segments were found:

- In sagittal and lateral bending, anteroposterior and lateral shear displacements were observed.
- In compression and shear displacements there were no significant coupled motions.

Removal of the posterior elements tended to increase all mean principle motions and so decrease segment stiffness. Cervical disc segments were as much as 50 % less stiff than intact segments in all modes of loading except posterior shear, where this effect was not evident. It was also found that removal of the cervical posterior elements approximately doubled the compression displacements.

No relationship of cervical motion segment stiffness to disc level was evident.

Severely degenerated cervical disc segments were 50 % less stiff in compression and three times stiffer in shear than were less-degenerated segments.

A3.2 Siegmund et al. (2000)

A3.2.1 Test Method

The purpose of this study was to examine the effect of axial preload on the kinematics of the motion segment and to examine the effect of multi-axial loading on the potential for facet capsular ligament injury. Two motion segments (C3-4 and C5-6) from seven female donors (50 ± 10 years) were exposed to quasi-static posterior shear loads of 135 N applied to the superior vertebra on four occasions while under compressive axial preloads of 0 N, 45 N, 197 N and 325 N. The vertebral body motions and the full Lagrangian strain field in the facet capsular ligament were measured. The facet joint of each motion segment was then isolated and failed in posterior shear. The grade of degeneration of the intervertebral discs in the thirteen specimens was between 2 and 4 (3.2 ± 0.7).

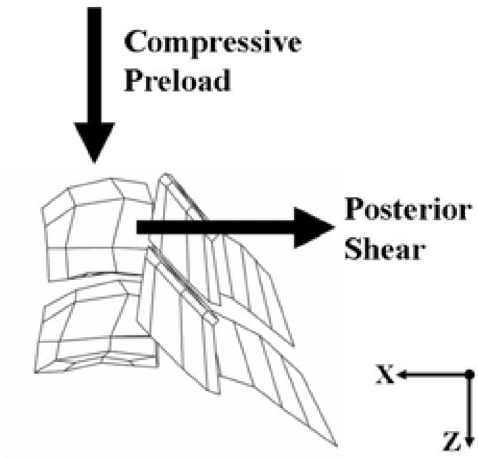


Figure A3.2 Loading of the C5/C6 motion segment used by Siegmund

The four axial preloads were based on four potential loading conditions: a preload of 0 N was used for comparison to previous research; a preload of 45 N was used to represent a static head mass; a preload of 200 N was used to represent a combination of static head mass and an inertial neck compression which develops due to a torso-seat-back interaction in low-speed rear-end collision at 8 km/h; and a preload of 325 N was used to represent the potential combination of static head mass, inertial compression and reflexive muscle forces.

The maximum extension moment (1.15 ± 0.12 Nm) at the level of the intervertebral disc in the current study was less than the extension moments (6 and 10 Nm) calculated for the atlantooccipital joint in human subjects exposed to speed changes of 8 km/h without head restraint protection. A significant portion of that moment, however, is supported by the neck musculature.

A3.2.2 Results and Discussion

Intervertebral extension angles of about 2 and 5 degrees have been reported for the C3-4 and C5-6 joints respectively in a single human subject undergoing a speed change of 8 km/h without a head restraint. The extension angles observed in the current study were 2.9 ± 1.8 degrees for C3-4 and 3.4 ± 2.2 degrees for C5-6 and were less than the extension observed in dynamic ligamentous spine tests but in the range of those observed in the human subjects. The similar intervertebral extension angles observed in both the human subject data and the present study further suggest that moment applied through the disc of the specimens in the current study was representative of the moments transferred through the intervertebral joints during actual whiplash loading conditions.

The results of these present tests demonstrated that posterior translation and extension of the superior vertebral body and maximum principal strain in the facet capsular ligament all increased with applied shear load. Changes in applied axial compression produced some reduction in posterior translation and extension, though these reductions were eliminated when two specimens that may have sustained ligamentous micro-damage were removed from the analysis. Flexibility coefficients calculated from the force-displacement and force-rotation data also did not vary with axial compression.

Based on the mean values, the maximum principal strains observed in the facet capsular ligaments during the flexibility tests were significantly less than the maximum principal strains observed in the corresponding failure tests. This suggests that on average, the facet capsule is not injured by the vertebral motions that occur during a whiplash exposure. However, two specimens exceeded their sub-catastrophic failure strains under the whiplash-like loads used in the flexibility tests, suggesting that capsular ligament fibres within these specimens may likely have experienced sub-catastrophic failures. Given the presence of nociceptive nerve endings in the cervical facet capsule ligament, such failures provide a mechanical hypothesis for the development of pain observed in a clinical population previously exposed to whiplash loading. Assuming the ratio of flexibility strains to initial failure strains is normally distributed, approximately 7.3 percent of specimens exposed to these spinal loads will undergo sub-catastrophic failure of a facet capsular ligament.

Inter-specimen variability in the maximum principal strains was larger for the sub-catastrophic failure tests (SD = 0.21) than for the flexibility tests (SD = 0.059). This suggests that different individuals under similar loading conditions might possibly produce different clinical outcomes. Moreover, the relative insensitivity of maximum principal strain to increasing posterior shear suggests that even relatively low posterior shear loads could exceed the threshold for sub-catastrophic failure in some individuals.

Compared to other previous studies, the increase in maximum principal strains (0.168 ± 0.059 under the combination of 135 N shear and 1.15 Nm extension moment at the intervertebral disc) observed in the current data suggest that shear contributed to increasing the strain in the facet capsular ligaments of motion segments loaded in extension. Furthermore, combining shear, extension and axial pre-torque may further elevate the strain in the facet capsule and further increase the potential for capsular injury. Increased capsular strain in the pre-torqued configuration would be consistent with the increased symptom duration observed in patients whose heads were turned prior to a rear-end collision.

Axial compression had no clear effect on the intervertebral kinematics, despite Yang et al. (1997) have previously reported increased flexibility to horizontal shear with increasing compressive loads. Methodological differences between the two studies were used as the reasoning for this discrepancy.

A3.3 Winkelstein et al. (2000)

A3.3.1 Test Method

The purpose of this study was to examine the role of the cervical facet capsule in whiplash injury. Six unembalmed human cadaveric cervical spines were cleaned of musculature and the C3-C4 and C5-C6 motion segments isolated for mechanical testing. Careful dissection was performed to expose the right capsular ligament. For the motion segment levels tested in this study the casting orientation relative to horizontal were 9 degrees for C3-C4 and 17 degrees for C5-C6.

Bending tests were performed in a pure-moment test frame equipped with a six-axis load cell. Flexion-extension testing was performed for each specimen in three loading configurations: 1) In the absence of a pre-torque, 2) with an axial pre-torque directing the upper vertebra toward the contralateral facet (away from the right facet joint), and 3) with the pre-torque directing the upper vertebra ipsilaterally (toward the right facet joint). A nominal pre-torque of 1 Nm, 5 % of the axial torque to failure, was applied to the specimens in the study.

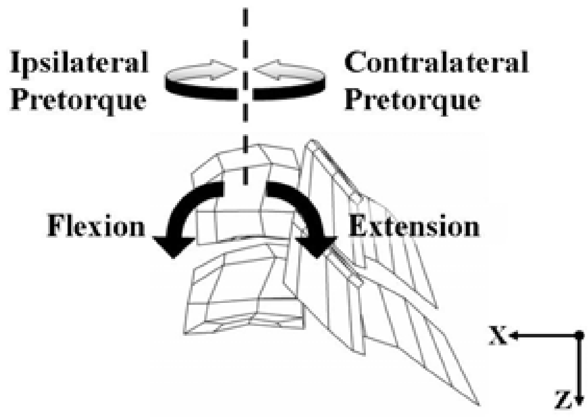


Figure A3.3 Loading directions of the C5/C6 motion segment used by Winkelstein et al. (2000)

Elongation-to-failure tests were performed using the isolated right capsular ligament. Facet joints were removed *en bloc* from the motion segments, and the bony articular processes superior and inferior to the facet joint were cast in polyester resin. Capsular ligament failure was defined as a decrease in force with a continued increase in joint distraction. Catastrophic failure of the facet joint was defined as the peak force before rupture.

A3.3.2 Results and Discussion

Mean flexion-extension range of motion was 11.6 ± 2.0 degrees and 9.8 ± 1.2 degrees for C3-C4 and C5-C6 respectively. Applied pre-torques resulted in a mean torsional rotation of 1.9 ± 1.0 degrees for all specimens. Mean flexion-extension range of motion after a pre-torque increased to 14.1 ± 3.7 degrees for the C3-C4 motion segments and to 10.3 ± 2.2 degrees for C5-C6. The increases in range of motion after a pre-torque were not statistically significant.

Axial pre-torque increased facet capsular strains for all loading scenarios. Catastrophic joint failure occurred at a mean failure force of 94.3 ± 44.4 N and 82.5 ± 33.0 N for C3-C4 and C5-C6 respectively. The mean actuator deflection at failure was 5.1 ± 0.73 mm for C3-C4 and 6.4 ± 0.66 mm for C5-C6. The site of failure was observed in the mid substance of the ligament for all specimens except for two. 7 of the 12 specimens tested had structural failure after subcatastrophic ligament failures.

During failure testing, mean maximum capsular strain at the subcatastrophic failure was 64.6 % ± 73.8 %. Catastrophic failure of the ligament occurred at 103.6 % ± 80.9 %

strain, four times the maximum strains during bending. Capsular strain distributions were highly non-uniform across the surface of the facet capsule during bending, illustrating the individual response differences to comparatively similar joint motions. In addition, the location of maximum strain did not exhibit any trend when specimen responses were compared.

Principal strain directions were commonly oriented across the joint, with the principal shear strains primarily directed along the joint line. This direction has been associated with the local sliding of the cervical bony facet surfaces during sagittal plane bending.

No overt pinching of the capsular ligament was observed in this study.

Capsular strains observed during bending, for all test configurations, were less than those required to produce catastrophic injury of the joint. It is therefore unlikely that vehicle occupants undergo gross failure of the capsular ligament resulting from low-speed, rear-end collisions. However, when subcatastrophic ligament failures are considered, the distinction is less clear. Capsular strains were significantly less in bending for the neutral and ipsilateral pre-torque groups than the subcatastrophic ligament failures. The contralateral facet strain was less but not significantly less than the strain to subcatastrophic failure. This finding indicates that some portion of the automobile occupant population may develop strains during bending with a pre-torque that are not different in magnitude from those required to cause subcatastrophic injury of the ligament.

A3.4 References

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APPENDIX A4 C5/C6 MOTION SEGMENT MODEL MADYMO DATASET

This Excel spreadsheet is in the attached CD as it is too large to present in hard copy.

APPENDIX A5 ACCIDENT RECONSTRUCTION MODELLING RESULTS

This Excel spreadsheet is in the attached CD as it is too large to present in hard copy.