Low Severity Neck Injury from Side Impact

Simone Lewis

School of Engineering and Sciences

Victoria University

Submitted in fulfillment of the requirements of the degree of Doctor of Philosophy

February 2018

Abstract

Typically, whiplash (low severity neck injury of for short LSNI) is associated with rear impacts. Due to this, there is a wide body of research investigating the mechanism of LSNI as a result of rear impact. Detailed studies into the prevalence of low severity neck injury show that this injury also occurs in front and side impacts (Stryke et al. 2012, Teamming et al., 1998, Jakobsson 1998, Morris et al., 1996, v Koch et. al. 1995,). This thesis is an investigation into low severity neck injury resulting from side impacts.

An initial investigation into the Monash University Accident Research Centre database (Australia) and the Loughborough University Co-operative Crash Investigation Study (UK) was undertaken to identify the typical factors associated with LSNI from side impacts. These factors were then used to determine the real-world cases to be reconstructed later in the thesis. As the occupants involved in side impacts are exposed to large lateral accelerations that do not occur in rear impacts, the factors that are associated with LSNI from rear impact cannot automatically be assumed to be a relevant in side impacts. This thesis makes a contribution to research by identifying the factors associated with LSNI that are unique to this side impact. This work can guide future research into the development of human surrogates/ human models to ensure that they more accurately replicate side impacts at multiple impact angles (such as oblique angles), as this thesis shows that LSNI occurs at various side impact angles and not just at 90 degrees.

A MADYMO human body model with detailed neck was used to simulate driver response in side impacts. To verify the output from the model, results from post mortem human subjects (PMHS) and live human volunteers, were used for comparison. The impact angles investigated in these trials were 90 degree lateral and 45 degree oblique. Six real world crashes were reconstructed using computer simulations undertaken in MADYMO (Mathematical Dynamic Modelling) and HVE (Human, Vehicle, Environment). Three different impact angles were analysed, namely 90 degree lateral near side, 90 degree lateral far side and 45 degree oblique near side. For each impact angle, two cases were reconstructed, one with an occupant receiving a low severity neck injury, and the other was a control case where the occupant did not receive a low severity neck injury.

The methodology used in this thesis of determining crash acceleration (crash pulse) by using HVE has been used previously by researchers (Franklyn et al. 2003, 2005a, 2005b and Hasija et al. 2007, 2009) to investigate head injury. Hasija et al. (2007, 2009), when investigating head injury also programmed the determined the crash pulse (from HVE) into a MADYMO to reconstruct the occupant mechanics using a crash test dummy model.

Declaration

Doctor of Philosophy by Publication Declaration "I, Simone Lewis, declare that the PhD thesis by Publication entitled Low Severity Neck Injury from Side Impact is no more than 100,000 words in length including quotes and exclusive of tables, figures, appendices, bibliography, references and footnotes. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. Except where otherwise indicated, this thesis is my own work".

Signature [redacted] Date 27 February 2018

Acknowledgements

The completion of this research is a tribute to the commitment and support of Dr. Eren Semercigil and Professor Gregory Baxter.

I would like to make special thanks to Professor Brian Fildes and Dr. Laurie Sparke for the opportunity to have been able to undertake this piece of work and for the valuable experience it has provided me.

The financial and technical support provided by General Motors Holden is gratefully acknowledged.

I am grateful for the support and encouragement from family and friends throughout the duration of this piece of work.

Simone Lewis.

Table of Contents

List of Figures

List of Tables

Chapter I
Introduction

1.1 Background

Low severity neck injury (also often referred to as whiplash) describes soft tissue injuries of the neck. These injuries occur when there is rapid acceleration of the head, such as occurs in motor vehicle accidents or sporting situations. For the purpose of this investigation the definition of Low Severity Neck Injury, is that classified as a level 1 neck injury on the Abbreviated Injury Scale (AAAM 1985). These injuries are the lowest classified neck injury on the scale and are described as “neck strain, acute with no fracture or dislocation” (AAAM 1985). In spite of the low severity rating, these neck injuries can be life altering, disabling injuries. As a result, low severity neck injury is a great cost to the community through insurance claims, rehabilitation costs and loss of income. Nolet et al. (2010) surveyed individuals reporting ongoing neck pain and found that 13% had experienced a motor vehicle collision in their history.

Most commonly, low severity neck injuries (LSNI) are associated with rear impacts. However, this thesis confirms that the injury occurs often enough in side impacts to warrant future research and development to understand the injury at this impact direction. Extensive research has been undertaken to determine the mechanism of injury and the factors associated with low severity neck injury from rear impact. Past research into LSNI from rear impact has not identifying conclusively the link between the injury pathology and the biomechanics of the neck during impact. In spite of this, there a number of measures that have been developed to test and reduce the risk of an occupant receiving a LSNI from a rear impact. These developments include: measurable injury criterion, New Car Assessment Program tests, improved biofidelic crash test dummy necks such as BIORID (Davidsson et al. 1998, Welch et al. 2010) and for risk reduction “anti-whiplash preventative measures” in seat and head restraint designs (Kullgren et al. 2007).

More recently, injury rate data presented in the literature has highlighted the need to investigate LSNI from different impact directions (Styrke et al. 2012, Teamming et al. 1998, Jakobsson 1998, Morris et al., 1996, von Koch et al. 1995). In this thesis, low severity neck injury resulting from side impact is investigated in detail. This thesis takes the approach of investigating real world crashes and identifying common factors amongst side impact cases where a LSNI occurred. A sample of such cases have been further investigated in detail using computer reconstructions. This approach is unique, as most investigations to determine occupant motion and injury from side impact have used laboratory based experiments (Mc Intosh 2007, Kallieris et al. 1981, 1990, 1991, 1996, Horsch et al. 1979, Faerber 1982, Bendjalla et al. 1987). These experiments have been mostly undertaken at an impact direction of 90° lateral impact angles, with limited research available to draw on at 60° and 45° impact angles.

Currently there is limited understanding of LSNI resulting from side impact. As a result of this, the research in lateral neck injury is far behind that of rear impact and there are no measures to test or reduce the risk of an occupant receiving a LSNI from a side impact. In this thesis, a detailed base of understanding about LSNI is provided as an enabling base for future development of vehicle safety measures and test criteria that may reduce the occurrence of injury in side impact.

Since its identification in 1926, the body of literature regarding neck injury from low acceleration motor vehicle crashes has expanded greatly. However, despite this volume, researchers have established few conclusions about the nature of this injury. The only concrete conclusion within the research community about these injuries is that the term *whiplash* is no longer suitable. Often the terms used are "deceleration injury", "whiplash syndrome", "cervical spine distortion" and "cervical spine trauma"(Teamming et al. 1998). For the purpose of this thesis the neck injury formally termed “whiplash” will be referred to a low severity neck injury. For the purpose of this research, the term *low severity neck injury* will replace the term *whiplash.*

There are different ways to describe side impact impacts, typically laboratory tests use a measure in degrees to identify the impact direction. In real world crashed and crash investigation that relies of measures such as the CDC (Collision Deformation Classification) direction of impact is identified using a clock face. As this research relies on both previous laboratory research as well as measurements and data collection from crash investigators both different descriptors are used where relevant. Figure 1.1 outlines the relationship between the different measurements.

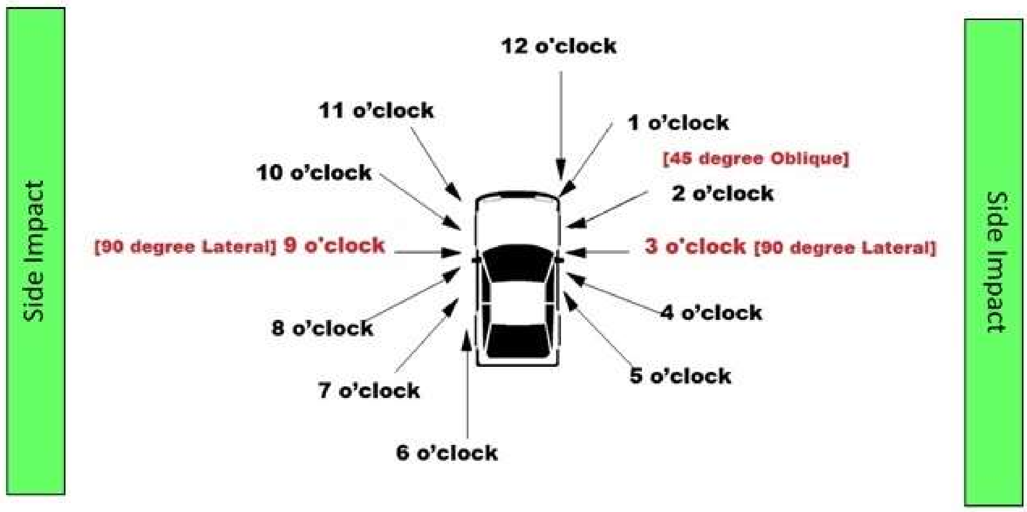


Figure 1.1 Descriptors of direction of side impact crashes

1.2 Research Objectives

Currently there is little research that focuses solely on low severity neck injury from side impact. The purpose of this work is to;

a) Address the gap in research that investigates LSNI from side impact to begin a base of understanding regarding the injury at this impact angle (such as incidence of injury, human factors, crash characteristics and occupant mechanics).

b) To undertake a detailed analysis of low severity neck injury from side impact, so as to determine the unique factors associated with this injury at this impact angle.

c) To undertake an analysis of occupant mechanics using computer simulations of real-world side impacts that have been identified as exhibiting common factors associated with LSNI. This work will be undertaken using previously validated mathematical models of the human body with an included detailed neck model, used as a combined human model.

1.3 Thesis Structure

This thesis is made up of 7 chapters. The first two chapters are an introduction and a review of the literature. Due to the volume of literature available regarding rear impact neck injury, a summary of literature has been presented in this thesis showing the common themes suggested by authors as contributing to LSNI during rear impact. This body of work stands as a comparison to the findings, regarding the investigation into LSNI from side impact presented later in the thesis. A research framework has been presented in Chapter 3, to provide an overview of the methodologies used in the thesis. A detailed methodology has been presented with each of the individual investigations undertaken in the thesis.

There are three aspects to the investigation undertaken into LSNI. The first part is an analysis of real-world crashes to determine the prevalence of LSNI in side impact and to identify the unique factors within these crash cases for later reconstruction. Analysis of side impacts from two crash databases, the Monash University Accident Research Centre database (Australia) and the Loughborough University Co-operative Crash Injury Study (UK) was is presented in Chapter 4 and includes a methodology that describes how the real-world cases were selected and analysed. The results presented include general human factors such as age and gender, vehicle factors such as vehicle mass and factors associated with other injuries received by the occupants such as head injuries. This chapter provides a foundation for the following two aspects of this research, as well as pointing the way for future research into side impact.

In the second and third aspects of the investigation, the TNO, Nederlandse Organisatie voor Toegepast Natuurwetenschappelijk Onderzoek, (*Netherlands Organization for Applied Scientific Research*) human body model was combined with detailed neck model to reconstruct human motion in lateral and side impacts. This model was chosen as it is a validated computational model that has been used in previous research (Van der Horst et al., 2001). The use of the computational human model with detailed neck (TNO Netherlands) allowed for an inexpensive, safe and ethical method to investigate low severity neck injury in side impacts. This model as a combined model was chosen as it offers the benefit of being multi-directional, as well as allowing for variable muscle activity. This capability permitted for muscle activity to be simulated in the reconstructions of the driver's head and neck kinematics during impact. For the purpose of this thesis, the human body model with detailed neck will be referred to as the combined human model.

In Chapter 5, a validation of the combined head and neck with the human body model is undertaken to supplement the existing validations of the model. The output of the model with passive muscle settings is compared against a 90° lateral cadaver tests undertaken by Wayne State University (Cavanaugh et al. 1990 1993, Irwin et al. 1993). To also compare the model's reliability in replicating human volunteers, a simulation including muscle activity in the neck was compared to 90° lateral and 45° oblique volunteer test undertaken by the Naval Bio-dynamics Laboratory (Wiseman et al. 1986). The kinematic data shows a similar event occurring in the model trial compared to the experimental data.

In the third and final part of the investigation (presented in Chapter 6) the combined human body model is used to reconstruct the occupant kinematics of three real world side impact crashes identified with the common factors associated with LSNI that were reported in Chapter 4. The cases are at 3 different impact angles; 90° lateral near-side, 90° lateral far-side and 45° oblique near-side. The results of these cases were compared to three control cases where the occupant reported that they did not receive a neck injury. Each of these cases have been reconstructed using two computer programs. Initially, the cases are reconstructed using Human, Vehicle, Environment (HVE) to obtain the vehicle's acceleration. The acceleration generated from HVE was then used in MADYMO 5.4.1 with the combined human model in the crash simulations. Previous research validating the HVE software is also presented in this chapter. The kinematic output from the cases explored and the lateral head and first thoracic acceleration are compared in for each case. In the 45° oblique near-side case, the frontal head and first thoracic acceleration are compared. This chapter provides a unique in-depth analysis of driver head and neck kinematics in real world crashes resulting in low severity neck injury, together with recommendations for future work.

The investigation undertaken in this thesis is discussed and concluded in Chapter 7, with five appendices following the conclusions. Appendix 1 provides an overview of the anatomy of the cervical spine. This has been included to provide a base of understanding of this anatomical region. This appendix also provides a definition of some of the terms describing motion used throughout this piece of research. The second appendix provides an explanation of the crash deformation classification, this is to support the use of this classification in Chapter 5 and Chapter 6. The co-ordinate system for the combined human model has been presented in Appendix 3. This has been provided to support the work undertaken with the human body model with detailed neck undertaken in Chapters 5 and 6. Appendix 4 contains the sled accelerations for the Naval Bio-dynamic Laboratory sled tests used for comparison against the model in Chapter 5. The final, Appendix 5 presents the kinematic data that has been presented in Chapter 4 with an additional row of kinematic data, the model presented with the facet skin layer presented as transparent to show additional detail of the neck during impact.

Chapter II
Literature Review

2.1 Introduction

There is a broad base of understanding into low severity neck injuries from rear impacts. The research into rear impacts is vast, ranging from identifying the incidence of injury through insurance and hospital data through to volunteer trials, cadaver specimen tests, specially developed crash test dummy necks and sophisticated neck models. In spite of this, body of work, the mechanism of low severity neck injury (LSNI) from rear impacts is not definitive (Chen et al. 2009). When investigating low severity neck injury from different crash directions, such as side impact, it is difficult to determine how much of this body of work can be relied upon to determine what advances need to be made to reduce low severity neck injury from side impacts.

The key difference between rear and side impacts is the large lateral component in side impact at all crash angles. This results in there being great variation in how an occupant interacts with the interior of a vehicle including the seat and lap sash belt when comparing a rear impact and a side impact. This would suggest that any advances in rear impact research that have resulted in modified seat, head restraint, dummy designs, new car assessment tests and computational neck models may not provide any benefit in neck injury mitigation during side impacts. How side impacts influence the biomechanics of the neck and the relationship with low severity neck injury has not been researched before.

This chapter provides a review of previous literature regarding the common factors associated with low severity neck injury from all impact angles. This is followed by a review of research into the biomechanics of the head and neck during rear, side and front impacts and where reported the proposed cause of LSNI. All of the factors discussed in this literature review were chosen as they are the same factors analysed from the real world cases presented in the later sections of this thesis. The factors examined are direction of impact, gender, age, seating position, side and direction of impact, striking object, seatbelt use, airbag deployment and delta V. The other injuries received by occupants with a low severity neck injury have also been presented in detail. Head restraint positioning and the occupants initial head position during impact have not been investigated although the literature regarding these factors were presented for completeness.

2.2 Factors Associated with Low Severity Neck Injuries

The factors associated with low severity neck injuries have been investigated extensively for many decades, yet much of this research focuses solely on rear impacts. What is accepted by researchers about rear impacts resulting in LSNI is that there are a number of common factors of those crashes and the occupants involved in those crashes. These factors have been presented in this are later compared to the findings from an investigation (presented in Chapter 3), that investigates the common factors found in real world side impacts where an occupant received a low severity neck injury (Abbreviated Injury Score, AIS 1, AAAM 1985) This section presents research focusing on the factors direction of impact, gender, age, seating position, side and direction of impact, striking object, seatbelt use, airbag deployment, delta V (rate of change of impact velocity) and occupant injuries received in real world side impacts.

2.2.1 Direction of Impact

The weight of this piece of research relies on low severity neck injury occurring frequently enough in side impacts to warrant concern. The literature (Styrke et al. 2012, Teamming et al. 1998, Jakobsson 1998, Morris et al., 1996, von Koch et al. 1995,) reports that there is a high percentage of low severity neck injury from side impacts. Styrke et al. (2012) analysed the data of individuals presenting to the emergency department of Umeå Sweeden. An analysis of longitudinal data from 2000 to 2009 showed that LSNI has increased at a rate of approximately 1% per annum. What is most interesting about the findings of Styrke et al. (2012) in regard to this piece of research is that in spite of the 1% increase for all crash directions, the incidence from rear impacts has decreased annually. It's acknowledged by Styrke et al. (2012) that whiplash prevention systems have been designed to be most effective in rear impacts. Figure 2.1 Shows that the increases in incidence of LSNI is contributed to from an increase in side (left and right) and front impacts.

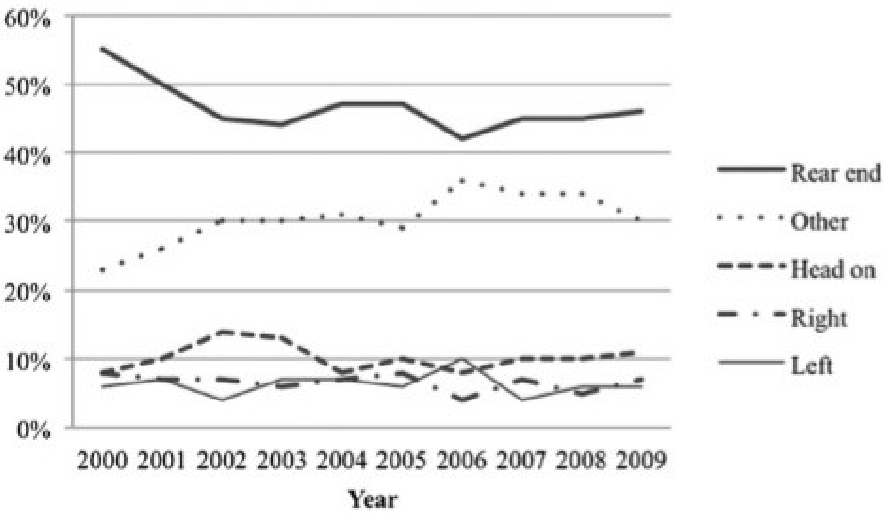


Figure 2.1: Direction of Impact in Car Crashes (Styrke et al. 2012)

The proportion of low severity neck injuries from four international accident databases separated for each impact type (Teamming et al., 1998, Jakobsson 1998, Morris et al., 1996, von Koch et al. 1995,) are presented in Table 2.1. The databases reported are the Volkswagen accident database (German), the Volvo accident database (Sweden), Co-operative Crash Injury Study (United Kingdom) and Folksam Insurance data (Sweden).

Table 2.1: Percentage of low severity neck injury by impact type.

	Volkwagen Database (Teamming et al., 1998)	**Volvo Database** (Jakobsson et al., 1998)	**Co-operative Crash Injury Study** (adapted from Morris et al. 1996)	**Folksam Insurance Data** (von. Koch et al., 1995)
Front	38%	35%	55%	23%
Rear	16%	17%	13%	63.7%
Side	12%	16%	25%	8.8%
Rollover	1%	7%	5%	2.6%
Multiple	33%	21%	NA	NA
Other/ Unknown	NA	4%	2%	1.8%

The data illustrates that low severity neck injury occurs in all impact directions at a varying frequency. The four databases report low severity neck injury occurring from side impacts to be between 8.8% and 25%. The first three databases (Volkswagen, Volvo and CCIS) show that low severity neck injury occurs mostly from frontal impacts, between 35% and 55% (Teamming et al., 1998, Jakobsson 1998, Morris et al., 1996). The Folksam Insurance data shows that most occupants (63.7%) with a low severity neck injury are involved in a rear impact (von Koch et al. 1995). The higher values for frontal impacts in the CCIS, and for rear impact in the Folksam Insurance data may be explained by these databases not including multiple impacts, as the absence of this category would increase the percentage values in the other categories. Jakobsson et al. (2000, Volvo data) found that 21% of occupants in frontal impacts and 26% occupants in side impacts received a low severity neck injury. It is important to note that all of the data available about the incidence of low severity neck injury side impact comes from databases, these databases are not a full representation of the whole population in each country they are sampled from.

The rate of low severity neck injury in Australia for each impact angle has been presented by Fildes et al. (1995) in Figure 2.2. The data includes an additional variable, severity of injury. This data compares non- chronic cases to chronic cases for each impact direction. Fildes et al. (1995) defined chronic neck injury as an injury persisting for greater than 6 months. It is shown that low severity neck injury has a higher total incidence from rear 28% and front impacts 21% compared to side impacts (total 16%). Although, it is shown that rear (20%) and side impacts (10% left and right combined) are more likely to result in chronic neck injury compared to front impacts (7%).

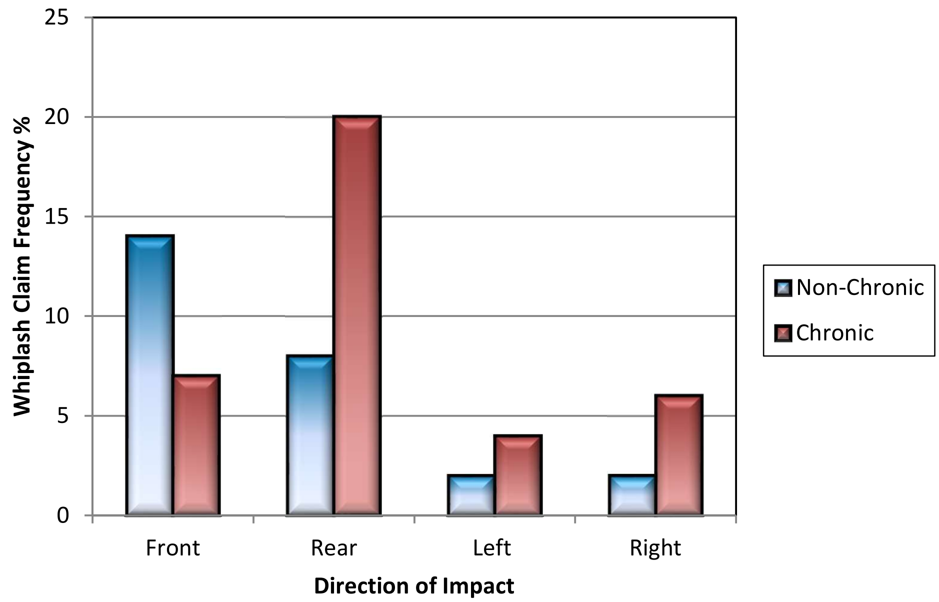


Figure 2.2: Low severity neck injury by direction of impact- Australian data (Fildes et al., 1995).

2.2.2 Gender

Gender has been widely discussed as one of the most important influencing factors with regard to low severity neck injury. The literature consistently shows that females are more likely to receive a low severity neck injury than males (Nolet et al. 2010, Teamming et al., 1998, Morris et al., 1996, von Koch et al. 1995, Fildes et al. 1995, Gibson et al. 2000 and Bring et al. 1996). Table 2.2 contains data from the Volkswagen accident database (Teamming et al., 1998) illustrates that although males experience more vehicle impacts, females have twice the risk of males. Minton et al. (1997) also found that males are less likely to experience disability than females as a result of a low severity neck injury. This finding is supported by Nolet et al. (2010) who found that individuals with a long history of neck pain, has experienced a motor vehicle collision in their past.

Table 2.2: **Low severity neck injury and occupant gender, (Teamming et al., 1998).**

	All Occupants	Male	Female	Male/Female Relationship
Occupants involved:	12 011	7321	4418	1/0.60
Occupants with LSNI (AIS 1 Neck Injury)	1620	735	884	1/1.20
Injury Risk	13.5%	10%	20%	1/2

Gibson et al. (2000) also reported a rate of chronic neck injury by type of impact presented in Table 2.3. This data was sourced from matching police accident data to data in a clinical database resulting in a sample size of 273. This table shows different results to Fildes et al. (1995) that was previously presented in Figure 2.1 with rear (40%) and frontal impact (34%) as the most common directions of impact to result in chronic low severity neck injury compare to side impact (22% left and right combined). The data presented by Gibson et al. (2000) has been further broken down to show the distribution of gender. While the total number of occupants with a chronic injury show that more females are injured than males, when the data is broken down into impact angles males have a higher representation at some angles. In rear (male 46%: female 35%) and left side impacts (male 12%: female 10%) males receive more chronic injury in front impacts. In right side impacts both genders experience a similar rate of chronic injury at 12%. In front (males 29%: females 37%) and rear (males 6%: females 0%) impact females experience more chronic injury than males.

Table 2.3: Impact type and gender for chronic low severity neck injury cases, Gibson et al. (2000).

Impact Direction	Total Percentage of Occupants	Percentage of Female Occupants	Percentage of Male Occupants
Front	34%	37%	29%
Left	11%	10%	12%
Rear	40%	35%	46%
Right	12%	12%	12%
Rollover	3%	6%	0%

Recent analysis of 10 years of Emergency Department data in Umeå Sweeden by Styrke et al. (2012) shows that there is less differentiation between the genders with 51.9% of women and 48.1% of men presenting with LSNI.

2.2.3 Age

Previous research into the human factors associated with low severity neck injury shows that the injury is likely to occur to adult occupants more than children and adolescents (Lövsund et al. 1988, Teamming et al. 1998, Schuller et al. (1999). The average age for males with a low severity neck injury from a frontal impact is 33 years, for females it is 35 years. Morris et al., (1996). There is little difference between these values, and the average age of occupants with no low severity neck injury in a frontal impact (males 32.9 years and females 36.3 years). For rear impacts, the average age of males is 41.5 years and that of females 36.0 for receiving injury. Males with no low severity neck injury from rear impacts are on average younger. Females with no low severity neck injury from rear impacts are on average older. Figure 2.3 illustrates the distribution of age separated by gender for the occupants in the Volkswagen database (Teamming et al. 1998).

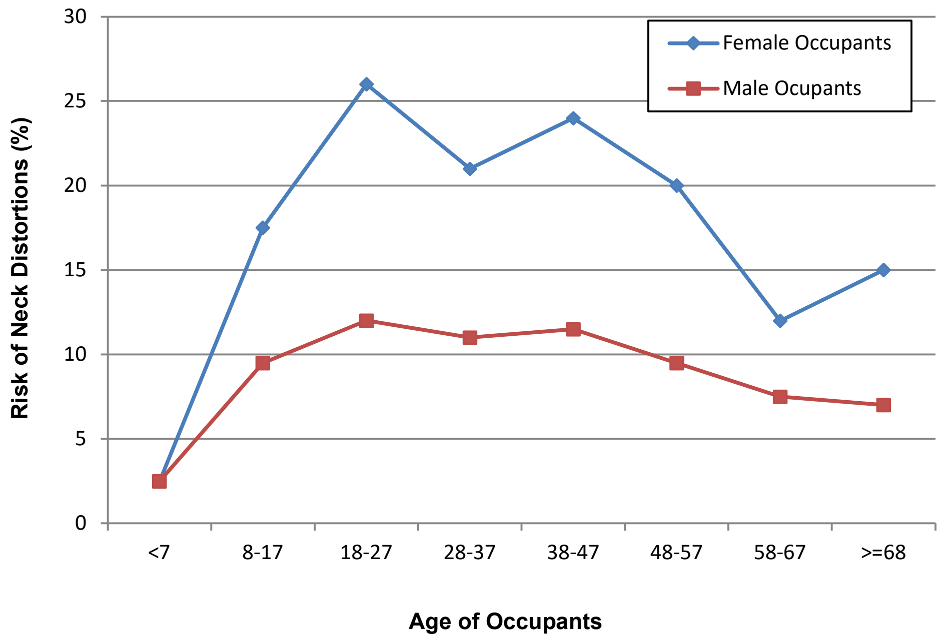


Figure 2.3: Age of occupants with low severity neck injury (Teamming et al. 1998).

A similar trend in incidence of LSNI and age that has been presented in figure 2.3 has been found by Styrke et al. (2012). Longitudinal data shows that the age category with the highest incidence of low severity neck injury (for all impact angles) is 20- 24 years.

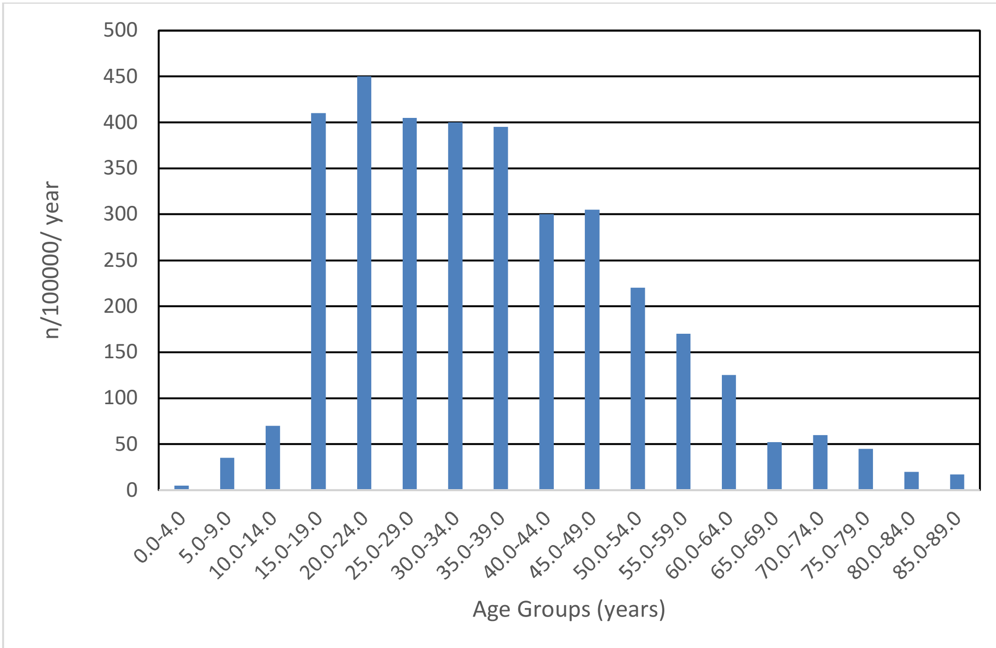


Figure 2.4 Average Incidence of LSNI for different age groups Styrke (2012)

2.2.4 Occupant Seating Position in Vehicle

The seating position in the vehicle for occupants' receiving a low severity neck injury has been reported by Lösvund et al. (1988), Galasko et al. (1993) Styrke et al. (2012) and Jonsson et al. (2013). Lösvund et al. (1988) shows that front row occupants (driver and front passenger) are at a greater risk of receiving a low severity neck injury than a rear seated occupants. Galasko et al. (1993) compared the incidence of low severity neck injury between drivers and front passengers (Table 2.4) and found the risk of front row occupants receiving a low severity neck injury is dependent on type of impact. The data in Table 4 highlights the little difference in the rate of injury between driver and passenger for each impact direction. Rear impacts produced the most injury with 51.9% of drivers and 54.3 % of passengers. Frontal impacts injured 22.7% of drivers and 21.3% of front passengers. Side impacts injured 16.4% of drivers and 12.2% of passengers.

Styrke et al. (2012) found that of those presenting at the Emergency Department with LSNI who were occupants in a car, 71.6% were drivers, 19.9% were front-seat passengers and 7.9% were rear seat passengers with 0.5% of seat positions unknown.

Table 2.4: Percentage of low severity neck injury from each impact type for front seat occupants Galasko et al. (1993).

	Driver	Front Passenger
Front	27.2	21.3
Rear	51.9	54.3
Side	16.4	12.2
Unknown	4.5	12.2

Jakobsson et al. (2000) found that that the risk of injury for each of the seating positions also differs between the genders. Male rear seat passengers and drivers have a greater incidence of low severity neck injury than that of front passengers. In addition, female front seat and rear seat passengers are at the greater risk of low severity neck injury than female drivers. In front, impacts the risk for males in all of the seating positions is equal. However, female drivers are reported to be at a greater risk than both the front and rear seat passengers.

Jonsson et al. (2013) investigated the incidence of risk of males and females receiving a LSNI in a rear impact with respect to their position in the vehicle as front seat passengers or drivers. Insurance claim data from Folksam Sweeden between 1990-1999, was used to analyse rear impacts when occupants in both front seats where on received a medical impairment from a neck injury. The medical impairment was required to be present 12 months following the crash and was required to have been diagnosed by a doctor. Jonsson et al. (2013) found that both gender and seating position influenced the occupant's likelihood of receiving a LSNI with symptoms presenting for a year or more. When comparing the incidence of LSNI of drivers compared to their front seat passenger, it was found that drivers experience double the rate of injury.

2.2.5 Mass of Vehicle (striking car vs struck car)

An in-depth investigation by von Koch et al. (1995) was made into the relationship between the vehicle's mass and the likelihood of an occupant receiving a low severity neck injury. It is shown that the relationship between the striking vehicle and the struck vehicle has an influence of the risk of injury. The ratio of risk for occupants in striking vehicle (frontal impact) and the struck vehicle (rear impact) of different masses are presented in Table 2.5. The results show that when vehicle masses are equal, the risk of neck injury is 1.4 to 2.32 times as high for the occupants in the struck vehicle. The maximum risk Table 2.5 for occupants in the struck vehicle, is almost 4.2 times higher when the striking vehicle weighs 1300kg and the struck car weighs 900-1000kg. Even when the mass of the striking vehicle is less than the mass of the struck vehicle the risk of the occupants in the struck vehicle is 1.4 times higher. It is also reported by von Koch et al. (1995) that females are at more risk than males when they are in the struck vehicle. Males are at more risk in the striking vehicle when the striking vehicle weighs 300-400 kg less than the struck vehicle. Finally, von Koch et al. (1995) also found that vehicle mass is not the only vehicle factor that may contributes to injury, the shape of the acceleration pulse from the impact affects injury. This suggests that vehicle stiffness as well as mass is also a factor.

Table 2.5: Risk or injury and relationship of mass of striking vehicle compared to struck vehicle, von Koch et al. (1995).

	Striking Car					
Struck Car	**900kg**	**1000kg**	**1100kg**	**1200kg**	**1300kg**	**1400kg**
900kg	1.46	1.37	2.50	4.00	4.16	3.13
1000kg	1.43	1.61	2.37	3.03	4.17	3.21
1100kg	1.09	1.27	1.83	2.20	2.54	3.37
1200kg	0.83	1.02	2.27	2.32	2.25	2.44
1300kg	0.90	1.03	1.09	1.30	1.51	2.04
1400kg	0.60	1.05	0.98	1.59	1.48	1.81

Desapriya et al. (2004) investigated the injuries in crashes where one vehicle was a pick-up truck and the other vehicle was a passenger sedan using insurance data (from British Columbia). Injury data was analysed to see the influence that vehicle mass had on rate of injury. It was found when there was an impact between a pick-up truck and a passenger sedan, occupants in the passenger sedan were more than twice as likely to receive a whiplash injury. LSNI injuries recorded for passenger sedan occupants totalled 316 compared to 128 recorded for the occupants of the pick-up truck. Impact direction of crashes were not specified. These findings support that the greater mass of an impacting vehicle can increase the incidence of LSNI.

2.2.6 Angle of Impact

The risk of receiving the initial symptoms of a low severity neck injury in a side impact is shown to be influenced by the impact angle and the direction of impact (near or side). One way of describing direction of impact in side impacts is using position of numbers on a clock face. A diagram of the direction of impacts has been presented in Appendix 2 figure A2.1. The angle of impact for a sample of side impact cases that resulted in low severity neck injury have been presented in Table 2.6 (Jakobsson et al. 2000). Occupants involved in near side impacts at 7-8 o'clock (drivers and left rear passengers) and 4-5 o'clock (right front and right rear passengers) are shown to have the greatest risk of receiving a low severity neck injury. On the other-hand for 90° side impacts (3 o'clock and 9 o'clock) far side impacts pose a greater risk than near side impacts at the same angle. The angle of impact with the lowest risk of injury is far side 1-2 o'clock (drivers and left rear passengers) and 10-11 o'clock (right front and right rear passengers).

Table 2.6: Risk of initial symptoms based on angle of impact (Jakobsson et al. 2000)

Impact Direction	1-2 o'clock* 10-11 o'clock° Far Side	3 o'clock* 9 o'clock° Far Side	4-5 o'clock* 7-8 o'clock° Far side	7-8 o'clock* 4-5 o'clock° Near Side	9 o'clock* 9 o'clock° Near Side	10-11 o'clock* 1-2 o'clock° Near Side
Risk	9%	31%	25%	38%	25%	24%
Total no.	33	81	8	8	119	67

*Angle of impact for Swedish Drivers and Left Rear Passengers
° Angle of Impact for Right Front Passengers and Right Rear Passengers
% of occupants within the sample of that impact direction with initial symptoms of LSNI

2.2.7 Head Restraints

Since 1972, head restraints have been a compulsory part of vehicles in Australia, although their introduction did not decrease the rates of low severity neck injury (Maher, 2000). Literature has highlighted the ineffectiveness of standard head restraints. There are two major concerns with the current design of standard head restraints. The first concern is the overall effectiveness of the head restraint due to its dated design. Head restraints are designed to be effective in rear impact. The effectiveness of head restraints in a side impact has not been investigated. The second concern is the incorrect use of the head restraint with occupants who fail to adjust their restraints properly. It has been reported that 88% of US drivers, and US and UK passengers adjust their head restraints to be too low. In 24% of occupants investigated, the head restraint was too far from the head Cullen et al. (1996). The limitation of standard head restraints has led to new head restraint and seats designs to reduce the incidence low severity neck injury. These head restraints have mechanisms that are activated in an impact, with the goal of providing additional protection to the occupant primarily by reducing the relative motion between the occupant's head and torso (Trempel (2014), IIHS (2007), Lundell et al. (1998), Jakobsson et al. (1998, 2000), Sekizuka et al. (1998) Wirklund (1998), Wirklund and Larsson (1998).

Research suggests that active head restraints and energy absorbing seat designs are reducing low severity neck injuries in rear impacts. Insurance Institute of Highway Safety (IIHS, 2002) found that the SAAB Active Head Restraint (SAHR) reduced low severity neck injury claims by 43%. In addition to this, improved head restraint geometry reduced claims by 18% and the Volvo WHIPS seat reduced claims by 49%.

Later research undertaken by Insurance Institute of Highway Safety- IIHS (2007) investigated all improved head restraint and energy absorbing seat design and found that two out of three designs were marginal or poor. Of 75 vehicles (USA models) tested by measuring geometry and undertaking simulated crashes, 22 were rated good, the other 53 were rated marginal or poor. These findings suggest that there is still much work to be done in the area of neck injury prevention and head restraint and seat design. When investigating injury rate in rear impacts comparing head restraint ratings, the Trempel (2014) found that the injury rates were lower for occupants in vehicles with head restraints rated good compared to poor. The rate of reduction in risk was found to be 12.7% for females and 8.9% for males.

Kullgren et al. (2007) found that vehicles fitted with a protection system designed to reduce LSNI were effective lessening the risk of developing long term symptoms from LSNI after a rear impact by 50 %. Other research suggests that the protective measures of seats and head restraints may now reduce the risk of neck injury for all demographics of occupants. Linder et al. (2013) investigated the effectiveness of seat and head restraint design and using occupant models with female anthropometry. It was reported that in the cases where a female model was simulated the neck loading would result in a greater risk to that occupant compared to a male occupant in the same conditions. This finding is relevant to any future design measures proposed to address LSNI in side impact as these measures may need to be modified to address anthropometric differences in gender.

Dehner et al. (2013) investigated the relationship between neck muscle activity and head restraint function in 8 female subjects (age 19-27 years). Rear-end collision laboratory sled tests were conducted at 6.3 km/h. The tests were conducted using a seat fitted with a (non-active) standard head restraint. The interesting findings in this work with respect to head restraint and occupant interaction is that there is muscle function detected in the sternocleidomastoid muscle (median 81ms), before there was contact with the head restraint (median 84ms). This work highlights the importance of good head restraint positioning as a greater gap between the occupant's head and the head restraint would result in a longer period of muscle activity before the head restraint can support head deceleration.

Ziraknejad et al. (2014) outlines work to develop sensors, that can detect head position and lead to automatic repositioning of the head restraint. This technology differs from other Active Head Restraints (AHRS), that repositions upon the occurrence of a rear impact. This technology will work to auto-position the head restraint according to the Institute for Highway Safety (IIHS) head restraint positioning guidelines in rear impacts. The focus of this technology will be to eliminate the human error of manual positioning and will only be effective in rear impacts. This work is still in the developmental stage.

2.2.8 Seatbelt Use

It has been reported that the incidence of low severity neck injury increased after the compulsory introduction of seatbelts (von Koch et al. (1995), Galasko et al. (1993). Figure 2.5 illustrates the increasing incidence of injury following the introduction of seatbelts (marked with A). It is shown that the incidence individuals attending a UK hospital emergency department with LSNI was below 10% in 1983 prior to the introduction of seatbelts. This increased to a maximum of over 40% by 1991.

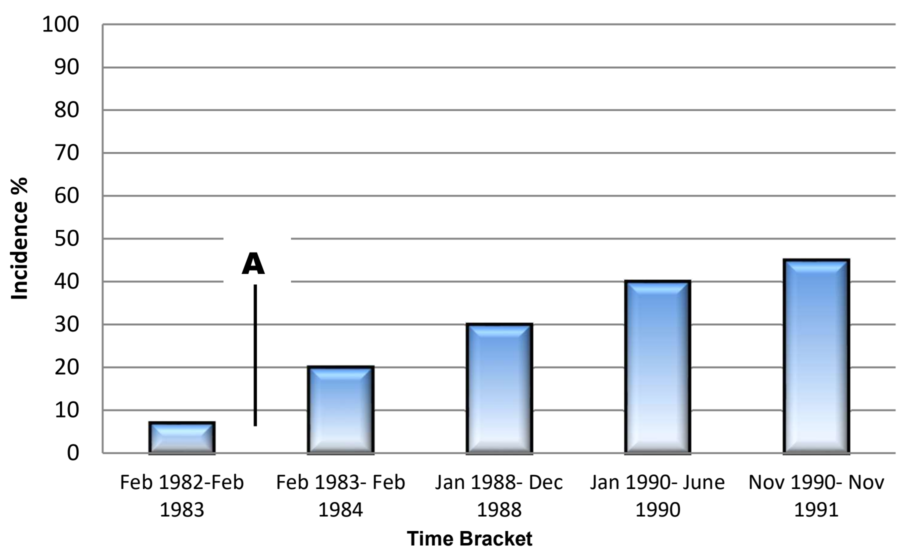


Figure 2.5: Incidence of low severity neck injury after the introduction of the seatbelt, Galasko et al. (1993).

von Koch et al. (1995) suggests that occupant interaction with the seatbelt is a common factor between front and rear impacts that result in low severity neck injury. Morris et al. (1996) found that occupants (regardless of gender) are significantly more likely to receive a low severity neck injury in a frontal impact if they are restrained by a seatbelt. With regard to rear impacts, only males are found to be significantly more likely to receive a neck injury if restrained by a seatbelt. While also considering that throughout the decades there has been an increase in vehicles on the roads and an increase in insurance claims, this data also suggests that the occupant kinematics are influenced adversely by interaction with the seatbelt.

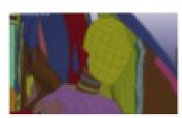
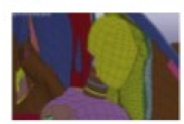
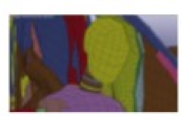
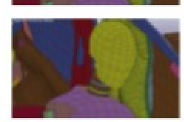
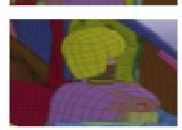
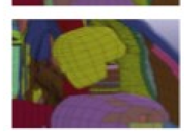
2.2.9 Airbag Deployment

There is limited literature available regarding the effectiveness of airbags in reducing low severity neck injury. Furthermore, what is available provides conflicting views with respect to the effectiveness of airbags to reduce low severity neck injury. An investigation undertaken by Fildes et al. (1995), comparing same model vehicles (VR Commodores , 1993 model) with and without airbag deployment, the only conclusion that could be made was that airbags did not have a negative influence on soft tissue or other neck injuries. Kullgren et al. (2000) found more conclusive results when examining frontal impacts where vehicles with airbags were also fitted with seatbelt that had pre-tensioners. The combination of seatbelt pre-tensioners and airbags were found to reduce the risk low severity neck injury by 41%. The benefit of pre-tensioners and airbags was found to increase in impacts with lower velocity. The risk of low severity neck injury in impacts with velocity of impacts between 1 and 30 km/h was reduced by 59% (Kullgren et al., 2000), compared to other impact speeds.

Otte (1995) has reported that frontal impacts with airbag deployment have a higher incidence of AIS1 cervical spine injuries. It is speculated that airbag inflation contributes to injury, as it induces a powerful motion of the head and cervical spine (Otte 1995). The side impact cases investigated in this study didn’t have as high incidence of AIS 1 neck injury as the front impacts. This may be because the side impacts investigated appear to be of a higher severity, indicated from the other injuries received by the occupants. Morris et al. (2000) found different results to Otte (1995) with the data showing that airbags reduce low severity neck injury in frontal impacts, by reducing head acceleration. For Australian drivers, soft tissue neck injury rates for non-airbag cases was 40% (compared to 15% for cases with airbags).

Recent research undertaken by Potula et.al (2012) found that in some simulated side impact trials where an occupant was out of position with a curtain airbag that there was greater risk of neck injury. Using Finite Element Analysis of a Hybrid III 50th percentile male in a 1996 Dodge Neon Vehicle with a curtain airbag. Four different occupant positions were investigated these were; in position with a normal seating posture with head forward, Out of Position 1 (OOP1) with the occupant that is laterally closer to the door without modification to the rotation of the head and neck, Out of Position 2 (OOP2) with the occupant laterally closer to the door and with laterally bending of the head and some head rotation and Out of Position 3 (OOP3) with the occupants torso rotated so the posterior of the head is against the door. Each of these body positions were tested in three different air bag configurations. One with no airbag and two with inflating airbags. The first airbag trial is labelled uniform pressure (UP) that simulates a situation where there is constant specific heat from the gas inflating the airbag. The other airbag trial is labelled Smooth Particle Hydrodynamics (SPH) that more realistically simulates the unfolding of the airbag due to the modelled particle dynamics. The contact point in each of the trails is presented in Table 2.7. The findings of Potula et.al (2012) were that both the UP and SPH modelling methods produced similar results. In all of the air bag cases the head injury criterion and the peak head accelerations were significantly lower than the no airbag trial. This supports that curtain airbags reduce the risk of head injury regardless of the occupant being in or out of position in an impact. The finding in this study that is most interesting regarding side impact and neck injury was that in two of the trials OOP2 and OOP3, the curtain airbag increased neck flexion forces, resulting in a greater risk of neck injury than if there was no airbag installed in the vehicle.

Table 2.7 Simulated Initial Position Airbag Neck Injury Potula et al. (2012)

	No Airbag	Uniform Pressure	Smooth Particle Hydrodynamics
In Position (~56ms)			
OOP1 (~50ms)			
OOP2(~52ms)			
OOP3(~40ms)			

<u>2.2.10 Speed, Delta V, Equivalent Barrier Speed (EBS), (EES).</u>

There are different measures available to quantify the severity of an impact. These include speed of struck or striking vehicle, delta V and equivalent barrier speed (EBS). Delta V is defined by the international standard 12353-1 for Road Vehicle accident analysis (ISO,

2002) is *the vectorial difference between impact velocity and separation velocity.* Knowing delta V provides an indication of the severity of the crash. Equivalent Barrier Speed is the speed at which the vehicle would be impacted with a test barrier to produce the same amount of crush (Tomasch 2004). The Energy Equivalent Speed (EES) as defined by International Standard (ISO/DIS 12353-1:1996(E)) is "the equivalent speed at which a particular vehicle would need to contact any fixed rigid object in order to dissipate the deformation energy corresponding to the observed vehicle residual crush" (Tomasch 2004).

Gibson et al. (2000) has reported the mean estimated speed for the struck vehicle for occupants with chronic low severity neck injury in various impact directions. The mean estimated speed was a subjective measure reliant on the experience of the police officer attending the accident scene. Of the cases investigated by Gibson et al. (2000), it was found that the struck vehicle was stationary in 34% of all cases. The estimated speed for the rear impact cases was 3 km/h, as most of the struck vehicles were shown to be stationary. The average estimated speed for frontal impacts was 55km/h. Side impacts were separated into left and right impacts. The average estimated speed for left side impacts was 45 km/h and for right side impacts is 42 km/h. This indicates that the struck vehicles were traveling a reasonable forward speed when struck and this forward motion would be expected to influence the occupant's kinematics. The average estimated speed for rollovers was 63 km/h. Hell et al. (1998) reports the most common estimated impacting speeds were between 16 and 25 km/h, followed by 26 and 35 km/h in crashes where an occupant received a Low severity neck injury. Very few impacts were slower than 16 km/h or above 35 km/h.

The delta V for these rear impact cases have been presented in Figure 2.6. Most low severity neck injury occurs at a delta v of less than 25 km/h. Impacts with a delta V of less than 15 km/h had the highest number of cases. Hell et al. (1998) used this figure to conclude that 10-15 km/h is the most critical delta v for low severity neck injury in rear impacts, although they made the recommendation that all low severity neck injury countermeasures be tested at a delta v between 12 and 25 km/h. This is supported by Meyer et al. (1998), who has also reported that a delta v above 10 km/h can produce a low severity neck injury. Gibson et al. (2000) found that crashes of a much higher delta v can lead to injury with the average delta v for chronic low severity cases as 34 km/h reported.

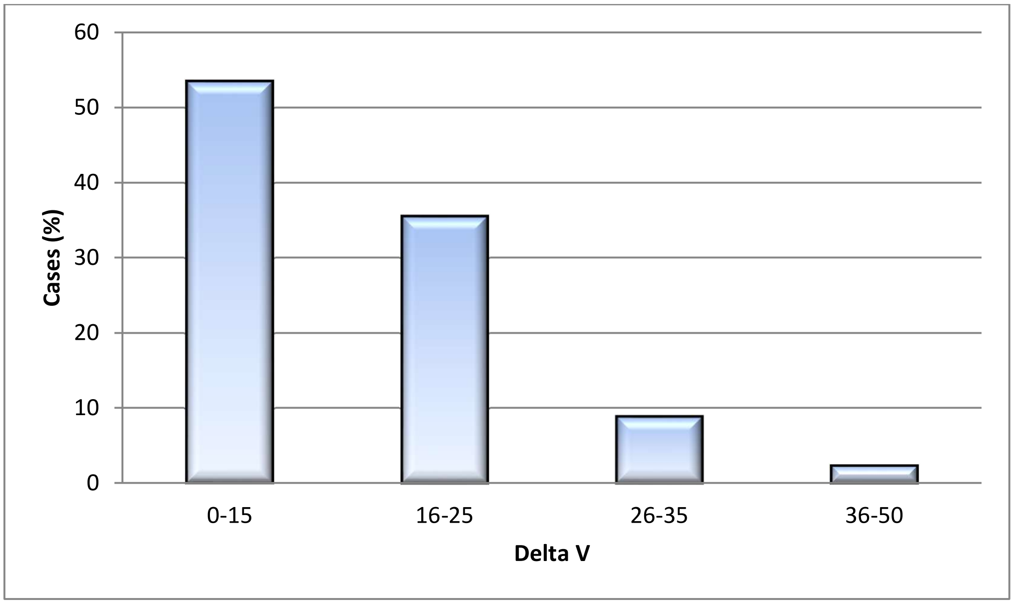


Figure 2.6: Distribution of delta V of impacts resulting in low severity neck injury (Hell et al.1998)

Equivalent barrier speed (EBS) and energy equivalent speed (EES) is a measure of the vehicles velocity, predicted from measures of damage from equivalent impacts of a vehicle into a rigid barrier Nordhoff (2005). Morris et al. (1996) reported the average EES (equivalent EBS) for frontal impacts as 29.9km/h for occupants with low severity neck injury, and 33.7 km/h for occupants without low severity neck injury. The EES for rear impacts is 31.8 km/h for occupants with low severity neck injury, and higher (36.2km/h) for occupants without low severity neck injury. This finding and the others prior to it suggest that low severity neck is more likely to occur in impacts of lower impact severity.

2.3 Mechanism of Injury

Extensive research has been undertaken into the mechanism of low severity neck injury focusing on rear impact, with some limited studies investigating front impact. Some research has provided investigation into neck injury from lateral (side) impacts. An overview of this research has been presented in this chapter to a base of understanding into the possible mechanism of injury of LSNI from different impact angles. This research is also presented for comparison with the findings of this study, presented in later chapters.

What sets side impacts apart from rear and frontal impacts (other than a large lateral component in the impact) is the range of impacts within the category. Side impacts encompass near and far side impacts as well as offset impacts either side of the midline of each side of the vehicle. Angle of impact is an important factor to determine when investigating low severity neck injury from side impact as it greatly impacts the occupant's kinematics. Where the occupant is sitting in relation to the side impact is also important, as the interior of the vehicle differs between the driver and passengers. As this the 3-point seatbelt configuration differs depending on which side of the vehicle the occupant is sitting on and how the occupant interacts with this will depend on the impact being a near or far side impact.

In this section a summary of the mechanism of injury research has been presented for all impact directions. This includes experimental data from rear, front and side/lateral impacts that have been undertaken to determine the possible mechanism of LSNI. This work is preceded by a section that outlines the natural range of motion of the cervical spine, as a reference point for the later impact related research.

2.3.1 Range of Motion of the Cervical Spine

The natural range of motion of the cervical spine varies across different the type of motion that the head and neck is able to produce. The neck can produce a variety of movement including, pure translation, pure rotation or coupled motions as illustrated in appendix? A number of research investigations have been undertaken to establish the natural range of motion of the cervical spine during different these types of motion. This testing has primarily been done on volunteer subjects, due to its low risk of injury. Ordway et al. (1993) tested the static range of motion of the whole cervical spine for 25 subjects (12 women and 13 men). Subjects were shown to have the greatest range of axial rotation with 140.2°. The range of motion for flexion/extension was 118.5° and for lateral bending as 82.5°. What is of interest for this research thesis with respect to side impacts is that Ordway et al. (1993) also found that the range of motion for flexion, extension and right and left lateral bending decreased when the head was initially rotated. The role of rotation in side impacts and LSNI is explored further in Chapter 6.

Panjabi et al. (1998) tested segmental motion between each vertebrae in the cervical spine. The types of motion tested were "combination flexion/extension (+/- x axis rotation)", "one side lateral (bending z-axis rotation)" and "one side axial rotation (y-axis rotation)". The variation in range of motion reported between each vertebrae for three different motions is illustrated in Figure 2.7. The figure shows flexion-extension is greatest between C0-C1, followed by C4-C5 and C5-C6. Axial rotation is also greatest at C1-C2. Lateral bending is greatest at the mid-vertebrae between C3-C4 and C4-C5. It is also shown that the lower vertebrae have the lowest range of movement for lateral bending and axial rotation.

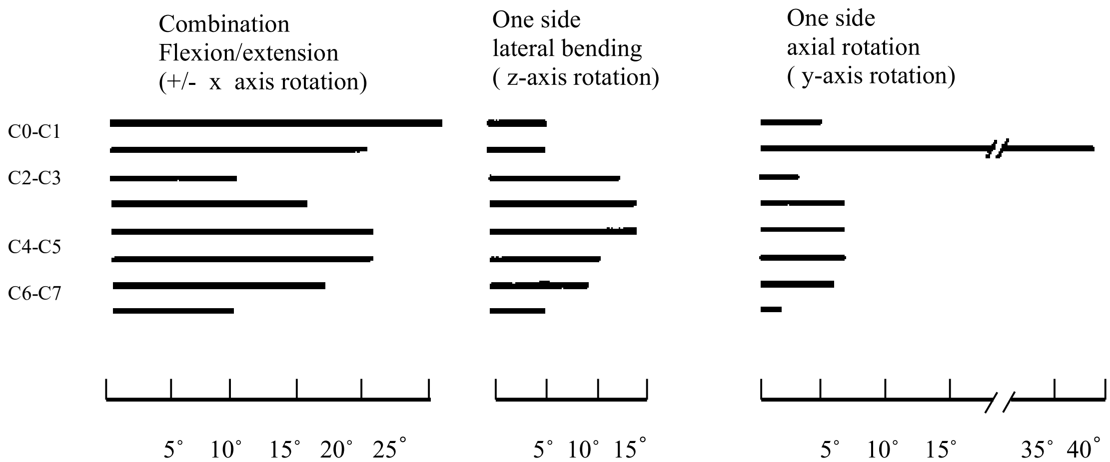
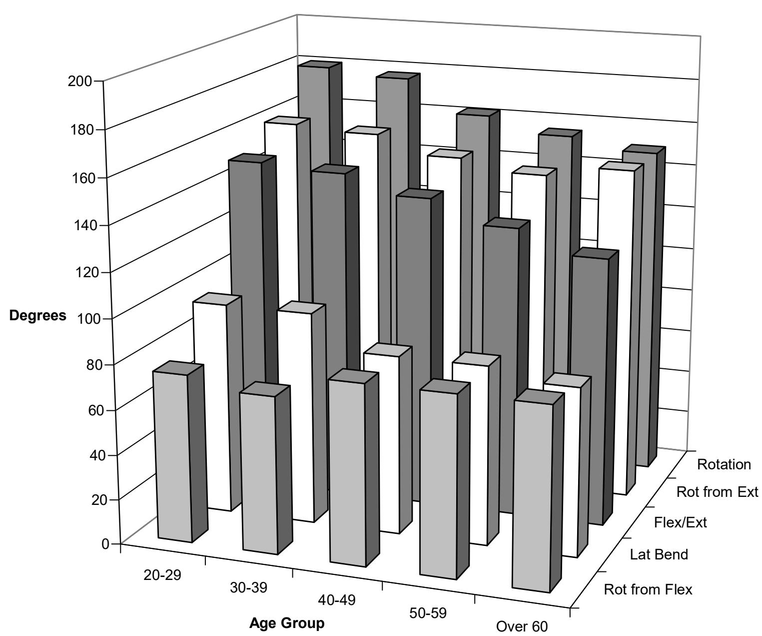


Figure 2.7: Range of motion between each vertebrae Panjabi et al. (1998)

Dvorak et al. (1992) provide a detailed illustration of the relationship between neck range of motion and age (see Figure 2.8). Older age groups were measured to have a lower range of motion than the younger age groups for rotation, rotation from extension, flexion/extension and lateral bending. A reduction in motion can be seen from age 40. There is no difference between the age groups for rotation from flexion. When comparing all neck motion, rotation and rotation from extension have an overall greater range of motion than lateral bending and rotation from flexion

Figure 2.8: Range of neck motion and age Dvorak et al. (1992)



2.3.2 Rear Impact

Extensive research has been undertaking investigating the mechanics of the head and neck during rear impact. This section provides a sample of some of the work investigations exploring what mechanism could lead to injury in rear impacts. This work is the foundation work for the later developments of protective systems in vehicles to prevent LSNI that have been mentioned previously in section 2.2.7. Some literature shows that the spine produces an s-shape during impact. It has been suggested by Harrison et al. (2000) that this is also relevant in lateral impacts. Volunteers were tested to measure their unforced lateral head translation. The average motion reported was 51 mm. This motion has been shown to be specific to each individual vertebrae. The change of direction in the cervical spine to produce an s-shape occurred at C4-C5. It was reported that lateral bending with lateral translation was the major coupled motion (Harrison et al. 2000).

The mechanism of low severity neck injury from rear impacts has been investigated extensively. There is no conclusive opinion among researchers as to the exact mechanism of injury. In spite of this researchers have raised a variety issues regarding head and neck mechanics during rear impact. The issues; torso ramping, the s-shape of cervical spine, range of motion reached during impact and the effect of seat properties have been discussed in this section as these factors may be present in cases of LSNI from side impacts.

Researchers have presented the occupant's kinematics during rear impact in a series of phases (Mc Connell et al. 1995, Ono et al. 1997, Bertholon et al. 2000). The description of the phases presented by Mc Connell et al. (1995) has been listed in Table 2.8.

Table 2.8: Phases of occupant kinematics- rear impact (Mc Connell et al. 1995)

Phase 1 Initial Response (0-100 milliseconds)	0 to 50-60 ms	-No occupant motion detected after contact with the bumper. -The seat response to the impact by moving forward 6.4-7.6cm with seat back deflection of 3-5 degrees.
	60 to 80 ms	-The vehicle and seat base moved about 10.2 cm. -Seatback deflection of 6-7 degrees rearward. -The upper body was pulled forward, this motion was initiated by the pelvis and lower trunk as it moved rearward with the seatback. -Ramping up of shoulders and upper thoracic spine occurred also, as a result of seat motion.
	80 to 100 ms	-At 80 milliseconds the forward movement at T1 was recorded at 2.54 cm. -No head movement had been recorded until 100ms, although base of the neck moved with the moving T1. This motion was attributed to the muscles of the neck exerting a forward pull on the head. -At 100 milliseconds the head has just begun to move. T1 and the base of the neck have moved forward by another 1-2 cm.
Phase 2 Principal Forward Acceleration (100-120 milliseconds)	100 to 110-120 ms	-Seat maximum rearward deflection of 10-14 degrees had occurred -Maximum acceleration at top of head was between 5-15g. -Vehicle had travelled forward 15.2-17.8 cm with T1 moving forward 8-10.2 cm. -Some subjects showed visible sternocleidomastoid function from the load applied to the muscle. -Early head motion reported to be rotation (rearward) close to the heads center of gravity.
	110 to 170 ms	-Head rearward rotation 10-15 degrees followed by forward translation initiated by the necks forward motion.
	180 to 200 ms	-Head has reached maximum rearward rotation between 18°-51°.(It was noted that taller subjects peaked 10-30 milliseconds later) -Seatback, torso and T1 angle had decreased 5-6 degrees.
Phase 3 Head Over Speed/ Torso Recovery (200 -300 milliseconds)	200 to 280 ms	-Seat back returns to its pre-impact angle -Torso has achieved vehicle's forward velocity (or higher). -Head had achieved greater velocity than torso of 1.5-2.5g and was actively slowed down by the neck.
Phase 4 Head Deceleration/ Torso Rest (300-400 milliseconds)	300 to 400 ms	Head continues forward movement (faster than shoulders), while the torso and lower body returns to post impact position.
Phase 5 Restitution Phase (400 -600 milliseconds)	400 to 600ms	Body had achieved vehicles impact related velocity change and began to return to post impact position. The trunk and hips were reported to be slightly higher than in the initial post impact position.

The work undertaken by McConnell et al. (1995) identifies upward motion of the torso (torso ramping) occurring at 60-80 milliseconds. Matsushita et al. (1994) reported ramping of the torso with an x-ray analysis of human subjects exposed to low speed rear- end impacts. From this analysis it was observed that the cervical and thoracic spines' natural curves straightened and the cervical spine is shortened. This shortening was attributed to a compressive load being placed on the cervical spine due to the upward movement of the thoracic spine (torso). This is believed to be due to subjects leaning forward with "stooped shoulders", created by contact with the seat back. During these investigations range of motion was always below the subject's natural range of motion. In spite of this, the subjects in Matsushita et al. (1994) study reported discomfort from injury. Davidson et al. (1998) illustrate this upward movement of the torso by mapping the motion of T1 in Figure 2.9.

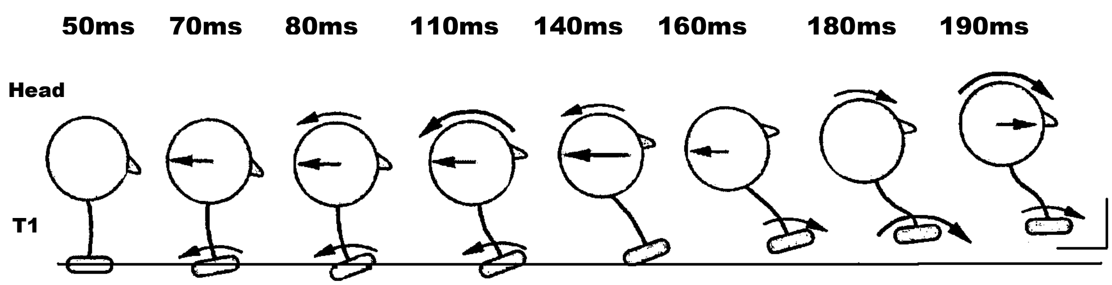


Figure 2.9: Diagram of torso ramping- rear impact (Davidson 1998)

Panjabi et al. (1998) and Deng et al. (2000) having undertaken cadaveric specimen tests into rear impacts, found that the range of neck motion went beyond the subject's natural tolerance. Panjabi et al. (1998) took segments of the cervical spine (from occiput to C7 or T1) and mounted them to a base in a neutral position (NP) with a steel head representing the 50th percentile male attached. Motion was measured with a head restraint. Panjabi et al. (1998) reported that there is a point where following impact the spine produces an s-shape. An illustration of the s-shape is presented in Figure 2.10.

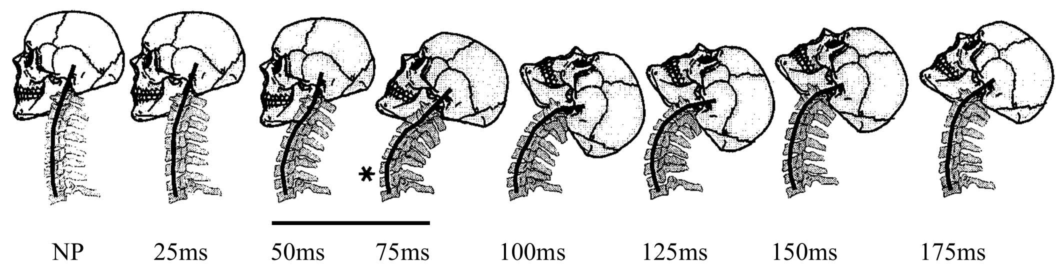


Figure 2.10: S-shape (*) in cervical- rear impact (Panjabi 1998)

It is shown that the s-curvature of the spine is shown occurred between 50-75ms. This is early on in the phase of neck motion during a rear impact. It is also reported that each of the joints of the cervical spine were pushed beyond their physiological range of movement. Panjabi et al. (1998) speculate that the s-shape curvature of the cervical spine in rear impacts is potentially more damaging than the following head-neck extension that was reported as mechanism of injury in early literature. Panjabi et al. (1998) illustrates the neck motion data obtained from the specimen experiments in terms of each individual vertebrae. This is shown in Figure 2.11. The light grey represents the vertebrae, the white represents the relationship between the adjacent vertebrae and the dark grey is the motion beyond the physiological range.

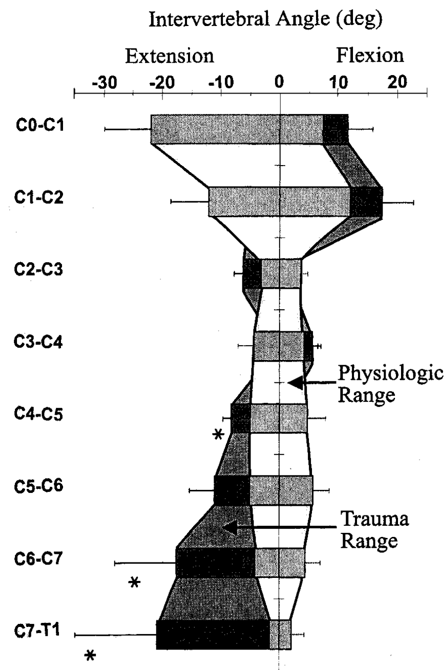


Figure 2.11: Range of motion of cervical spine joints- rear impact (Panjabi 1998)

When comparing the range of motion of individual vertebrae, each joint is extended beyond its physiological range and moves into what Panjabi et al. (1998) terms as trauma. Differences between the joints are seen when comparing the type of movement experienced at each joint. The joints C0-C1, C1-C2 and C3-C4 experience flexion beyond their physiological ranges. The other joints C2-C3, C4-C5, C5-C6, C6-C7 and C7-T1 all experience extension beyond their physiological ranges. It is the point where the motion between the joints changes from flexion to extension where the s-curve occurs. Kaneoka et al. (1999) investigating human volunteers and Bertholon et al. (2000) investigating cadavers have also reported the cervical spine producing an s-shape during simulated rear impacts. Deng et al. (2000) also found that in rear impacts the cervical spine to go beyond natural motion. When investigating the facet joints, it was found in all test cases the peak strain occurred before seat contact. The average maximum strain in the facet joints as reported was between 22% and 60%, with the highest occurring at C5-C6. This magnitude of strain is believed to be great enough to stretch capsular ligaments beyond their natural tolerance. Typically, the s-shape of the spine is reported as occurring in phase 2 although, Bertholon et al. (2000) reports it occurring earlier in phase 1. It was also found that the addition of padding to the rigid seat made the s-shape more pronounced.

Ono et al. (1998) also found that seat stiffness had a noticeable effect in the kinematics of the spine in rear impact. Seat properties are shown to influence the timing of the straightening of the spine. It is suggested by Ono et al. (1998) that there is an optimal seat stiffness to protect the spine in rear impact. When there is increased stiffness there is a greater upward movement of the torso, straightening the spine, placing the neck under compression. With decreased seat stiffness there is greater torso rebound increasing the shear at the upper cervical spine (Ono et al. (1998)).

The effect of seat back angle was also observed to influence the initial curvature of the spine (Deng et al. 2000). With a seatback angle of 0 degrees more initial curvature was reported. When the seatback angle was increased to 20 degrees more, ramping of the torso was evident with the increase in upward movement of the pelvis marker (Deng et al. 2000).

2.3.3 Front Impact

The kinematics of the head and neck of a restrained occupant in a frontal impact have been described by Morris et al. (2000) in two phases; the phase 1: protraction motion and phase 2: flexion motion (see Figure 2.12).

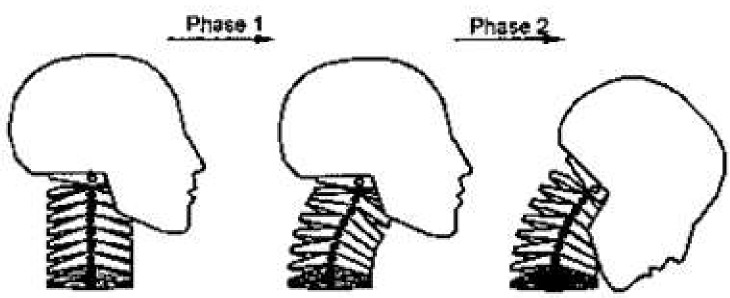


Figure 2.12: Head and neck motion- front impacts (Morris et al. 2000)

In a frontal impact the occupant is forced forward (retraction motion phase). The occupants the torso of is then suspended by the seatbelt and the head continues to travel forward (flexion/hyperflexion phase). This forward motion of the head will force the cervical spine into flexion and even hyper-flexion. Hyper-flexion has been described as a potential mechanism of injury in frontal impacts. (Morris et al. 2000). Unlike in rear impact, during frontal impacts the motion of the neck has a natural stopper when the chin contacts the chest (Morris et al. 2000). Watts (1999) reports that when the chin contacts the chest the load is removed from the neck and transferred to the chest. It is also suggested that in some cases this creates a new point of rotation leading to an increased neck load (Watts 1999).

Kallieris (1991) investigated in detail the injuries to the cervical spine in cadavers from simulated front impact at velocities of 30, 50 and 60km/h. Three different types of seatbelts were used, a 3-point standard belt, a double shoulder or an inverted V-pelvic belt. The kinematics for a subject restrained by a standard 3-point belt is illustrated in Figure 2.13. Three images of the occupant's kinematics have been presented as an overlay at time 0ms, 90ms and 160ms after the impact. The occupant's initial position at 0ms is shown in white. The occupant's motion at 90ms is show in light grey with forward torso and head motion and some neck flexion visible. The occupant's position at 160ms shown in dark grey, highlights extensive neck flexion.

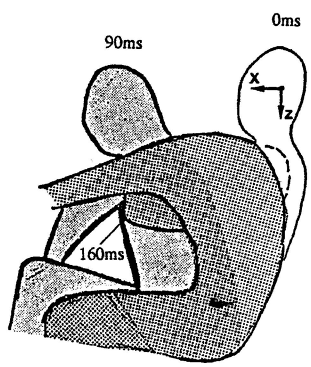


Figure 2.13: Kinematics of belted occupant- front impact (Kallieris et al. 1991)

Kallieris et al. (1991) also reported that during frontal impact there an initial translation phase lasting between 30ms and 70ms. The type of restraint and impact velocity affect the duration of this translation phase. Following translation, the cervical spine is reported to be simultaneously compressed at the front and extended at the back, leading to an increase in the overall length at of the cervical spine. Injuries were found to occur at all levels of the cervical vertebrae. The most common injuries were shown to be to the intervertebral discs, ligamentum flavum, joints and posterior transverse ligament.

The pain symptoms of rear seated volunteers who were exposed to low speed frontal impacts have been reported by Croft et al. (2011). Nineteen subjects (17 males, 2 females) with an average age of 37 years were exposed to full scale frontal impacts with a Delta V ranging from 1.4 mph to 3.9 mph. Subjects were pre-screened for any pre-existing spinal injuries or abnormalities. All subjects were rear seated in the bullet vehicle with the belt fastened, with 9 subjects positioned in the middle of the seat with only a lap belt. All but two subjects experienced discomfort resulting from the impact. The predominant symptom identified for was posterior neck discomfort followed by shoulder and thoracic discomfort. None of the subjects rated the pain as severe, but did report the pain to be between minimal to slight, with symptoms 15 minutes to a week.

2.3.4 Lateral (Side) Impact

The most detailed work investigating neck mechanisms in lateral impacts has been undertaken by Kallieris et al. and presented in several publications (1981, 1990, 1991, 1996). This work details the kinematics of post mortem human subjects tested in lateral impacts and provides a full report of the anatomical structures injured. In the earliest work by Kallieris (1981), post mortem human subjects were placed on a sled and accelerated into either a rigid wall or a wall with various levels of padding. The speed of impacts into the unpadded wall was either 15 mph or 20 mph, whereas and the speed into the two padded walls was 20 mph. It was found that subjects received injuries in both padded and unpadded situations. Although, the padding of the wall did have an influence on the magnitude of the injuries recorded. Injuries in the padded cases were a maximum of AIS 1 as haemorrhages to the epidural space, posterior longitudinal ligament or vertebral discs. The injuries in the unpadded cases are of a maximum AIS 3, with the most serious injury a haemorrhage between C0-C2. The results showed that, for the rigid wall tests, the resultant head acceleration for the 15 mph tests peaked at 190g for 20ms duration, and for the 20 mph test 130g for a maximum of 40ms. For the T1 lateral acceleration, for the lower speed test the peak acceleration was 110g for 10ms; and for the higher speed test it was 120g for 10ms. The results for the padded cases show greater variation between subjects. The resultant head acceleration for the trial with the padded wall (20mp/h) was between 110-190g for 20-30ms for the first padding condition (APR Pad, urethane block) and for the second padding condition (fibre glass matrix pad) 170g for the duration of 20ms. The peak T1 lateral acceleration for the first condition was 90g for approximately 20 ms, and for the second condition the peak was 170g for approximately 20ms. The results show that padding the decreases the acceleration of the torso but does not decrease the acceleration of the head. This explains why vertebral column injuries are received by subjects in both padded and unpadded conditions (Kallieris et al. 1981).

Kallieris et al. (1990) also investigated occupant kinematics in a side impacts, paying special attention to their interaction with the seat belt. It was hypothesised that the conventional three point seatbelts offer little protection in a far side impact, allowing the occupant to slide out of the belt. This work is of great interest to this study as the occupants in all six simulations have been restrained by a three-point belt. The work by Kallieris et al. (1990) was expanded beyond 90° side impact angles to include impacts at 60° angles, at a speed of 50 km/h. In all tests, rear seated cadavers were restrained by three-point belts mounted with the shoulder belt in the opposite direction to a standard shoulder belt (Figure 2.14). These occupants were impacted on their far side. In one test, a US SID dummy was positioned as a near side front passenger for comparison. The results show that neck bending angle was within biological tolerances with a range of 45-65° with an average of 53°. The results from cadavers showed that maximum neck injuries sustained were of AIS 1. These injuries are attributed to seatbelt induced loading. It was found that the 60° test produced higher loading from the belt compared to the 90° tests. Comparisons with the near side front passenger (dummy) show that with the standard three-point belt, the occupant will experience higher angular head and neck velocities and accelerations (Kallieris et al. 1990).

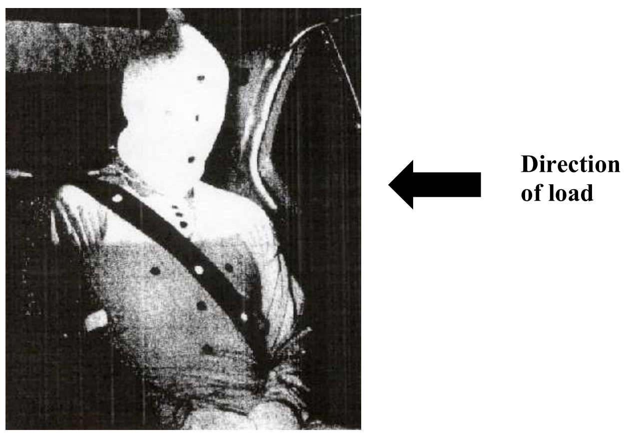


Figure 2.14: Seatbelt configuration and direction of load (Kallieris 1990)

A follow up study was undertaken Kallieris et al. (1991). In this study, cadavers were examined in 90° lateral impacts as both near side front (with standard belts) and far side rear belted occupants. A diagram of the subject's kinematics is shown in Figure 2.15. It was found that regardless of the restraint used, it was observed that there was an initial lateral displacement of the head in relation to the torso, followed by lateral bending towards the direction of the impact. During the initial loading phase, a tension was applied on the side of impact while a compression force occurred on the opposite side of neck. It is important to note that Kallieris et al. (1991) reported that in these lateral impacts a rotation of the head was observed around the vertical axis, although subjects with obvious head rotation were excluded from the study.

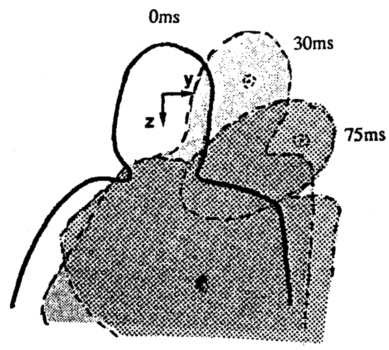


Figure 2.15: Occupant kinematics with standard restraint, side impact Kalleris et. Al. (1991)

Kallieris et al. (1991) reported differences in the neck mechanics between near and far side impacts. In a lateral impact at 50 km/h, the head bending for near side front passengers was greater at 62 ° than that for far side rear passengers at 57°. The mean maximum angular velocity and mean maximum angular acceleration were also higher for near side front passengers than for rear side rear passengers. Timing differences were also observed between the two occupant groups. Near side front passengers head/neck motion was completed in a shorter time at 20-30 ms as compared to that of far side rear impacts at 80-100 ms. Post impact autopsies showed that these later impacts produced AIS 1 neck injuries. An overview of the nature and location of these injuries have been presented in Figure 2.16.

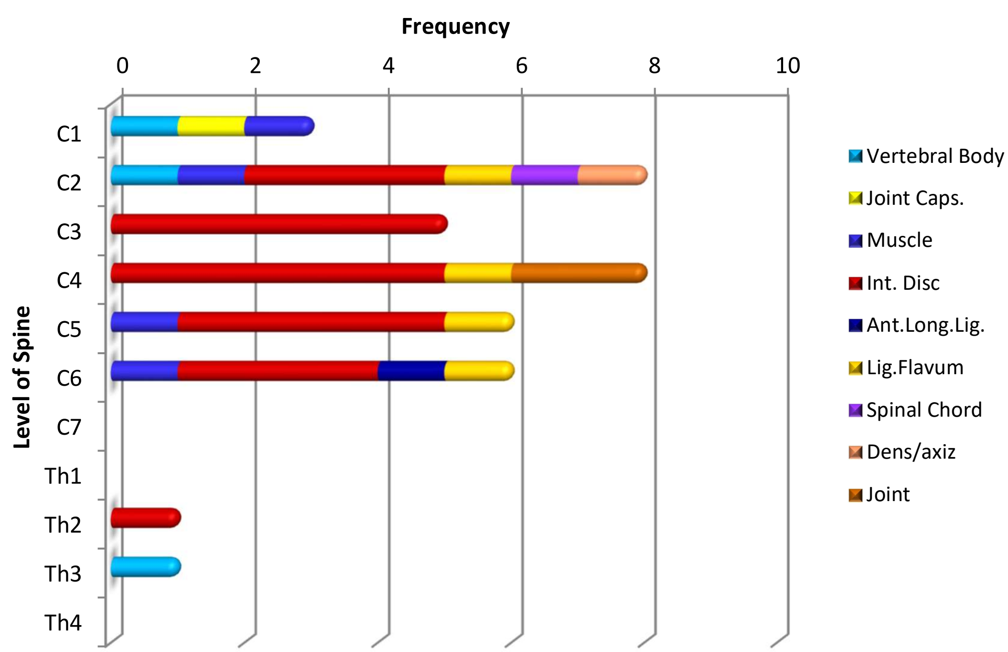


Figure 2.16: Site and type of cadaver spine injuries Kallieris et al. (1991)

The most common regions of the cervical spine injuries are C2 and C4, followed by C5 and C6 (Figure 2.16). The most frequently occurring injury is to the inter-vertebral discs. Other structures injured along the cervical spine are joint, muscle, vertebral body, joint capsule, dens, spinal cord, ligament flavum and anterior longitudinal ligament.

McIntosh (2007) investigated the relationship between biomechanical response and injury severity using data from cadavers exposed to lateral impacts. The cadaver tests were undertaken by the University of Heidelberg's Institute for Legal Medicine in the 1980's, presented in by Kallieris et al. (1981, 1990, 1991, 1996). The lateral sled accelerations ranged from 12g-23g and cadavers were impacted into either a rigid or padded wall configuration. Neck injuries observed in the cadavers chosen for analysis by McIntosh (2007) were either no injury, AIS 1 injury or AIS 2 Injury. When comparing the AIS 0/1 cases to the AIS 2 cases, McIntosh (2007) found that there were significant differences in the mean head acceleration, neck force and moment. The correlation between the injury groups for lateral neck bending moment and/or lateral shear force was found to be low.

Horsch et al. (1979) also investigated the interaction between the three-point seatbelt and cadaver subjects in lateral impacts. In these tests, different sides of anchor point were examined (near side or far side impact). When comparing different anchor points of the shoulder belt, it was found that the subject's kinematics varied. In 35 km/h 10g tests, it was found that restrained subjects in far side impacts, only received initial lateral protection. Following this, the subject rotated out of the shoulder belt. When the shoulder belt was anchored at the shoulder closest to the impact (as in near side impact) there was less displacement of the subject's torso as they moved laterally into the belt. Although, in spite of the additional protection of the subject's torso, the neck experienced greater loading from the slipping belt (see Figure 2.17).

Figure 2.17: Cadaver- Seatbelt interaction, near side lateral test (Horsch et al. 1979)

The loading of the belt against the occupant's neck was found to be dependent on anchor height, anchor fore-aft position and the subject's seating posture and height (Horsch et al. 1979). The maximum shoulder belt loading on the neck was reported to be as high as 8kN. The injuries to the cervical spine recorded for both belt anchorage positions were to the ligament and other overlaying soft tissue structures. It was also noted that, with the shoulder belt anchored on the side of the impact (near side), subjects received greater injuries.

Faerber (1982) investigated side impacts at angles of 60°, 90° and 120° investigating the interaction between the seatbelt and front seat positioned dummies. The findings of Faerber (1982) support the findings of Horsch et al. (1979) that seat belts provide limited protection in lateral impacts. It was found that in side impacts belts provide the best protection in angled impacts (Faerber, 1982).

Bendjella et al. (1987) investigated head and neck responses of cadavers under the influence of low and high g lateral sled acceleration. It was found that the magnitude of acceleration had an influence on neck mechanics. During the low g tests it was observed that initial shoulder deflection was followed by pure translation of the head. The final phase of head movement was rotation about the anterior posterior and inferior/ superior axes (3 dimensional movement). The high g tests produced greater head displacements. During the high g tests only one subject received injuries to the neck (cervical fractures), no injury occurred in the low acceleration tests.

Lateral tests of human volunteers undertaken by the USA Naval Biodynamics Laboratory (NBDL) and have also investigated the influence of acceleration on head and neck response Wismans et al. (1983). This data was used for comparison to the human body model in Chapter 3. Subjects were seated in a sled and restrained with a harness consisting of a lap belt, two V shape shoulder straps and an additional chest strap. All tests were performed below injury causing threshold. Subjects were exposed to high onset-long duration, low onset-long duration and high onset-short duration sled acceleration pulses. The First Thoracic Vertebra (T1) acceleration in each type of test was shown to be similar. The head accelerations were shown to be higher in the high onset-long duration test. An analysis of the same data was undertaken by Wismans et al. (1983) to provide a series of recommendations for dummy necks. Sled velocity for these subjects investigated ranged from 3.1 to 6.6 ms with peak accelerations ranging from 4.9 g to 10 g. The results show that the T1 acceleration has an initial peak 2-3 times higher than the sled acceleration. It is important to note that a variation in T1 acceleration between the subjects was reported. This is believed to be due to a variation in the way in which the subjects interacted with the harness. When examining the subject's head displacement

Wismans et al. (1983) observed that following initial motion there was head rotation in the impact direction (lateral bending) followed by head rotation about the vertical axis. It was also viewed that for the more severe tests there was greater head excursion (translation).

The subjects tested by the NBDL were also exposed to oblique (45 ˚) impacts. These tests have been reported by Wismans et al. (1986) and compared to the results for lateral and frontal tests. The T1 acceleration (in impact direction) observed for oblique impacts shows an initial peak, reported to be a result of interaction with the harness. The peak mean T1 acceleration is shown to be greater than the lateral acceleration. The relative head motion between the head and T1 show that the head excursion (translation) in the oblique tests is greater than in the lateral tests. In spite of this, the head excursion (translation) is not as great as that in the frontal tests. It was found with cases of all impact directions that flexion (in direction of sled acceleration) and head rotation is dependent on the peak sled acceleration. It has been reported that in the lateral and oblique cases that head flexion (in impact direction) is followed by a head rotation about the vertical axis. This does not occur in frontal impacts. Guccione et al. (2001) compared the kinematic response of three subjects tested by the NBDL in simulated oblique impacts and compared them to lateral and front impacts. It is shown that 45 ˚ oblique tests have frontal and lateral components of equal magnitudes.

2.3.5 Head Contact

The issue of head contact coupled with low severity neck injury, is an important one. Head contact can occur in side impacts as the head is in close proximity of the door. Intrusion during a side impact of the vehicle structures, such as the A pillar and B pillar can enter into the vehicle striking the occupant. Jakobsson et al. (2000) shows that in frontal and side impact cases where an occupant experiences a head contact, there is an increased risk of an occupant also receiving a low severity neck injury. This is the case for occupants with neck injury that shows initial symptoms as well as persistent symptoms (less than 3 months) and passing symptoms (1 year). It is reported that an occupant's initial symptoms are far greater when the occupant has a head impact (compared to no head contact). It is important to note that in this study the head contacts recorded were any contact with in the interior of the vehicle including airbags, but did not include head restraints. This occurrence may be able to be explained by the research that suggests that neck injury occurs with direct loading to the head by Rizzetti et al. (1997). Cadavers were exposed to frontal, lateral and occipital (rear) head contacts with padded and unpadded impactors. The magnitude of the impact forces were of a high severity at 20000N for the un-padded impacts. The magnitude was smaller between 3600N and 8400N for the padded impacts. The results show that the unpadded impacts produced skull fractures. The padded impacts produced no skull fractures although brain injury was detected. The most interesting finding relating to this research was that (combined with the severe head injuries from both padded and unpadded impacts) cervical spine injuries were reported in all but two subjects. The cervical spine injuries reported were strains of the soft tissue and ruptures of the ligaments. As the data in this study is separated by impact direction, it is possible to assess the effect that lateral head impact has on neck injury. It is shown that the subjects exposed to head impacts, including lateral head impacts were likely to receive strains to the soft tissues and ruptures of the ligaments in the neck.

Ono et al. (2002) also examined the effect that head impact has on cervical spine injury by testing human volunteers. The directions of impact tested differ from those examined by Rizzetti et al. (1997). The load directions examined were, upward load to the chin, rearward load to the chin (with the subject wearing padded mask), rearward load to the chin (with the subject not wearing mask) and rearward load applied to forehead. These impact directions would compare to contact with an airbag. The impact loads were much lower than those tested by Rizzetti et al. (1997) at approximately 210N. Ono (2002) investigated the neck strain between the cervical vertebrae in volunteers while they received low magnitude, short duration impacts to the face and head. These have been presented against the normal strain measures in Figure 2.18. The response of the cervical spine differs depending upon the loading direction. The magnitude of strain differs at each level of the cervical spine and the magnitude is greater for some impact directions (Ono et al. 2002).

The chin rearward results (displayed as black stripes) suggest an s-shape is created in the cervical spine. This is because there is strain recorded in opposite directions. For C2/C3 and C3/C4 there is strain found in the "expanded direction" or positive direction. Between vertebrae C4-C7 the strain is in the negative direction and compressive.

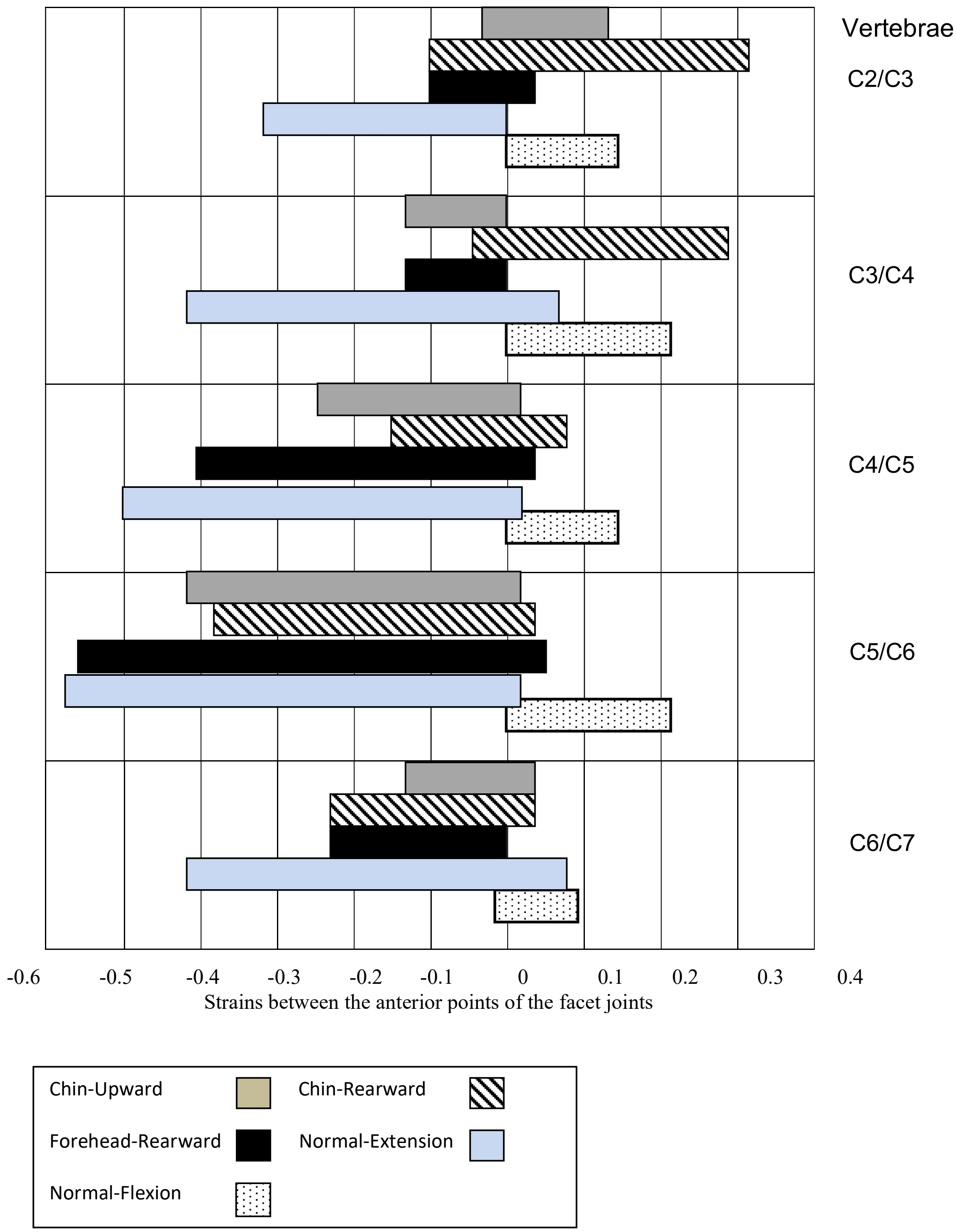


Figure 2.18: Strain at facet joints for different head contact positions (Ono et al. 2002).

Neck injury resulting from impact to the forehead was investigated by Ivancic (2013). A rearward force was applied to a custom design dummy head that was attached to rear impact BioRIDII dummy with a cadaver cervical spine (C1 to C7) fitted. Two different magnitudes of force were tested of 249N and 504N. The force of the larger magnitude 504N applied to the forehead was found to produce macroscopic injuries to C6 to T1. The injuries include bone fractures, disc and ligament injuries are listed in Table 2.9. In considering these injuries it is important to note that the average age of the cadaver specimens was 86 years and age related degenerative changes were noted. The impact of lower magnitude force 249N into the forehead of the human dummy with cadaver neck did not result in any macroscopic injury.

Table 2.9: Neck injury from impact to forehead Ivancic (2013)

Specimen #	Injuries
1	C6/7 All and anterior disc injury C6 fracture through flag screw
2	C7/T1 All and anterior disc injury C7 anterioinferior vertebral body fracture
3	T1 fracture through mount fixation screw
4	T1 fracture through mount fixation screw
5	C6/7 All and disc injury C7/T1 ISL, SSL, posterior CL injuries
6	C7 fracture through flag screw and lateral rod

2.3.6 Head Turned Posture/ Head Rotation

Occupant seating posture has been shown to influence occupant kinematics and risk of neck injury in impacts. Head turned posture occurs when the occupants head in not in a neutral position facing forward, but is rotated laterally at the time of impact. Occupants with turned head posture are usually in this seating position due to awareness of the pending impact. In rear impacts this may be as a result of viewing the rear view mirror in rear impacts, looking out of a side window in side impacts or turning at an intersection in any impact direction. Head turned posture and head rotation and LSNI has been investigated in rear, front and side impacts. Jakobsson et al. (2000) found that, in almost all side impacts when an occupant reported to have their head turned, it was turned towards the impacting object. This head turned position was shown to increase the risk of initial symptoms, and persistent symptoms of low severity neck injury in the occupants.

Watts et al. (1996) also reports that turned head posture in a rear impact will result in more serious injuries, as the initial turned head condition increases the stiffness of muscles and ligaments. It is also believed that a turned head puts the discs under a combined shear-tensile load. Recent literature also works to identify the structures in the neck that are injured when the head is turned or rotated during an impact.

Computational modelling of the neck in rear impacts by Storvik et al. (2011) found that when there was head turned posture, strain in the ipsilateral facet joints increased when compared to forward facing head position. The computational model used was the same MADYMO Head and Neck Model also used in this thesis in Chapter 4 and Chapter 5. The range of angles of the head rotation measured was from 0-60 degree with the rotation applied to the C1-C2 spinal level. The results showed that with the 60 degree heard rotation, the strain in ipsilateral facet joints increased by 47-196%. It's important to note that in these trials the head contacted the head restraint at varying times dependant on impact speed. It was reported by Storvik et al. (2011) that head rotation did not influence the time of head restraint contact.

Ivancic et al. (2006) also found that head rotation also had an adverse effect on the neck in rear impact. In undertaking experimental tests using cadaver specimens, it was found that there was greater elongation of the vertebral artery when comparing head turned to head forward facing conditions. The elongation of the vertebral artery can be attributed to the symptoms of headaches, blurred vision, tinnitus, dizziness and vertigo following a low speed impact.

2.4 Summary of findings

Research investigating the incidence of low severity neck injury from side impacts shows, the injury occurs frequently enough to be of concern. There has been a wide range of research undertaken into LSNI from rear impacts. This work provides a base of understanding for work to be undertaken at other impact angles. In addition to the research available on rear impacts, there is a body of work regarding LSNI from front impact that can also be drawn to provide a base of understanding for work undertaken into side impact, especially oblique side impacts that have a frontal component.

The factors that have been shown in this chapter to be common amongst individuals who receive a LSNI from a rear or front impact are;

- gender- female
- age- adult (under 40 years)
- seating position- driver
- greater mass of striking vehicle
- low Delta V
- poor head restraint positioning

The additional factors have been shown to influence the likelihood of LSNI and the severity of the injury are head contact or head/ neck rotation during impact. Work has not been previously undertaken to determine the factors associated with Low Severity Neck Injury. An investigation into the factors associated with the injury from side impacts has been presented in the following Chapter 3.

In addition to the research into the factors associated with LSNI, there is a wide body into the mechanism of LSNI. This work also focuses on rear impacts. The mechanism of injury is still inconclusive, yet a number of hypothesis have been investigated for rear impact injury. The hypothesis presented in this chapter that may have relevance to the mechanism of injury in side impact are;

- neck motion beyond natural range,
- s-shaped torso ramping,
- torso ramping,
- interaction between the head restraint or seatbelt,
- injury to the facet joints or ligaments.

There is limited literature available investigating the mechanism of injury from side impacts. Most of the literature available investigates 90° impacts, with little investigating 60° and 45° impacts. No known research exists investigating the mechanism of injury, using the crash configurations from real world impacts. To provide a greater understanding of the mechanism of LSNI in side impacts, in Chapter 5 an investigation into head and neck motion has been undertaking using computer simulations of real world side impacts that have been identified as common using the information from Chapters 2 and Chapters 3.

Chapter III
Research Framework

3.1 Introduction

This thesis uses a novel methodology to investigate neck injuries from real-word side impacts. This methodology has not been used before to investigate low severity neck injuries (whiplash) from side impacts. A detailed methodology for each of the individual studies undertaken in this thesis are presented in each chapter. An overview of the research framework has been presented below in Table 3.1.

Table 3.1: Research Framework

Analysis of Real-World Crash Data Chapter 4
• Two accident databases were analysed and compared to provide a base of understanding of the common factors associated with LSNI from Side Impacts. • These databases are the Monash University Accident Research Centre (MUARC) accident database, containing Australian crash data and the United Kingdom, Loughborough University Co-operative Crash Injury Study (CCIS).
Case Selection- Chapter 6
• Six cases were selected from the Monash University Accident Research Centre database. In three of these cases an occupant received a LSNI and in three of the cases the occupant did not. • The cases with an occupant receiving a LSNI were chosen as they displayed the typical characteristics shown in Chapter 4.
Validation of Combined Human Model with Detailed Neck- Chapter 5
• The MADYMO (Mathematical Dynamic Modelling) human body model combined with detailed neck model (termed in this thesis as combined human model) • The human model and the detailed neck model have been extensively validated as separate models (Happee et al. 2000). The two models combined have been used in rear impact research (van der Horst et al. 2001) and the adapted neck model has been used to research into injury causation and injury management (Sim 2004, Storvik et al. 2011). • Additional validations of the combined model were undertaken at lateral and oblique directions. One dataset contains cadaver data for a lateral impact provided by Wayne State University (Cavanaugh et al. 1990 1993, Irwin et al. 1993) the other dataset has volunteer data for lateral and oblique impact from the Naval Biodynamics Laboratory (Wismans et al. 1986).

Determining Vehicle Acceleration- Chapter 6

- To determine the acceleration pulse of the crash *Human, Vehicle, Environment (HVE)* crash reconstruction software was used to reconstruct the six selected cases.
- HVE has been validated by Day et al. (2004, 2001, 1996, 1990, 1989).
- The vehicle model used to reconstruct the LSNI was an Opel Omega to replicate the Holden Commodore.
- The environment models chosen from library to match the crash configurations of the real world crashes.

↓

Computer Simulation of Side Impacts- Chapter 6

- The MADYMO (Mathematical Dynamic Modelling) human body model combined with detailed neck model as programmed with a vehicle model of a Holden Commodore.
- The crash accelerations from the HVE reconstructions were programmed into the model to replicate the real world crashes in the selected cases.
- The model output for the 90 degree, near-side and 90 degree far-side impacts is lateral acceleration for T1 and head.
- The model output for the 45 degree, near-side impact is lateral acceleration for T1 and head and frontal acceleration for T1 and head.
- For all of the computational reconstructions an animated kinematic file was available, the still images from this file has been presented.

3.2 Data Analysis to Create a Base of Understanding

To develop a base of understanding of low severity neck injury from side impacts in addition to undertaking the literature review presented in the previous chapter, an analysis of two crash data bases has been undertaken and presented in Chapter 4. This analysis was undertaken to determine the unique factors associated with low severity neck injury resulting from side impacts. The outcomes from this analysis guided the later selections in Chapters 6 of the real-world crashes that were reconstructed.

The two accident databases have been used in this research are the Monash University Accident Research Centre (MUARC) accident database, containing Australian crash data and the United Kingdom, Loughborough University Co-operative Crash Injury Study (CCIS). Due to there being slight variations in how the crash data was collected. The results from each of the databases have been presented separately in the following section, as each data base has slightly different criteria for collecting information.

The review of literature presented in Chapter 2 shows that for crashes where an occupant has received a low severity neck injury there are a number of common factors. These factors include gender, age, the seating position in vehicle, angle of impact, severity of impact and head contact. Other factors within the vehicle such as seat belt use, airbag deployment and the head restraint are also shown to have an influence on injury. The detailed analysis of the side impact crashes of the two databases provided the opportunity to investigate what of the aforementioned that are listed in literature as common for low severity neck injury are relevant when this injury occurs in side impacts.

At the time of this study, the Monash University Accident Research Centre (MUARC) accident database contained crash and injury data for 671 adult occupants. Each accident case in this database contained a detailed medical report and vehicle inspector's report. The data collected in each of these reports have been described in detail in Section 4.2.3. The United Kingdom, Loughborough University Co-operative Crash Injury Study (CCIS) database is a much larger with up to 1600 cases being collected each year at the time this analysis of data was undertaken. The case data collection is similar to that of the Australian data with each crash recorded containing a detailed medical information and a vehicle inspection, and is also discussed in detail in Section 4.2.3.

The injury data in each data set is coded according to the Abbreviated Injury Scale (AIS) (AAAM 1985). Injuries with the code for neck strain acute with no fracture or dislocation (low severity neck injury), 640278.1 were the focus of this research.

3.2 Real World Crash Analysis

Two computational crash reconstruction studies have been undertaken in this thesis. One a detailed reconstruction of 6 real world crashes which included both and vehicle and an occupant model.

3.2.1 Occupant Model

The human body model combined with detailed neck model (termed in this thesis as combined human model) was developed by MADYMO (Mathematical Dynamic Modelling) are validated models that have been used in previous research (de Jager et al. 1996, Happee et al. 2000, van der Horst et al, 2001). To understand the capabilities of the model, additional validations of the combined human model with detailed neck model in lateral and oblique impacts have been undertaken. The purpose of these extra validations is to compare the combined model to available experimental data. The experimental data used for comparison and the methodology of these simulations are presented in detail in Chapter 5 and include the comparisons of two different data sets. One dataset contains cadaver data for a lateral impact provided by Wayne State University (Cavanaugh et al. 1990 1993, Irwin et al. 1993) the other dataset has volunteer data for lateral and oblique impact from the Naval Biodynamics Laboratory (Wismans et al. 1986).

3.2.2 Case Selection

To undertake detailed analysis of real world crash cases that resulted in Low Severity Neck Injury six cases were selected from the Monash University Accident Research Centre database. In three of these cases an occupant received a LSNI and for comparison these were compared to three of the cases the occupant did not. The cases with an occupant receiving a LSNI were chosen as they displayed the typical characteristics shown in the data analysed from the crash databases undertaken in Chapter 4. This data showed that LSNI was more likely to occur from aside impact that was at an angle of between 0 and 90°. Due to this crash configurations modelled in this study are; near side 90° impact, far side 90° impact and a near side oblique impact of 45°.

The data in the real world data bases analysed in Chapter 3 showed that female occupants are more likely to receive a LSNI from a side impact compared to males. In spite of this only cases with male occupants were considered in this study because the human model used was only has the geometry of a 50th percentile male. In addition to this only cases with drivers were modelled as the data in Chapter 4 shows that males are most likely to be injured when they are drivers. A summary of the case data has been presented in Tables 6.4, 6.6 and 6.8, including occupant age, weight, seat belt usage and airbag deployment. The case data also includes information regarding injuries received, the vehicle details, and impact details.

3.2.3 Computational Crash Reconstruction

A detailed methodology of the computational vehicle and occupant model and the reconstructions undertaken to analyse low severity neck injury has been presented in Chapter 6. The occupant had a seatbelt in all of the crash models for all of the impact directions. An steering wheel airbag was modelled in the 45 degree oblique near side crash as airbag deployment was a factor in the real world cases being reconstructed.

Before the real world reconstructions could be undertaken, the vehicle accelerations before the crash had to be determined using crash reconstruction software. Human Vehicle Environment (HVE) was the crash reconstruction software used to determine the vehicle acceleration. This software and the methodology used is explained in detail in Chapter 5. Day et al. (2004, 2001, 1996, 1990, 1989) has presented literature of the validation of HVE software. Human, Vehicle, Environment software is suitable to reconstruct real-world side impact crashes at the range of different angles. This is due to the software having been validated for both front and side impacts at both 90° and oblique angles.

For each of the computational reconstructions the still images from the kinematic file has been presented. The presented output of the lateral 90 degree, near-side and far-side models is the lateral acceleration of the first thoracic vertebrae (T1) and the head. The output of the 45 degree, oblique near side model is lateral acceleration of the first thoracic vertebrae (T1) and the head and the frontal acceleration of the first thoracic vertebrae (T1) and the head.

Chapter IV
An Investigation into Real World Side Impacts Resulting in Low Severity Neck Injuries

4.1 Introduction

An in-depth investigation into the real world crash data from two databases, the Monash University Accident Research Centre database (Australia) and the Loughborough University Co-operative Crash Injury Study (UK) has been presented in this Chapter. This investigation works to identify the common factors associated with low severity neck injury in side impacts and compares these to those factors presented in the Chapter 2 Literature Review. For the purpose of this research, side impacts are defined as a vehicle collision where the predominant impact and damage is at the right side or the left side panels of the vehicle. The definition of low severity impact is that identified as level 1 neck injury using the Abbreviated Injury Scale (AAAM 1985). These neck injuries result in a strain or sprain of the neck, although there is no dislocation or fracture. Understanding the common factors associated with this injury provides a base of knowledge, to understand the injury and allows for further investigation in the following chapters. This knowledge assisted with the selection of six real world crash cases, identified as typical, that have been investigated in detail in Chapter 6. While this chapter focuses on the factors associated with the types of crashes and occupants injured, the later chapters in this thesis, focuses on the research into the occupant's kinematics during cases where an injury was reported. This chapter is structured with a methodology, results section concluded with a summary of findings.

4.2 Method

The published research presented in Chapter 2 reported that there are a number of common factors for occupant's receiving a low severity neck injury. These factors include gender, age, the seating position in vehicle, angle of impact, severity of impact and head contact. Other factors within the vehicle such as seat belt use, airbag deployment and the head restraint are also shown to have an influence on injury. In this study the data from two accident databases have been used to investigate the factors associated with low severity neck injury from side impacts. Australian data from Monash University Accident Research Centre (MUARC) accident database has been presented with data from the United Kingdom, Loughborough University Co-operative Crash Injury Study (CCIS). The database from the UK is larger than the Australian database, and has been included in this study to supplement the Australian results. The results from each of the databases have been presented separately in the following section, as each data base has slightly different criteria for collecting information.

Initially all of the cases in the databases were sorted by their type of impact. Following this, all of the cases where an occupant received a low severity neck injury were sorted by the type of impact. Doing this allowed for comparison between this data and the findings reported in existing literature previously presented. The results presented in this chapter were prepared in the following manner;

1. The side impact cases from each database were extracted and sorted based on the occupant's injuries.
2. Occupants with a low severity neck injury were grouped together to create a low severity neck injury group.
3. The occupants without a low severity neck injury were sorted into groups according to their severity of maximum injury received these were categorised a no injury and MAIS0 to MAIS 6 (fatal).

A description of the two databases has been presented in the following two subsections, this methodology also describes how the data has been sorted, analysed and presented in this study.

4.2.1 Australian Data- Monash University Accident Research Centre:

The Australian data consists of crash cases from the Monash University Accident Research Centre (MUARC) accident database. The cases are sourced from members of the community, reporting the crashes as a response to advertising. If a reported case meets the collection criteria of the database, it is investigated and included in the database. The collection criteria of the MUARC database are listed below.

- Only particular impact directions are of interest to the Monash University Accident Research Centre to meet the needs of their research projects. Front and side impacts are predominant cases collected with very few rear impacts collected. Side-swipes, with the damage to the left or right side of the vehicle, which has occurred from an impact that is predominantly from a frontal direction are not collected.
- Only particular makes and models of vehicle (for this study predominantly Holden Commodore and Ford Falcon) were collected, with an age of less than 7 years at the time of crash.
- The vehicle must be occupied during the impact, although occupants do not have to be injured for the case to be included.

At the time of this study, crash and injury data for 671 adult occupants were available. There were two main parts to each accident case: a detailed medical report and vehicle inspector's report. The data collected in each of these reports have been described in detail in Section 4.2.3.

4.2.2 UK Data- Loughborough University Co-operative Crash Injury Study

The UK accident data are from Loughborough University Co-operative Crash Injury Study (CCIS). Cases were collected only from a predetermined geographical region. The set of criteria to be met for a crash to be included in this database are different to the Australian database and are listed below;

- Vehicle damage needs to be extensive enough to require towing from the accident scene.
- A minimum of one occupant to be injured.
- All impact directions are considered.

The only collection criteria that is similar to the Australian data is that the case vehicle could not be older than 7 years. This database is a much larger than the Australian database with up to 1600 cases being collected each year (from 1983- 2002). The data collection within each case is similar to that of the Australian data with detailed medical information and a vehicle inspection, and is also discussed in detail in Section 4.2.3.

4.2.3 Case Data Collection

This section outlines how the date from each individual crash case was collected for both the Australian and United Kingdom Databases. Once the cases were identified to be suitable for inclusion, the specific information collected from each of the databases is the same. There are two main parts to a case, the medical data and the vehicle inspection data. The medical data is collected during an interview between the vehicle occupants and a qualified nurse. During this interview, the occupant's personal details such as gender, age, height and weight (human factors data) recorded as well as all injuries sustained in the accident (injury data).

All injury data is coded according to the Abbreviated Injury Scale (AIS) (AAAM 1985). Using the abbreviated injury scale allows a code to be applied to each injury, taking into account the location, type and severity of the injury. This severity ranking identifies the potential of the injury to cause death. There is five parts to an AIS injury code as illustrated in Figure 4.1. The first number indicates the body region injured. The body is divided into nine (with 6 as the code for the spine). The second number describes the type of anatomical structure injured. The third number describes the specific anatomical structure this, such as a burn or an abrasion. The following number indicates the level of injury. The final number following the decimal point is the injury severity rating. This final number ranks the injury's likelihood of resulting in death.

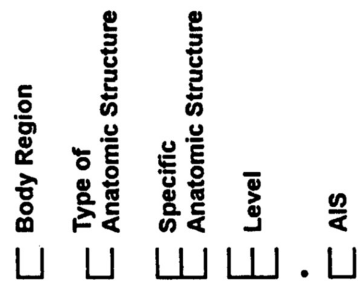


Figure 4.1 Overview of AIS code categories (AAAM 1985).

An example of the abbreviated injury scale is the code for neck strain acute with no fracture or dislocation (low severity neck injury): 640278.1. The 6 at the beginning of the code indicates that it is an injury to the spine. The second number 4 shows that it is an injury to the organs, including muscles and ligaments. The 02 reports that the location of injury is the cervical spine. The 78 is an identification number for this cervical spine injury. The .1 indicates that this injury is rated as a minor injury. The order of severity is described in Table 4.1.

Table 4.1: Injury severity codes according to Abbreviated Injury Score (AIS)

Code	Injury Severity
1	Minor
2	Moderate
3	Serious
4	Severe
5	Critical
6	Maximum (Death)
9	Unknown

The human factors data (also collected during the medical interview reported in this study) are gender, age, and the seating position in the vehicle were recorded. To provide consistency when comparing the crash cases, only adult cases (age 17+) were chosen in this study to eliminate any issues regarding the physical differences in the neck between adults and children. Cases where the occupant's gender was unreported were also removed. As an occupants gender has been reported as an important factor contributing to the likelihood of an occupant receiving a low severity neck injury.

The other injuries received by the occupants with a low severity neck injury have also been presented, as they are of interest and they may indicate contact with the interior of the vehicle or impacting vehicle. Knowing these contacts may allow assumptions about the occupant's kinematics to be made.

A detailed vehicle inspection for each case vehicle was undertaken by an experienced vehicle crash inspector. During this inspection, the following factors were recorded, side and angle of impact, impacting object, seat belt use, airbag deployment and Delta V. The side of impact and occupant position in the vehicle were combined to deduce either near or far side impacts as shown in Figure 4.2. The angle of impact was established from the crush measured on the vehicle. Angle of impact is presented in terms of a clock face, and is illustrated in Appendix 2.

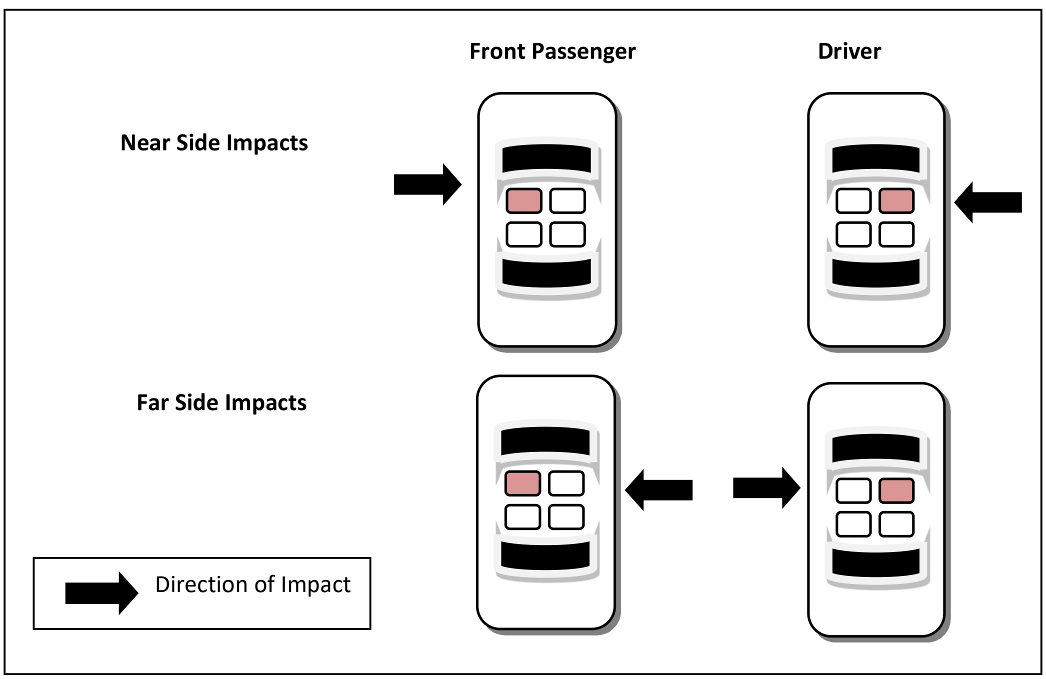


Figure 4.2: Diagram of near and far-side impacts.

The striking object for the Australian data and the UK data are grouped in slightly different categories. The Australian data classifies poles and trees as one category, whereas the UK data classifies all stationary objects as either narrow objects (less than 41cm) or wide objects (greater than 41cm), as suggested in Table 4.2.

Table 4.2: Comparison of categories for striking object

Australian Categories	UK Categories
Car	Car
Pole/Tree	Narrow object (<41 cm) Wide object (>41 cm)
4 WD/Truck	Light Goods Vehicle
Bus/Truck	Heavy Goods Vehicle
Not Applicable	Two Wheeler

Knowing the occupant's seatbelt use is important, as the incidence of low severity neck injury has been shown to increase since the introduction of seatbelts in vehicles (Galasko et al. 1993). The occupant is only reported to be wearing a seatbelt, if the evidence of seatbelt use, such a damage to the belt webbing was visible during the vehicle inspection. The occupant's self-reporting of seatbelt use alone was not used as an indicator of seatbelt use. Airbag deployment is described in three categories as, no airbag installed or airbag deployed/ not airbag deployment.

The Delta V has only been presented for the Australian data only. The Delta V was calculated using CRASH 3 software (National Highway Safety Administration, 1986). Using this method of calculation of Delta V, requires the same crush profile measurements that is used in equivalent barrier speed (EBS) it is also measured in km/h. Delta V can only be calculated when both vehicles involved in the accident are known and a crush profile for each has been created. It can also be measured when the crash involved a tree, pole or other stationary object. There are some limitations with delta V, as it cannot be calculated in a rollover accident and when the vehicle has sustained extreme damage.

4.2.4 Sample and Control Groups

This section discusses the ways in which the data from the Australian and UK databases have been analysed and presented in the following section. The Australian database consisted of 227 side impact. The UK database, being a larger database, had 1069 side impacts. The Australian and UK data sets have presented separately, although each of the datasets have been analysed in the same manner. All of the side impact cases were divided into three groups: those with only a low severity neck injury, those with no injury, and those with other injuries. The group with other injuries (but no low severity neck injury), was further divided according to their maximum severity of their injuries. Sorting the data in this manner provided a group of occupants with low severity neck injury to be compared to a number of other groups (controls) involved in impacts of varying severity.

4.3 Results

In the following sections the results from the analysis of the Australian data from Monash University Accident Research Centre (MUARC) accident database and from the United Kingdom Loughborough University Co-operative Crash Injury Study (CCIS) are presented. The results include information related to the direction of impact, injury severity, gender, age, side and angle of impact, impacting object, seatbelt use, airbag deployment, delta V and head contact. The results presented for each of the sample groups. It is important to note that the results are representative of each sample group and specific generalisations cannot be automatically made for the whole population due to the size of the samples and the entry criteria into the sample groups.

4.3.1 Direction of Impact for All Cases

The direction of impact data for all cases is presented for the Australian and UK databases, presented Figures 4.3a and 4.3b. Knowing the percentage of impacts for each of the databases allows for comparison between the two databases as well as highlighting any trends that may be specific to a particular database due to the method of data collection.

The direction of impacts for Australian database is shown in Figure 3.3a. Occupants involved in frontal impacts account for 60% for the database, with occupants involved in side impacts accounting for 38%. Rollover and rear impacts account for 1% each. Multiple impacts are not applicable in this database as impacts were categorised by their major impact. The main objective of the Australian database was to collect and investigate front and side impacts, this explains why these are the main impacts represented. Rear impacts and rollover cases were not a priority for collection, so they account for very few cases in the database.

The UK database contains a wider variety of cases than the Australian database. The results in Figure 4.3b show the direction of impact for cases in the UK database. Occupants involved in front impacts account for 44% of the database, those involved in side impacts account for 17%, rear impacts 4% and rollover 5%. Multiple impacts account for 30% of the impacts in the UK database. The difference in the side impacts from 38% in the Australian to 17 % in the UK data, can be explained by the presence of the other impact angle categories in the UK database that are not in the Australian database. In spite of the differences, the side impacts from the Australian and UK databases are still comparable for this study, as the same human factors and injury data was collected in each database. Side impact is an impact direction that was well represented in both databases.

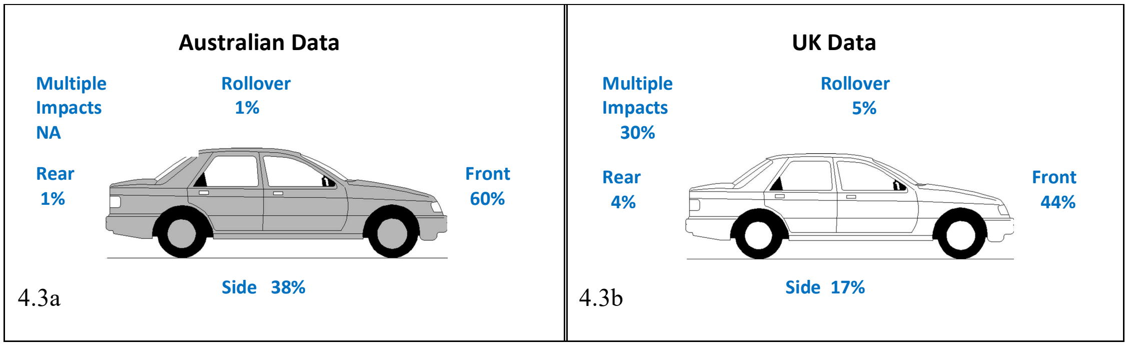


Figure 4.3a: Direction of impact Australian database
Figure 4.3b: Direction of impact UK database

4.3.2: Direction of Impact for Low Severity Neck Injury Cases

All of the cases where an occupant received a low severity neck injury, were extracted from each database and presented according to impact type in Figures 4.4a and 4.4b. The Australian data in Figure 4.4a shows that frontal impacts resulted in 60.5% of occupants receiving a low severity neck injury while side impacts resulted in 39.5 %. The low severity neck injury from rear, rollover and multiple impacts cannot be reported as these impact types are not represented in the database.

The percentage of cases from the UK database where an occupant received a low severity neck injury is shown in Figure 3.4b. It is shown that front impacts resulted in 45% of occupants receiving a low severity neck injury while side impacts resulted in 18 % of occupants receiving a low severity neck injury. Occupants receiving a low severity neck injury in a rear impact account for 4% of the database and occupants involved in rollovers account for 5%. Occupants involved in multiple impacts make up 29% of those receiving a low severity neck injury.

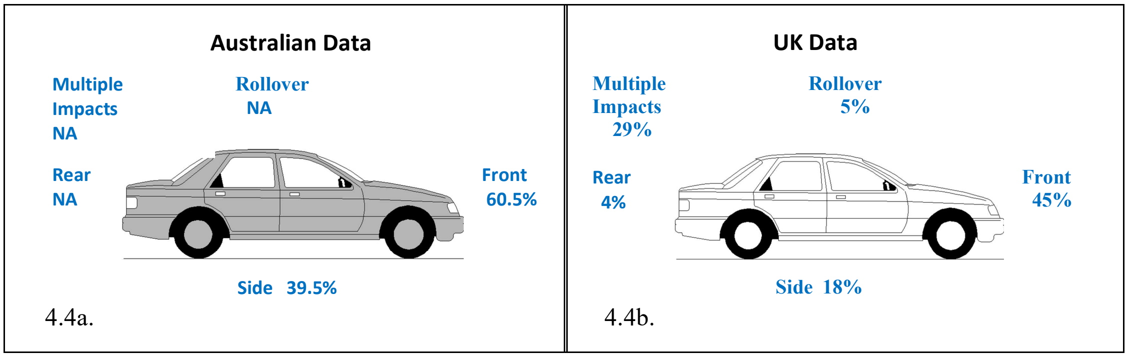


Figure 4.4a: Direction of impacts producing a low severity neck injury Australian data
Figure 4.4b: Direction of impacts producing a low severity neck injury UK data

The incidence of injury in the UK data compares more closely to the values reported in the literature as all impact types are represented. The literature shows that low severity neck injury from side impact ranges from 8.8% to 24% (Teaming et al. 1998, Jakobsson et al. 1998, Morris et al. 1996, von Koch et al. 1995). The UK data in this study falls within this range.

The results shown for the Australian data show a higher incidence of injury than what has been reported in literature (Teaming et al. 1998, Jakobsson et al. 1998, Morris et al., 1996, Von Koch et al. 1995). This is expected due to absence of rear, multiple and rollover impact in this database.

4.3.3 Injury Severity Groups

The side impact cases have been extracted from each database to create a collection of cases for investigation into the factors associated with low severity neck injury. All of the side impact cases that resulted in a low severity neck injury, were grouped together to make the low severity neck injury group. The side impact cases where an occupant did not receive a low severity neck injury were further grouped according to the severity of the maximum injury according to the Abbreviated Injury Scale (AIS). These groups have been titled according to the maximum injury received by the occupant MAIS 0 (no injuries) to MAIS 3+. The MAIS 3+ group contains all occupants with MAIS 3 to MAIS 6 injuries as each of these groups alone do not contain enough cases for comparison.

The percentage of occupants in each of the groups has been presented in Figures 4.5a and 4.5b. Occupants in the Australian database (Figure 3.5a) involved in a side impact received a low severity neck injury in 21% of the cases and 26% of occupants received no injuries. Occupants receiving a maximum AIS 1 injury (not to the neck) accounted for 38% of side impacts and 8% of occupants receiving a maximum AIS 2 injury. Occupants who received a maximum AIS 3 or greater injury, make up 7% of the database.

The UK database (Figure 4.5b) shows a higher percentage (33%) of occupants receiving a low severity neck injury than the Australian database. Those uninjured (MAIS 0) account for 5% of the side impacts. Occupants receiving a MAIS 1 injuries accounted for 31% of the side impacts, and occupants receiving MAIS 2 injuries accounted for 12%. Occupants receiving MAIS 3+, make up 19% of the sample.

The difference between the two databases with respect to the MAIS 0 values and the MAIS 3+ can be explained by the UK data base having the prerequisite that the impact was serious enough to be reported to police and required towing away. The Australian database does not have this prerequisite. The outcome of this difference is less uninjured cases in the UK database, but a greater number of cases with severe injuries.

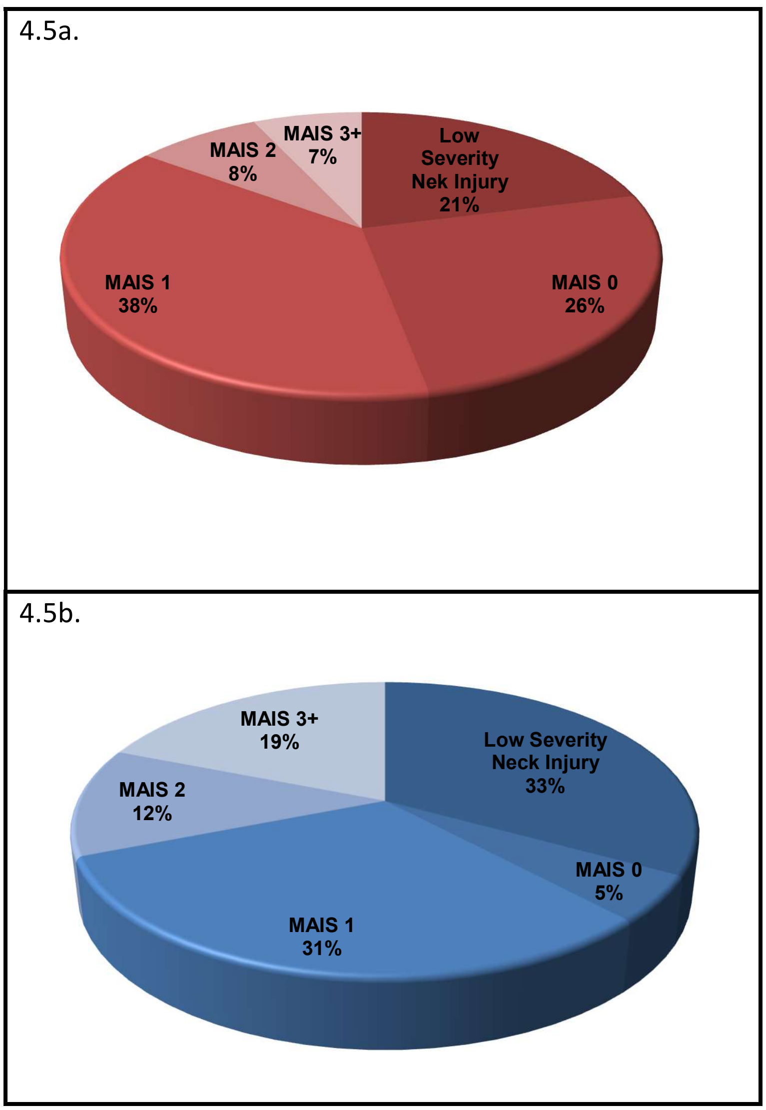


Figure 4.5a: Occupant injury severity for all side impacts Australian data
Figure 4.5b: Occupant injury severity for all side impacts UK data

4.3.4 Gender

The division of gender for Australian occupants involved in a side impact, is 60% males and 40% females. The UK data shows similar gender patterns for occupants involved in a side impact, 58% males and 42% females. The Australian data shows that 20% of the males and 22% of females involved in a side impact received a low severity neck injury. The UK data shows a higher number of occupants receiving a low severity neck injury with 27% of males and 39% of females. The UK data shows a difference between the genders for occupants reporting a low severity neck injury compared to the Australian data. This may be due to the small size of the Australian data sample. Females in the UK database are shown to receive more neck injury from side impacts than males. The literature supports this finding showing that gender is one of the most important factors for low severity neck injury and that females are more likely to be affected by low severity neck injury (Teaming et al. 1998, Morris et al., 1996, Fildes et al. 1995, Gibson et al. 2000, von Koch et al. 1995, Bring et al. 1996, Jakobsson et al. 2000.

4.3.5 Age

The average age of occupants with low severity neck injury has been presented in Table 9 against the average age of occupants from the MAIS0-MAIS3+ injury groups (where there was no neck injury). The average age for each group has been separated by gender. The standard deviation is presented in brackets. The Australian data has been presented in the first two columns and the UK data has been presented in the following two columns.

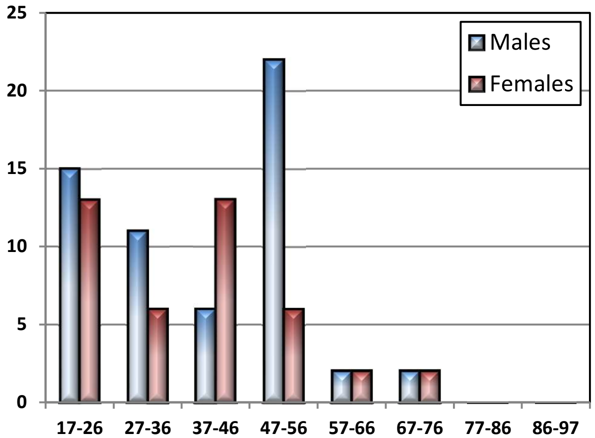
Table 4.3: Average age of occupants involved in a side impact by injury severity and gender

	Australia		UK	
	Male	**Female**	**Male**	**Female**
Low Severity Neck Injury	39 (16)	38 (15)	34 (16)	37 (16)
MAIS 0	40 (16)	42 (19)	41 (20)	40 (15)
MAIS 1	38 (15)	41 (15)	39 (19)	42 (18)
MAIS 2	35 (12)	41 (20)	39 (20)	43 (20)
MAIS 3 +	39 (18)	34 (11)	36 (19)	41 (20)

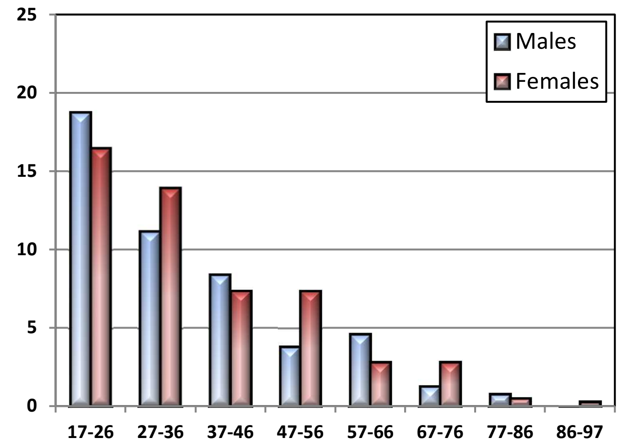
The average age of the Australian occupants receiving a LSNI for side impact was 39 years (S.D. 16) for males and 38 years (S.D. 15) for females. The average age of UK occupants receiving a LSNI is slightly lower for males and 34 years (S.D. 16). UK females have a similar age to Australian occupants at 37 years (S.D. 16).The high standard deviation across the genders in both data bases (S.D. 15-16) indicates great variation among the ages of the occupants injured. Due to this it is difficult to identify trends in the age data using only average.

When comparing the LSNI sample group to other injury groups for the Australian database there is also little difference. The average age of uninjured males is 40 (S.D. 16) and uninjured females 42 (S.D. 19). The results from the UK database uninjured occupants were slightly older than those receiving a low severity neck injury. Uninjured males averaged an age of 41 years (S.D. 20) and females average 40 years (S.D. 15). There is little difference in the average age of male occupants injured in a side impact with no LSNI in both databases. In the Australian databases there is only a 4 year difference between the occupants in the MAIS 1 (38 years), MAIS 2 (35 years) and MAIS 3 (39 years). This is similar for UK database. There is only a 3 year variation in the average age for injured male occupants, MAIS 1 (39 years), MAIS 2 (39 years), MAIS 3 (36 years). There is a slightly greater range in average age of injured females without a LSNI in the Australian database. Females receiving MAIS 3+ injuries were on average 6 years younger (34 years) than occupants receiving a MAIS 3 injuries (41 years). This is not evident in the UK database as injured female occupants have a similar age at all injury severities in MAIS 1 (42 years), MAIS 2 (43 years) and MAIS 3+ (41 years).

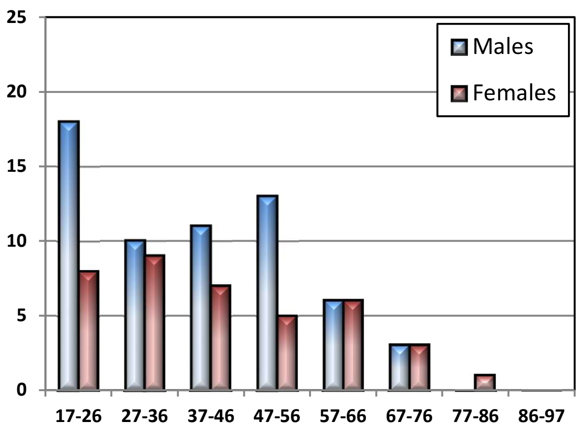
To address the inconclusive nature of the previous data, the data has been re-grouped in individual age brackets for occupants who received low severity neck injury and for occupants involved in minor side impacts (Occupants receiving a MAIS 2 or less with no neck injury). These results have been presented in Figure 4.6, the Australian data is on the left side of the page and the UK data is on the right side of the page. The Figures 4.6a to 4.6d show the distribution of age for occupants with a low severity neck injury against occupants with injuries less than or equal to MAIS 2 (with no low severity neck injury). This data presentation now indicates that particular age groups are more likely to be affected by low severity neck injury.



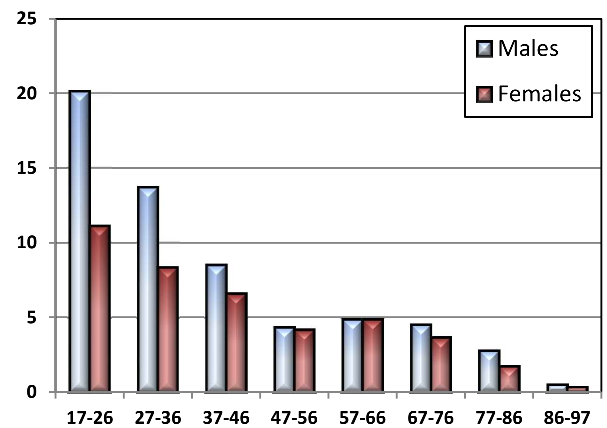
a.



c.



b.



d.

Figure 4.6a - Distribution of age for occupants with low severity neck injury- Australia
Figure 4.6b - Distribution of age for occupants with injuries up to MAIS 2- Australia

Figure 4.6c- Distribution of age for occupants with low severity neck injury- UK
Figure 4.6 d -Distribution of age for occupants with injuries up to MAIS 2- UK

The Australian data in Figure 4.6a shows that occupants aged between17-26 and occupants aged between 47-56 accounted for more than half of those receiving a low severity neck injury from a side impact. There is a similar number of occupants in the 27-36 and 37-46 age brackets. Few occupants over 57 are shown to receive a low severity neck injury and there are no occupants in the 77-86. When considering age and gender, males aged between 47 –56 years old received the most low severity neck injury. Females aged between 17-26 and 37-46 received the most low severity neck injury. Equal numbers of males and females injured were aged between 57-66 and 67-76.

The Australian data for age and gender for the combined controls group MAIS0-MAIS2, have been presented in Figure 4.6b. As with the LSNI group, males aged 17-26 have the highest representation, followed by males in the 47-56 age bracket. Females in the aged 17-26 in the control group have approximately half of the representation compared to females in the low severity neck injury group. This suggests that low severity neck injury is a greater concern for females of this age group. This is also case for females aged 37-46 years.

The UK age and gender data for the low severity neck injury group is presented in Figure 4.6c and shows a different pattern to that of Australian data. Generally, as the groups increase in age, the percentage of occupants in each category generally decreased. Within each of the age groups there are not large differences between males and females. When comparing the age groups with a low severity neck injury more males are aged between 17-26 and more females are aged 17-26 and 27-36 received the highest incidence of injury. This is a slight variation to the Australian database that shows older females 37-46 are also most likely to be injured.

The age and gender data for the combined group MAIS0-MAIS2 in Figure 4.6d shows a similar pattern to the UK data. As with the Low Severity Neck Injury the incidence of injury generally decreases with age. Males aged 17-26 and 27-36 have a much greater incidence of injury than males of any other age categories.

The finding in both data bases that males aged 17-26 have a high incidence of injury than females is this category is against what is reported in the literature. Teamming et al. (1998) found that young females 18-27 have a greater risk of low severity neck.

4.3.6 Occupant Seating Position and Gender

Occupant seating position has been presented with respect to the occupant's gender for the low severity neck injury group (Table 4.4) to investigate the influence on the incidence injury.

Table 4.4: Occupant position and gender of occupants with a low severity neck injury

	Australia			UK		
	Male	Female	Total	Male	Female	Total
Driver	26	10	36 (77%)	158	131	289 (73 %)
Front Passenger	1	8	9 (19 %)	22	55	77 (20%)
Left Rear Passenger	1	1	2 (4%)	4	5	9 (2%)
Right Rear Passenger	0	0	0	3	8	11 (3%)
Rear Centre Passenger	0	0	0	3	0	3 (<1%)
Unknown	-	-	-	-	-	6 (1%)

The Australian data, on the left side of Table 4.4 shows that 77% of occupants receiving a low severity neck injury from a side impact, were drivers. Front passengers account for approximately one fifth (19%) of the occupants. The UK data shows similar results to the Australian data, with 73% of drivers and 20 % being front passengers receiving a low severity neck injury. In both databases, more drivers were males and more females were front passengers. Databases have very few rear seat passengers receiving a low severity neck injury.

Australian and UK data drivers are shown to experience the highest incidence of low severity neck injury as compared to any other vehicle position. Both databases show that drivers were more likely to be males, and front passengers were more likely to be females. Very few occupants in either database were rear seated passengers. Literature is divided as to the influence that seating position has on low severity neck injury. Lösvund et al. (1988) shows that front row occupants (driver and front passenger) are injured more than rear seat passengers. This supports the findings in this study. Jakobsson et al. (2000) shows that the risk of a particular seating position differs between the genders. Males are shown to be at greater risk of injury when they are drivers and females are at greater risk when they are passengers

4.3.7 Side and Angle of Impact

In this section the side of impact (near-side or far-side) has been presented with respect to the angle of impact (position on clock face). This data also indicates if the occupant position in the vehicle and the influence of impact direction. The data has been divided into four groups: low severity neck injury, MAIS 0, MAIS 1&2 (combined) and MAIS 3+. The Australian data has been presented in Figures 4.7a-d, and the UK data has been presented in Figures 4.7e-h.

Australian occupants (Figure 4.7a) with low severity neck injury are more likely to be involved in a near side impact. This is in contrast with the group of occupants who received no injury -MAIS 0 (Figure 4.7b) who were more likely to be involved in a far side impact. The results for the MAIS 1 & 2 combined group (Figure 4.7c) and the MAIS 3+ group (Figure 4.7d), show that occupants in these injury groups are most likely to be involved in a near side impact.

The Australian data shows that the most common angles of impact for the low severity neck injury group were, in descending order 10 o'clock, 2 o'clock, 3 o'clock and 1 o'clock near side impacts. This differs from the most common impact directions for the uninjured group (MAIS 0) with 1 o'clock (near), 9 o'clock (far), 10 o'clock (far) and 11 o'clock (far) side impacts. The most common impact direction for the MAIS 1&2 combined group were 1 o'clock, 2 o'clock, 3 o'clock near-side and 10 o'clock far-side impacts. The impact direction for the side impacts that produce the most severe injuries (MAIS 3+) showed an equal number of 3 o'clock near-side and 9 o'clock far-side as the most frequently occurring.

The UK data is presented in Figures 4.7e-h. The data for the low severity neck injury group in Figure 4.7e shows that (as same the Australian data) more side impacts that lead to this injury are near side impacts, although the common impact directions differ. The most common impact directions are 12 o'clock, 2 o'clock and 3 o'clock near-side impacts. The results for MAIS 0 group (Figure 4.7f) also differs to those in the Australian data. Uninjured occupants are equally likely to be involved in a near or far side impact. The direction of impact also differs between the UK and Australian databases with 2 o'clock near-side, 9 o'clock and 10 o'clock far-side impacts being the most common. The results for the MAIS1 & 2 combined group and the MAIS 3+ group are the same as that of the Australian data, with more near-side impacts producing injuries of higher severity. The impact directions are the same between the two databases with the most common impacts for the MAIS1 & 2 combined group as 1 o'clock, 2 o'clock, 3 o'clock and 10 o'clock near side impact. The impact directions vary slightly between the two databases for the MAIS 3+ group with the most common impact directions 9 o'clock (far), 3 o'clock and 2 o'clock (near) side impacts.

These findings are not consistent with Jakobsson et al. (2000) who show that near side impacts at 7-8 o'clock for left sided drivers (Swedish) and left sided passengers and 4-5 o'clock for right front and right rear seated passengers has the highest risk of low severity neck injury. These impact angles are shown to rarely occur in the Australian and UK databases. It is difficult to determine the cause of the difference between the Australian and UK data and that reported in the literature regarding Swedish data with the information provided. This is due to the unique nature of 7-8 o'clock and 4-5 o'clock side impacts with the impacting vehicle striking from the rear to the front of the vehicle. The cause could range from road or intersection design to driver behaviour/ error at intersections. It is important to note that the measure of angle of impact in both the Australian and UK sample groups is determined from an appraisal undertaken by an accident investigator, this method can be subjective as it requires the investigators interpretation of the crash damage.

Australian Data

Low Severity Neck Injury

MAIS 0

MAIS 1 & 2

MAIS 3

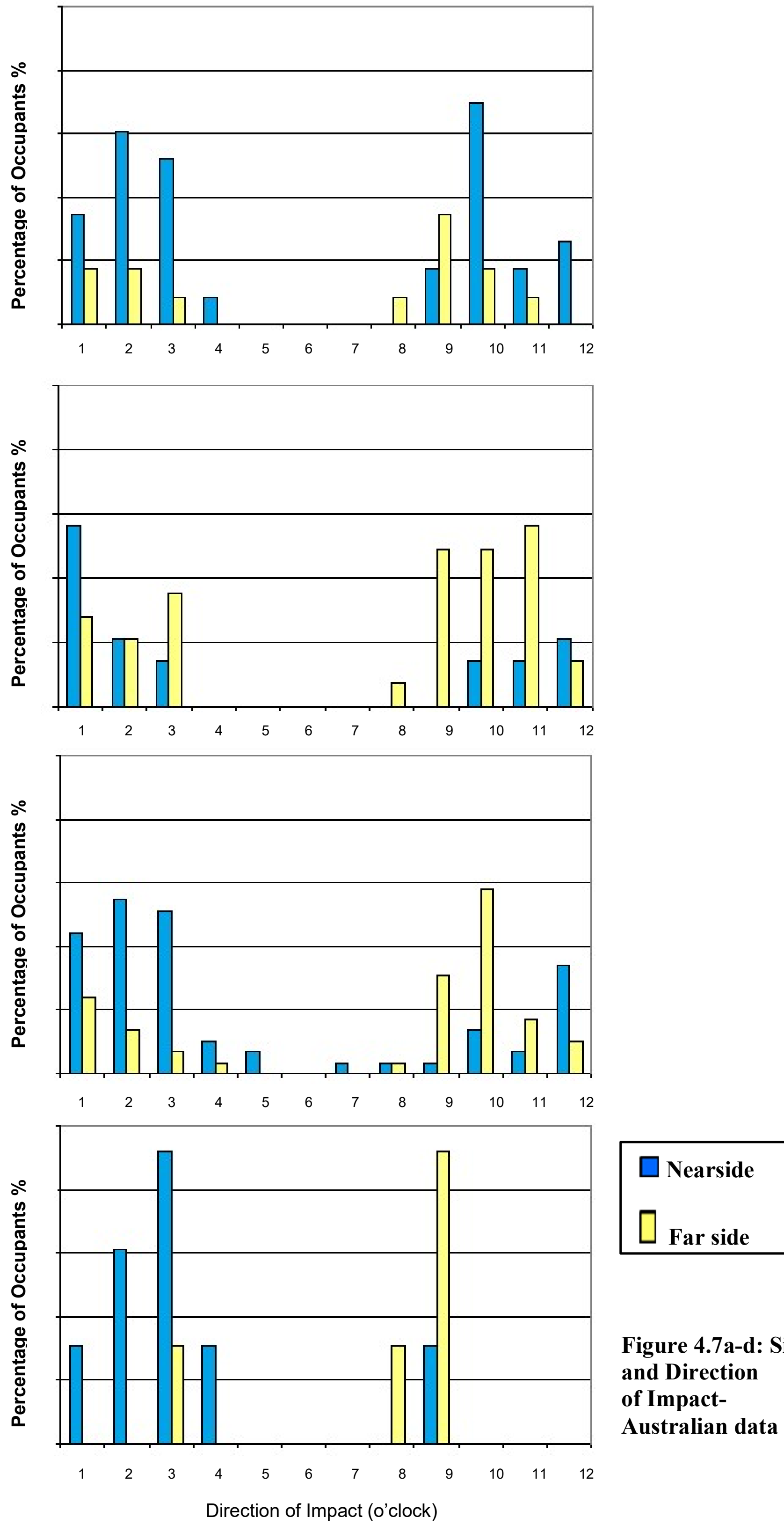


Figure 4.7a-d: Side and Direction of Impact- Australian data

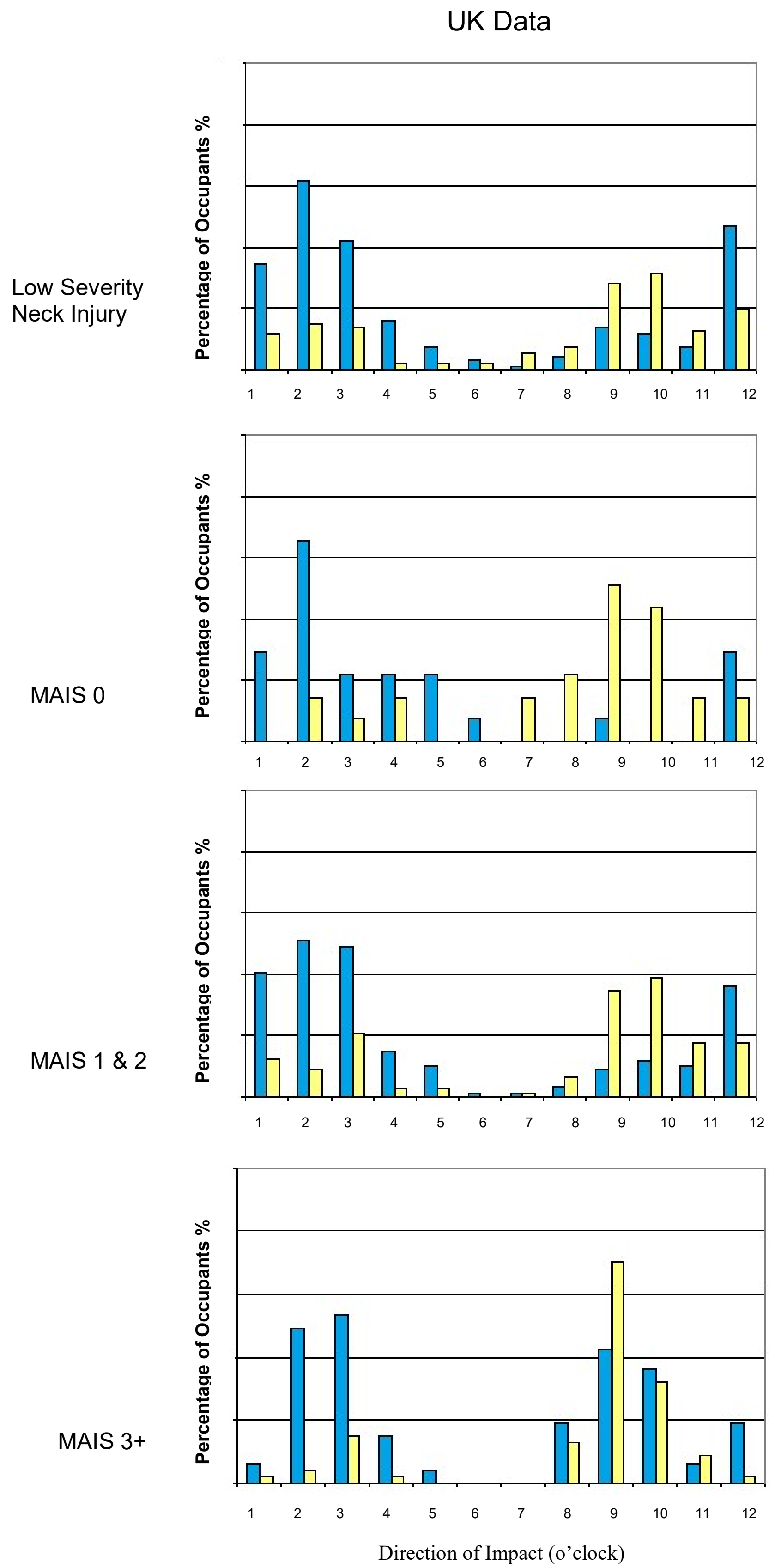


Figure 4.7e-h: Side and Direction of Impact- Australian data

4.3.8 Impacting Object

The type of impacting object for the for the low severity neck injury cases for the Australian data (Figure 4.8a) and UK databases (Figure 4.8b) have been presented for comparison. The Australian and UK databases have different categories for classifying impacting object as shown earlier in Table 4.2 although they are comparable.

The results for the Australian data show that cars are the most common striking object at 64% and make up almost two thirds of the sample. Heavier vehicles are represented in the sample with four wheel drives and vans accounting for 8% and trucks and busses 13%. Pole and tree impacts cause 15% of side impacts low severity neck injuries.

The UK data shows a similar pattern of impacting objects as the Australian data. Cars are shown to be the most common impacting objects with 64% in cases where an occupant receives a low severity neck injury. The main difference in the classification of the UK data is the classification of stationary objects as either wide objects (>41cms) or narrow objects (<41cms). Impacts with fixed objects account for 15% of impacts resulting in a low severity neck injury. This is the same as the occurrence of impacts with pole/trees in the Australian data. Impacts with heavy goods vehicles account for 12 % this number is similar number to the bus/truck data from Australia. Impacts with light goods vehicles are 6% of impacts. This number is similar to the number of impacts with 4WD/Vans in the Australian dataset. The UK data has one additional category that is not represented in the Australian data, two wheeled vehicles, accounting for 2% of low severity neck injuries.

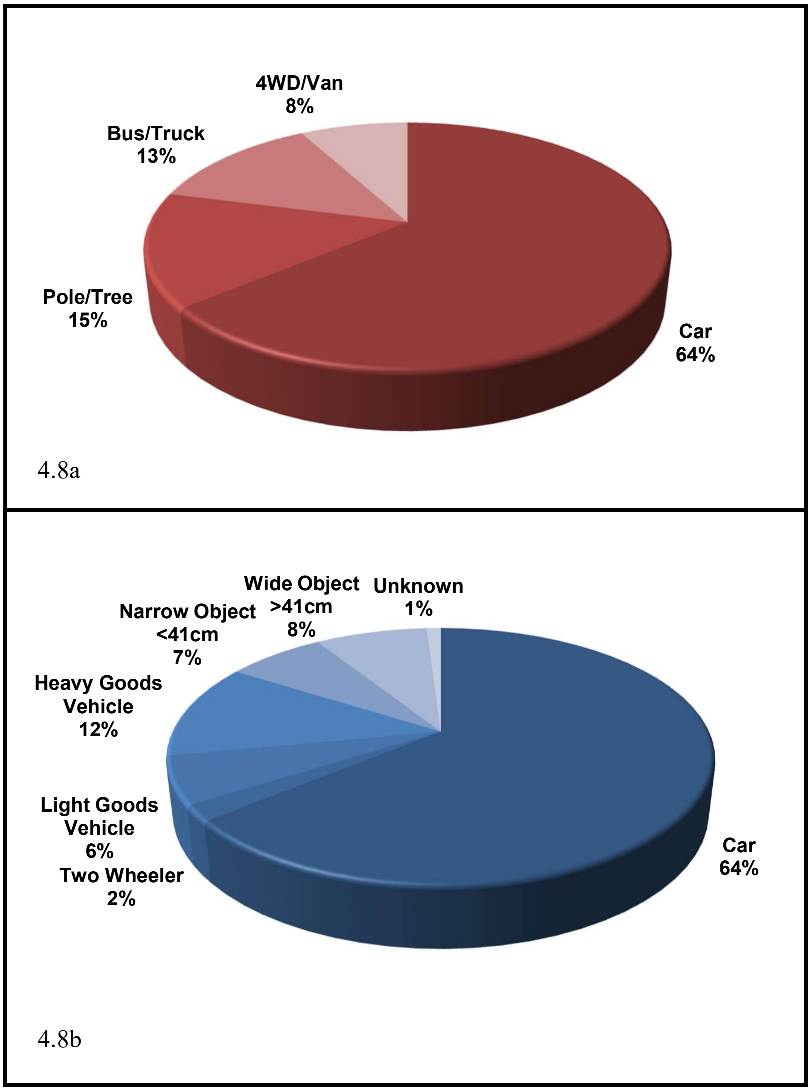


Figure 4.8: The impacting object for low severity neck injury cases for Australian data (a) and UK data (b).

4.3.9 Seatbelt Use

The data regarding the use of seatbelts by occupants receiving a low severity neck injury is presented in Figure 4.9. This data is only available for the Australian cases. The confirmed seat belt use is shown to be 89%. It is shown that for 7% of occupants it is probable that there was seatbelt use although, physical evidence is in conclusive. Four per cent of occupants claimed seatbelt use although there was no physical evidence to support their claim. There were no occupants who were confirmed not to use their seatbelt.

It is difficult to comment on the high use of seatbelts without a control group. Although, the high use of seatbelts among injured occupants in this sample is consistent with what is reported literature regarding rear and front impacts. It is reported that low severity neck injury is more likely to occur when the occupant is restrained by a seatbelt Galasko et al. (1993), von Koch et al. (1995), Morris et al. (1996).

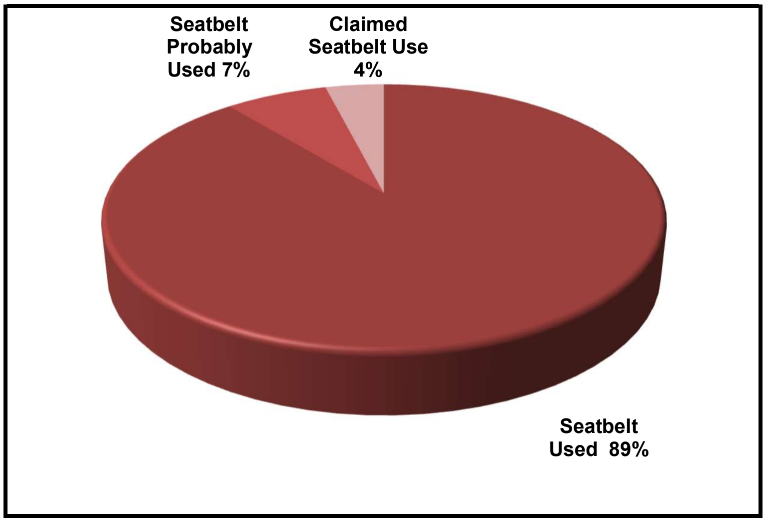


Figure 4.9: Seatbelt use (Australian data only)

4.3.10 Airbag Deployment

The data regarding the deployment of airbags during the impacts that result in a low severity neck injury is presented in Figure 4.10. This data is only available for the Australian cases. The data presented is for frontal airbags only as none of the case vehicles had side airbags installed. It is difficult to conclude the effectiveness of frontal airbags from the data available. For occupants receiving a LSNI in 38% of cases no airbag was installed. The airbag deployed in 36% of cases and did not deploy in 26% of cases.

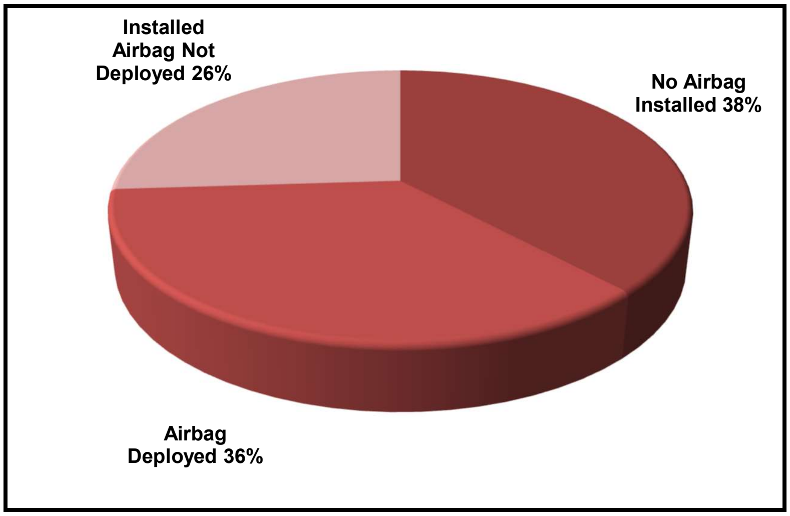


Figure 4.10: Airbag deployment (Australian Data Only)

The literature regarding the effectiveness of airbags in reducing the likelihood of receiving a low severity neck injury shows conflicting findings. Kulgren et al. (2000) and Morris et al. (2000) found that airbags reduce low severity neck injury. Otte et al. (1995) found that airbags increased the incidence low severity neck injury.

4.3.11 Delta V and Low Severity Neck Injury (Australian Data Only)

The delta V for the side impacts that lead to a low severity neck injury has been presented in Figure 4.11.

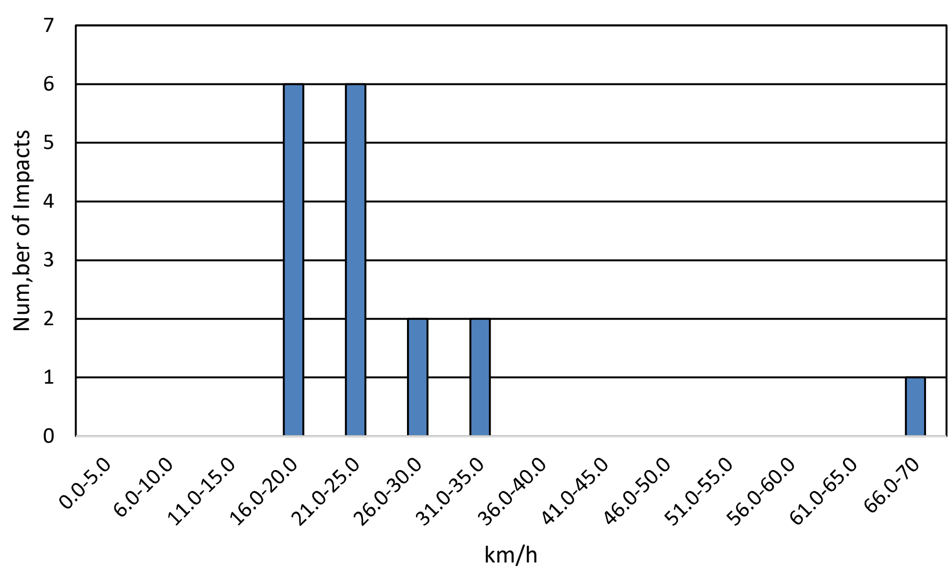


Figure 4.11 Delta V Low Severity Neck Injury- Australian Data Only

The range of delta V for side impacts is predominately between 16km/h and 35km/h, with one case occurring at each of 41-45km/h and 66-70km/h. Most impacts resulting in LSNI have a delta V of between 16 and 25 km/h. These results are consistent with the results for rear impact, presented by Hell et al. (1998). Hell et al. (1998) shows that majority of low severity neck injury from rear impacts occur at a delta V smaller than 25 km/h. It is important to note that due to the nature of side impacts and that work in this area is limited, delta V may not be the best descriptor of crash severity in a side impact, but currently no other comparable measure exists for side impacts.

4.3.12 Other Injuries (Occupants with a Low Severity Neck Injury)

The other injuries received by occupants with a low severity neck injury for the Australian and UK databases has been presented in Table 3.5. Understanding the other injuries received by an occupant in an impact with a LSNI provides an insight into the occupant's mechanics in a crash. The Australian data shows that no occupant with a low severity neck injury received any other injury above an (Abbreviated Injury Score) AIS 2. Occupants with a low severity neck injury with no other injuries only accounted for 21%. Most occupants (64%) with a low severity neck injury also received at least one other MAIS 1 injury. Occupants who received more severe injuries (at least one AIS 2 injury) made up 15% of the cases.

The UK data shows that occupants with a low severity neck injury have a greater range of more severe other injuries from MAIS 1 to MAIS 6. Occupants who only received a low severity neck injury and no other injury were appoximately one fifth of the cases (similar to the number in the Australian sample). Most occupants (63%) with a low severity neck injury received at least one other AIS1 injury. Occupants with other injuries up to maximum a severity of AIS2 with their low severity neck injury are 11% of the sample. Very few occupants received other injuries of a MAIS3 or above with their low severity neck injury. Only 3% of occupants received injuries with a maximum severity of AIS3, 2% of occupants received injuries up to AIS4. One per cent or less of occupants received other injuries of maximum AIS5 or AIS6.

Table 4.5: Percentage of Injury Severity of Other Injuries for Occupants with Low Severity Neck Injury

Australian Data		UK Data	
LSNI Only	21%	LSNI Only	19%
MAIS 1 (with LSNI)	64%	MAIS 1, (with LSNI)	63%
MAIS 2 (with LSNI)	15%	MAIS 2, (with LSNI)	11%
MAIS 3, (with LSNI)	0%	MAIS 3, (with LSNI)	3%
MAIS 4, (with LSNI)	0%	MAIS 4, (with LSNI)	2%
MAIS 5, (with LSNI)	0%	MAIS 5, (with LSNI)	1%
MAIS 6, (with LSNI)	0%	MAIS 6, (with LSNI)	1%

4.3.13 Site of Injuries for Occupants with a Low Severity Neck Injury

The data for the injuries received by occupants with a LSNI presented in the previous section has been sorted by body segments (i.e. head, chest and thorax). The Australian data is presented in Figure 4.12a and UK data is presented Figure 4.12b. The data has been presented in a visual format labelling the body part and the percentage of injuries for easy comparison.

Australian occupants with a LSNI received additional head injuries at a rate of 19%. In 26% of cases occupant received a thoracic injury as well as a LSNI. Fewer occupants also received an abdominal injury at 8%. Most head injuries received were a MAIS 1 (15%) with 4% of injuries at MAIS2. All thoracic injuries (26%) received by occupants were minor at MAIS 1. Abdominal injuries received were divided between MAIS 1 (4%) and MAIS 2 (4%).

Occupants in the UK database with a LSNI experience a higher incidence of head injury at 42%. This is also the case for thoracic injuries (44%) and abdominal injuries (24%). The other injuries experienced by occupants in the UK database also had a greater range of severity. Head injuries range from MAIS 1-MAIS 5 with 30 % of occupants experiencing MAIS1.

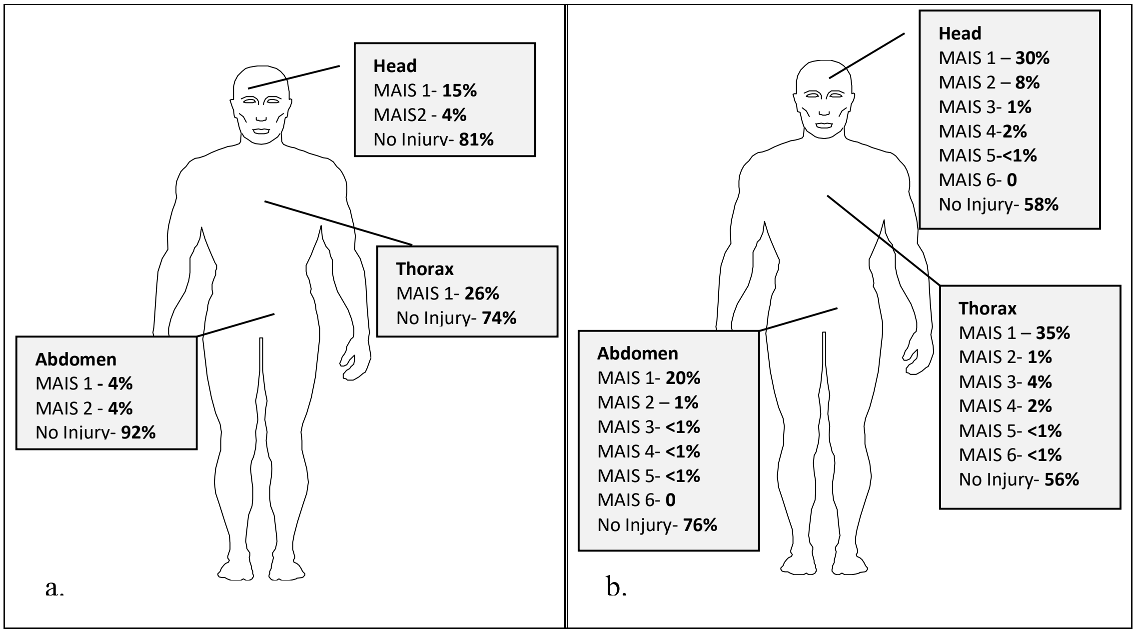


Figure 4.12: Location of other injuries received by occupants with a low severity neck injury

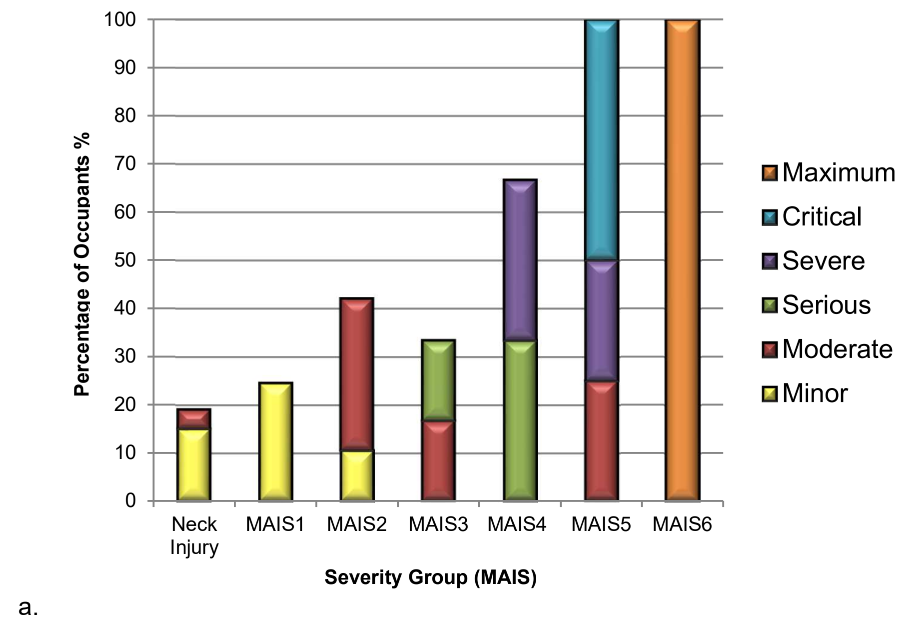
4.3.14 Head Injury

To compare the incidence of head injury in all side impacts, the results for the low severity group have been presented against all of the other severity groups. The MAIS 3+ group has been separated into MAIS 3, MAIS 4, MAIS 5 and MAIS 6 groups for this section to account for the variation in severity. These aspects are particularly important to keep separated for the head injury data as the outcomes of head injuries of these severities are very different, with some injuries resulting in permanent disability.

Head injury is an indicator of a head contact with either the interior of the vehicle or the impacting object. Therefore, it is an indicator of the occupant's kinematics in the impact. Literature shows that the head/neck kinematics are different when there is a head contact in an impact (Ono et al. 2002) compared to when there is no head contact. The data has been grouped by the most severe head injury received by the occupants with each of the severity group presented in Figure 4.13. The Australian data is presented in the top Figure 4.13a and the UK data is presented below in Figure 4.13b.

The Australian data generally shows that as an occupant's injuries become more severe there is a greater incidence of head injuries. Less than 20% of occupants with a low severity neck injury also received a head injury. The majority of these head injuries were either AIS 1 with a smaller amount AIS 2. Slightly over 20% of occupants in the lowest injury group category MAIS 1 control group received a head injury. Almost twice as many occupants in the next lowest injury group MAIS 2 (42%), received a head injury. Approximately three quarters of the occupants in this group received a maximum head injury rated moderate (AIS 2) with a quarter being minor (AIS1). Less than 40% of occupants in the first middle injury severity group MAIS3 received a head injury. The maximum head injuries in this group were divided between moderate (AIS 2 and serious AIS 3). Over 60% of occupants in the second middle injury group MAIS 4 received a head injury. These head injuries were equally divided between serious (AIS3) and severe (AIS 4). All of the occupants in the highest two injury groups MAIS 5 and MAIS 6 received a head injury. Occupants in the MAIS 5 group received a maximum head injury of either a serious (AIS 3), severe (AIS 4), or critical (AIS 5) head injury. All occupants in the MAIS 6 group received the maximum head injury. It is important to note that the MAIS 5 and MAIS 6 groups have a smaller sample size than that of the other groups. Hence, these results may not be representative of the whole population.

The UK head injury data also shows that as an occupant's injuries are more severe the number of occupants within that injury group with a head injury is greater. More occupants (over 40%) in the low severity impact group from the UK database also received a head injury than those in the Australian database. These head injuries are of a greater range of severity than the Australian data, although most occupants received a minor head injury (AIS 1) similar to the Australian occupants. More than half of the occupants in the lowest injury severity group MAIS 1 injury group received a minor head injury. Over 70% of occupants in the next lowest injury group MAIS 2 received a head injury. More than two thirds of these injuries were moderate (AIS2) and a third were minor (AIS1). Over 80% of occupants in the first middle injury group MAIS 3 injury group received a head injury. These head injuries ranged from minor (AIS1) moderate (AIS2) and serious (AIS3). Slightly more occupants in the second middle injury group MAIS 4 injury group received a head injury. These injuries ranged from minor (AIS1), moderate (AIS 2), serious (AIS 3) and severe (AIS 4). The most common of these injuries for this group were either minor (AIS 1) or severe (M\AIS 4). Most of the occupants in the second highest injury group MAIS 5 received a head injury, The most common injury for this group was critical (AIS 5), with less than half of the injuries were either minor (AIS 1) , serious (AIS 3) or severe (AIS 4). All of the occupants in the highest injury group MAIS 6 received a head injury. Approximately a quarter of these injuries were rated the highest rating maximum (AIS 6). A similar number of occupants in this group received a minor head injury (AIS 1) or a serious head injury (AIS 3). The other occupants in this group received a severe injury (AIS 4).



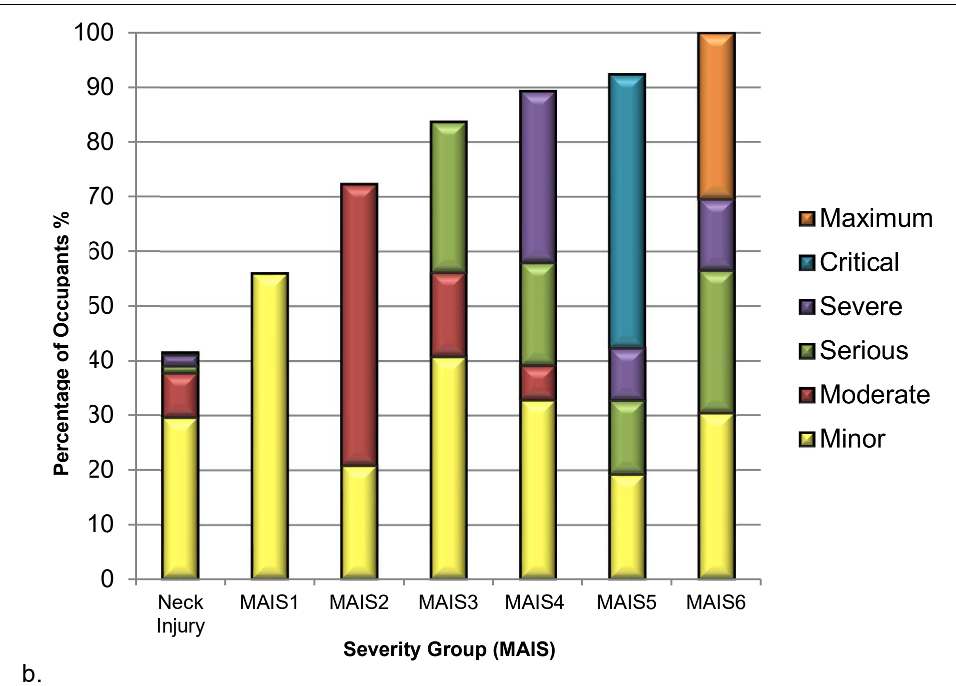


Figure 4.13a: Percentage and severity of head injuries for occupants with low severity neck Injury- Australian Data.
Figure 4.13b: Percentage and severity of head injuries for occupants with low severity neck Injury- UK Data.

The literature supports the findings in the UK database. Jakobsson et al. (2000) reports that occupants involved in a side impact are at more risk of receiving a low severity neck injury when there is a head contact. Literature investigating neck kinematics with head contact shows that only at certain impact directions and impact loads neck injury occurs (Ono et al. 2002). This is illustrated in the UK data as there is a high incidence of head injuries with low severity neck injury of varying severity. The reduced rate of head injuries in the Australian data may be a result of low severity neck injury being under reported with more serious head injuries, as the head injury may mask the neck injury.

4.4 Discussion and Conclusions:

An in-depth investigation into two real-world crash databases were undertaken in this chapter. These were the Monash University Accident Research Centre database (Australia) and the Loughborough University Co-operative Crash Injury Study (UK). Both databases were analysed separately due to there being slight variation in how the data was collected. This investigation worked to identify the common factors associated with low severity neck injury in side impacts and compares these factors to the factors presented in the Chapter 2 Literature Review.

The primary goal of this chapter is to provide a base of understanding into low severity neck injury from side impacts. A secondary goal was to establish typical side impact case scenarios that may lead to a low severity neck injury. These typical case scenarios are modelled in later chapters using a combined human model developed by TNO Netherlands for MADYMO 5.4.1.

The results in this chapter provide a detailed description of the factors associated with low severity neck injury from a side impact. There are differences in the results between the Australian database and the UK database. In spite of these differences typical side impact crash scenarios that lead to low severity neck injury can still able to be determined.

The results for each factor investigated are:

a) **Gender:** the Australian data shows that males and females are equally affected by low severity neck injury. However, the UK data shows that females are more frequently affected by low severity neck injury in side impacts. The variation in the gender results between the two databases could be a result of the smaller sample size of the Australian data. The higher incidence of females receiving more injury than males is consistent with that presented in literature. Research that investigates low severity neck injury and compares the crash incidence of injury between males and female consistently shows that females are more likely to receive an injury compared to males (Nolet et al. 2010, Teamming et al., 1998, Morris et al., 1996, von Koch et al. 1995, Fildes et al. 1995, Gibson et al. 2000 and Bring et al. 1996). More specifically the Volkswagen accident database (Teamming et al., 1998) showed that males experience more vehicle impacts compared to females, yet females have twice the risk of males of receiving a low severity neck injury. Minton et al. (1997) also found that females experienced more disability than males from a low severity neck injury, suggesting that these injuries are more debilitating for females.

b) Age: the results show that likelihood of being injured at a particular age is influenced by gender. The Australian data shows that females age 17-26 and 37-46, males 47-56 are more likely than other `age groups to receive a low severity neck injury. The UK data shows that females aged 17-26 and 27-36 groups are shown to experience more injury. For males no particular age group showed a higher incidence of injury. The Australian database shows the average age of males receiving a low severity neck injury is 39 years (S.D. 16 years) and for females 38 years (S.D. 15 years). The UK data shows that the average age for males is 34 years (S.D.16 years) and for females 37 years (S.D. 16 years).

The literature shows that the age of occupants is within a similar range to the age show in this research. The average age reported in the literature range in the mid-thirties for an occupant receiving a LSNI from a front impact. This was reported by Morris et al., (1996) with the average age of males 33 years and females 35 years. The average age of occupants receiving a LSNI from a rear impact was reported by Teamming et al. 1998 to be older for males at 41.5 years, but similar for females at 36.0. The literature supports that cases of low severity neck injury does occur in younger populations as shown in this research. When the data is sorted in age brackets the 17-26 years age bracket is shown to be one of the common groups in both the Australian and UK data sets. This is supported by Styrke et al. (2012) shows in longitudinal data that the age category 20- 24 years has the highest incidence of low severity neck injury for all impact angles.

c) Occupant Seating Position in Vehicle: when assessing seating position of occupants who receive a low severity neck injury in a side impact, drivers are more likely to be injured. This is supported by the literature that shows that front row occupants (driver and front passenger) are at a greater risk of receiving a low severity neck injury compared to rear seated occupants Lösvund et al. (1988). When comparing occupant position in the vehicle Styrke et al. (2012) found that 71.6% were drivers, 19.9% were front-seat passengers presented to the Emergency Department with LSNI.

d) Side of Impact and Angle of Impact: the Australian data and UK data shows that impacts resulting in a low severity neck injury are more likely to be near-side impacts. The Australian and UK data shows that 2 and 3 o'clock near-side impacts are the most common impacts that result in a low severity neck injury. With these impact angles the Australian data shows that 9 o'clock and 10 o'clock near-side impacts also commonly result in this injury. The UK data shows that 12 o'clock near-side impacts result in more low severity neck injury.

The risk of receiving a low severity neck injury in a side impact is shown in literature to be influenced by the impact angle and the direction of impact (near or side). When looking at near side impacts Jakobsson et al. (2000) found that occupants involved in near side impacts at 7-8 o'clock (drivers and left rear passengers) and 4-5 o'clock (right front and right rear passengers) are shown to have the greatest risk of receiving a low severity neck injury. When the impact angle changes to be a 90° side impact (3 o'clock and 9 o'clock) the literature shows that far side impacts pose a greater risk than near side impacts at the same angle Jakobsson et al. (2000). Far side at an offset angle 1-2 o'clock (drivers and left rear passengers) and 10-11 o'clock (right front and right rear passengers) are shown Jakobsson et al. (2000) to have the lowest risk of injury.

e) Striking Object: both the Australian and UK data shows that low severity neck injury is shown to more likely to occur in car-to-car impacts. The literature shows that there is a relationship the mass of the striking vehicle and the struck vehicle. The results show that when vehicle masses are equal such as in car-to-car impacts, the risk of neck injury is from 1.4 to 2.32 times as high for the occupants in the struck vehicle by von Koch et al. (1995).

f) Delta V: this data is only available for the Australian data. It is shown that side impacts that result in low severity neck injury are most likely to occur at a Delta V range of 16-26 km/h.This value range is less Hell et al. (1998) that conclude that 10-15 km/h is the most critical delta v for low severity neck injury in rear impacts. In addition to this Hell et al. (1998) made the recommendation that all low severity neck injury countermeasures should be tested at a delta v between 12 and 25 km/h. This recommended range includes the range of delta v that was found in this study. Meyer et al. (1998), reported that a delta v above 10 km/h can produce a low severity neck injury. This is also a lower value than the range found in this study.

g) Head Contact: The Australian data shows that 18% of occupants with a low severity neck injury also received a head injury. These head injuries were shown to be below AIS 2. The UK data shows that 42 % of occupants received a head injury with a low severity neck injury. These injuries were shown to be of up to a maximum of AIS 5 severity most injuries an AIS 1. The literature shows that there is a link between head contact and low severity neck injury. Head contact can commonly occur in side impacts due to close proximity of the head to the door. Jakobsson et al. (2000) shows that in side impact cases there is an increased risk of an occupant also receiving a low severity neck injury when an occupant experiences a head contact. The effect that head contact has on LSNI has also been reported to influence there being greater severity of symptoms when the occupant has a head impact (compared to no head contact). Rizzetti et al. (1997) reports that neck injury occurs as a result of the direct loading to the head from the contact.

.

4.5 Practical applications of the results

There are two practical applications of the data analysis undertaken in this chapter. The first was to address the gap in research that investigates LSNI from side impact to begin a base of understanding regarding the injury at this impact angle as outlined in the aims at the introduction to this thesis. The base of understanding that this chapter provides supports a detailed analysis of low severity neck injury from side impact which is also an aim outlined in the introduction of this thesis. The findings in this chapter are the basis for the case selection of the side impact crashes that are investigated in detail through computational reconstructions that have been presented in Chapter 5 and Chapter 6.

Chapter V
A Comparison of the Combined Human Model to Existing Lateral Impact Experimental Data

5.1 Introduction

Due to the complex nature of investigating low severity neck injuries in side impacts, human modelling was the method chosen to reconstruct the real world cases in this investigation. This was determined the best methodology as with the nature of the injury, often when an occupant experiences a low severity neck injury they report severe pain and limited movement (disability). This usually occurs with no visible physical signs of injury to the soft tissues, such as tears visible on scans. The patient's reporting of pain is a major part of the diagnosis. This poses a problem when using cadavers or dummies to reconstruct the motion leading to injury, as there is no scope for pain to be measured.

While it is acknowledged that more advanced dummy necks exist for rear impacts such as BIORID (Davidsson 1998, Welch 2010), there are limited tools available to reconstruct human neck motion effectively in side impact crashes. The simplified anatomy of the neck in the side impact dummies also provides limited information about the mechanism of the cervical spine at all levels. Using volunteers to test neck injury in side impacts was deemed too dangerous due to the risk of participants receiving a long term injury from the acceleration tests. In contrast, computational human modelling, while still limited by the lack of pain as a measure of injury, is an inexpensive way of investigating neck mechanics during side impact.

The MADYMO (Mathematical Dynamic Modelling) human body model combined with detailed neck model (termed in this thesis as combined human model) was chosen for this research as they are validated models that have been used in previous research (de Jager et al. 1996, Happee et al. 2000, van der Horst et al, 2001). MADYMO (MADYMO Manual 6.0, 2001) is a tailored crash analysis software that allows vehicle safety designs to be evaluated and optimised. This software is available through TASS, Netherlands. TNO, Nederlandse Organisatie voor Toegepast Natuurwetenschappelijk Onderzoek, (*Netherlands Organization for Applied Scientific Research*). The human model and the detailed neck model have been extensively validated as separate models (as outlined by Happee et al. 2000). The two models combined have been used in rear impact research (van der Horst et al. 2001) and the adapted neck model has been used to research into injury causation and injury management (Sim 2004, Storvik et al. 2011). TNO reports the benefits of the human model to be:

- Improved bio-fidelity over dummy models
- Multi-directional modelling capability in frontal, lateral and rear impacts as well as oblique impact directions. (MADYMO Manual 6.0, 2001)
- Capability to incorporate additional biomechanical data into the model (such as sitting posture or a varied head position).
- Option of programming muscle activity (MADYMO Manual 6.0, 2001)

It is important to have a model that closely replicates human anatomy including muscle activity. It is also important for the work to be undertaken in Chapter 5 that the model is multi-directional and functions in oblique directions. These capabilities allow for the simulation of both near and far side impacts, of varying angles. In addition to this, the detailed neck model as an established model required minimal adjustment as it already contains represented anatomic structures of the neck. The detailed neck model is of interest to this study because it models many of the anatomical structures in the human neck, such as bone, disc, ligaments and muscles. These structures have been described in detailed for comparison against the anatomy of the cervical spine in Appendix 1.

In this chapter, additional validations of the combined human model with detailed neck model in lateral and oblique impacts, are presented. The purpose of this chapter is to compare the combined model's capabilities with the existing experimental data. Any identified limitations of the combined model will be considered when using the model to reconstruct real world crashes in Chapter 5.

Two different data sets have been used in this chapter for comparisons. One dataset contains cadaver data for a lateral impact provided by Wayne State University (Cavanaugh et al. 1990 1993, Irwin et al. 1993) the other dataset has volunteer data for lateral and oblique impact from the Naval Biodynamics Laboratory (Wismans et al. 1986).

5.2 Computational Simulation Software Package- MADYMO (MAthematical DYnamic MOdel)

MADYMO (MAthematical DYnamic MOdel) is a computational software package which can be used to model crash situations occupant behaviour during car crashes. The level of detail that can be included in the computational models in MADYMO contribute to a high degree of accuracy. The output of the models allow for occupant behaviour during car crashes to be analyses to assess and predict the injuries that would be sustained victims if the same crash was to occur in the real world.

MADYMO allows for three-dimensional simulations and the program allows for one simulation to combines both multibody and finite element components. Examples of how a multibody system could be used within in a model is a crash test dummy with its complex joint structures. An example of a finite element structure that could be incorporated within a model is with an airbag deploying. MADYMO output allows for a broad range of standard output parameters, including accelerations, forces, torques and visual kinematic data. Predicted injury parameters can be derived from calculations using these variables. The specifications of the simulations undertaken in this chapter are specific to each study undertaken, these are presented in sections .Due to the diversity in modelling that MADYMO allow for accuracy of MADYMO is determined through the validations of each specific system modelled. The prior validations of the MADYMO models relied on in this research and new validations undertaken to support this research are discussed further in this chapter.

5.2.1 Combined Human Model- Anthropometry of the Human Body Model

This section provides an overview of the structure of the combined human model. This section has been provided for comparison to the human anatomy in the preceding Appendix 1. It has also been presented to provide a base of understanding about the model for the work undertaken in the following chapters.

The anthropometry of the body of the human model has been described in detail in the MADYMO Manual 6.0 and by Happee et al. (1998, 2000). The human body model has been developed to anatomically represent the human body, and therefore, has a head, spine, torso, upper and lower extremities. The human model's geometry comes from the RAMSIS anthropometric model (Happee et al. 1998), including joint locations, joint range of motions, segmental masses, centre of mass of body segments and dimensions for triangular skin connected to various body segments. Additional variables required to complete the human model were rotational inertia, joint resistance models and the joints in the spine which were derived from literature (Happee et al. 1998, 2000).

The spine of the human body model has been reported in detail by Van den Kronnenberg et al. (1997). The RAMSIS model was expanded from the 7 spine joints to 25 in the MADYMO human model (illustrated in Figure 5.1). The vertebrae are modelled as rigid bodies (Happee et al. 1998, 2000). The human body model compares to a human volunteer in that the thorax of the human model deforms in three dimensions from external impacts. The thorax is also responsive to spinal deformations. The shoulder in the human model consists of a scapula, clavicle, sternum and humerus attached by joints. A passive force model allows the scapula to glide over the thorax (Happee et al. 1998, 2000).

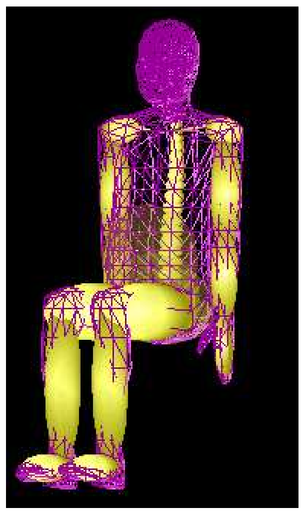


Figure 5.1: Human Body Model (showing detail of spine)

5.2.2 Bone

There are 9 parts of the detailed neck model that replicate bone. The model consists of a skull with 7 cervical vertebrae and the first thoracic vertebra (T1) visible in Figure 4.2. The first thoracic vertebra (T1) is the base of the neck model. All of the bones have been modelled as rigid bodies.

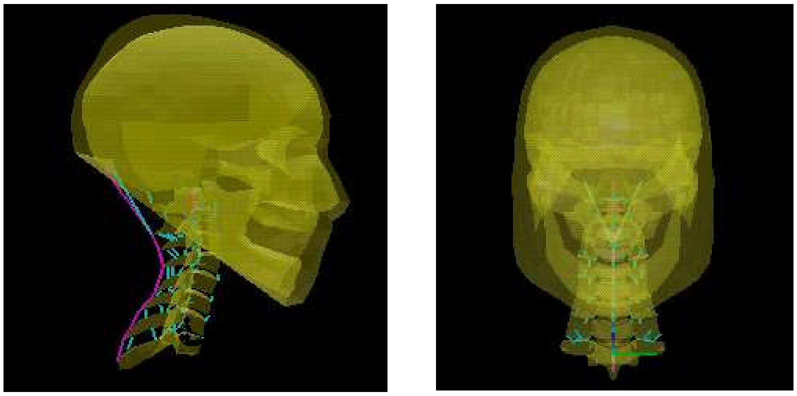


Figure 5.2: Detail neck model as modified by van der Horst et al. (1997)

The outer surfaces of the rigid bodies have been covered by a facet surface consisting of triangular and quadrangular facets with nodes. The geometry of the bones in this model come from a post mortem human subject and has been described in detailed by van der Horst et al. (2002). Each of the cervical vertebrae has a vertebral body, transverse process, articular facets and spinous process to compare to the human vertebrae (Figure 5.3). As with the human cervical spine, these anatomical structures are the attachment points for the soft tissues.

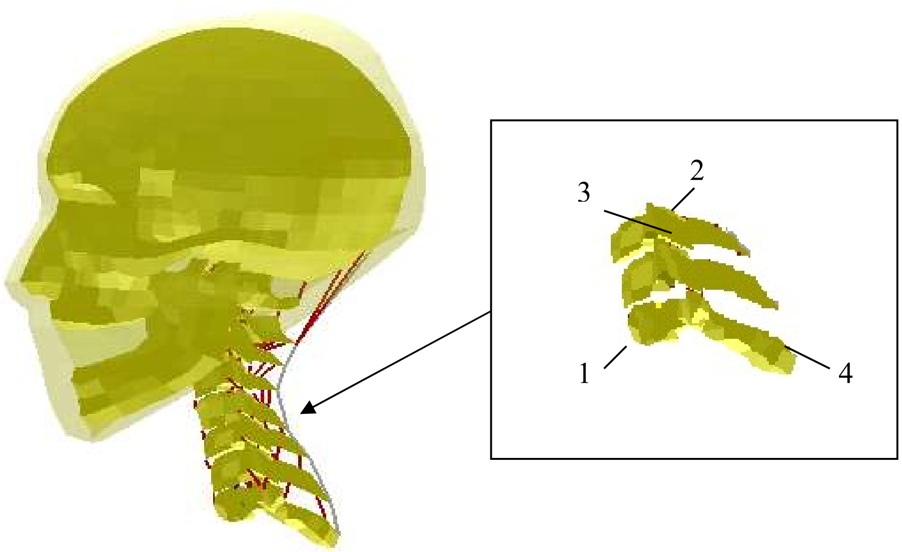


Figure 5.3: Detailed Neck Model Segment C5-C6 (van der Horst, 1997)
1.Vertebral Body, 2.Transverse Process, 3.Articular Facets, 3.Spinous Process

5.2.3 Facet Joints

The facet joints in the cervical spine are not modelled as surface to surface contacts. Instead facet joints are modelled with non-linear translational springs, this is due to there being limited literature on the geometry of facet joints. There is also limited available literature regarding the contact stiffness of the facet joints. Due to this, a modified value of the nonlinear static compression of the discs is used in the model instead. The joint capsules surrounding the facet joints are modelled as 4 single line elements van der Horst et al. (2002).

5.2.4 Occipital Condyles and Dens

The occipital condyle on the skull and the dens on the first cervical vertebrae have been meshed in detail. This creates a smooth convex-concave contact area. The contact between the occipital condyle and the dens is modelled as a stiff (rigid body) contact with the ligaments and the joint capsule creating resistance in tension and shear van der Horst et al. (2002).

5.2.5 Spinous Processes

The spinous process of each of the vertebrae is the attachment point for the ligaments. Each spinous process has been modelled similar to the facet joints with a non-linear spring model. This method was chosen to replicate contact between two neighbouring spinous processes that can occur in extreme extension. The same stiffness values are used for facet surface contact as no literature exists giving this value for spinous process contact van der Horst et al. (2002).

5.2.6 Intervertebral Discs

There are no discs between the occiput and C1 and between C1 and C2 and are therefore not included in the model. Discs are modelled between the vertebrae C2-C7. The discs are modelled with 6 degrees of freedom (translation and rotation) as parallel springs and dampers. Loads are placed on the discs by the adjacent vertebrae and are calculated at the mid-point of the disc. It is important to note that the discs do not allow for coupling between the different degrees of freedom. The values used for stiffness and damping are detailed by van der Horst et al. (2002).

5.2.7 Ligaments

The attachments points of the ligaments are based on the anatomical sites of the origins and insertions in the cervical spine. The ligaments are modeled as single line elements. Joint capsules between facet joints, are modeled as four single line elements. The ligaments are modeled to produce force only in tension. The ligaments included in the detailed neck model are Ligament nuchae, Transverse, Tectorial membrane, Cruciform , Alar, Anterior longitudinal, Posterior longitudinal, Interspinous, Flaval, Capisilar (van der Horst et al. 2002).

5.2.8 Muscles

The muscles are modelled using Hill type muscle elements. This method of modelling the muscles is described in the MADYMO 5.4.1. Theory Manual (1999). The Hill type model is illustrated in Figure 5.4. This model consists of a contractile element (CE) to replicate the active force generated by the muscle. Parallel to this is an elastic element (PE) to replicate the elastic properties of the muscle. Either side of these are masses to replicate the muscle mass (M1and M2) and either side of these at either end are elastic elements that replicate the muscle tendons (SE1 and SE2) .

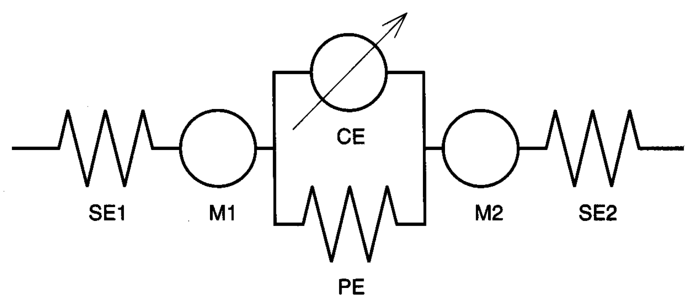


Figure 5.4: Hill type muscle model MADYMO 5.4.1. Theory Manual (1999)

This model has been developed to address some of the limitations of the model presented by de Jager et al. (1996). The neck muscles were originally modelled as straight lines, van der Horst et al. (2002) has developed the neck muscles to curve with the neck. Additional muscles have also been included in the model by van der Horst et al. (2002). A list of the muscles included in the detailed neck model are presented in Table 5.1.

Table 5.1: List of muscles represented in detailed neck model (Van der Horst 2002 & de Jager 1996)

Flexor muscles	Extensor muscles
Longus colli	Trapezius
Longus capitis	Sternocleidomastoid
Scaleneus anterior *	Splenius capitis
Scaleneus medius *	Splenius cervicis
Scaleneus posterior *	Semispinalis capitis
Lumped hyoids **	Semispinalis cervicis
	Longissimus capitis
	Longissimus cervicis
	Levator scapulae
	Multifidis cervicis
All muscles are modified by Van der Horst (1997) compared to the De Jager model unless specified. * Not changed by Van der Horst (1997) ** New muscles added by Van der Horst (1997)	

The muscles of the detailed neck can be modelled in two states, active or passive. Passive muscle functions are best used when simulating cadaver tests where there is no muscle activation. Active muscle functions are used when simulating volunteer tests when there is muscle activation. Muscle activation level can be set between 0-100%. A delay period can also be set to allow for reaction time. Two activations states have been used in literature, these have been outlined in Table 5.2.
de Jager et al. (1996) used an activation level of 95% and a delay of 50ms and van der Horst et al. (2002) used an activation level of 100% and a delay of 25ms.

Table 5.2: Activation level of muscles used in detailed neck model (van der Horst et al. 2002, de Jager et al. 1996)

	Activation Level	Delay Time
De Jager (1996)	95	50ms
Van Der Horst (2001)	100	25ms

5.3 Validation of the Combined Human Model

The combined human model is a validated model that has used in previous research studies (van der Horst et al. (2001). The co-ordinate system of the model has been presented in Appendix 3, as it is important for the interpretation of results.

Figure 5.5 shows the sequence of validations for the combined human model. The human body model has been validated as single model by (Happee et al. 2000) this work has been discussed in detail in the following section. The detailed neck model has also been developed and validated as a single model by De Jager (1996 & 1998). This work is also discussed in the following section. Additional new validations of this combined model against volunteer data for lateral and oblique angles have been presented in the following section, undertaken as a part of this investigation.

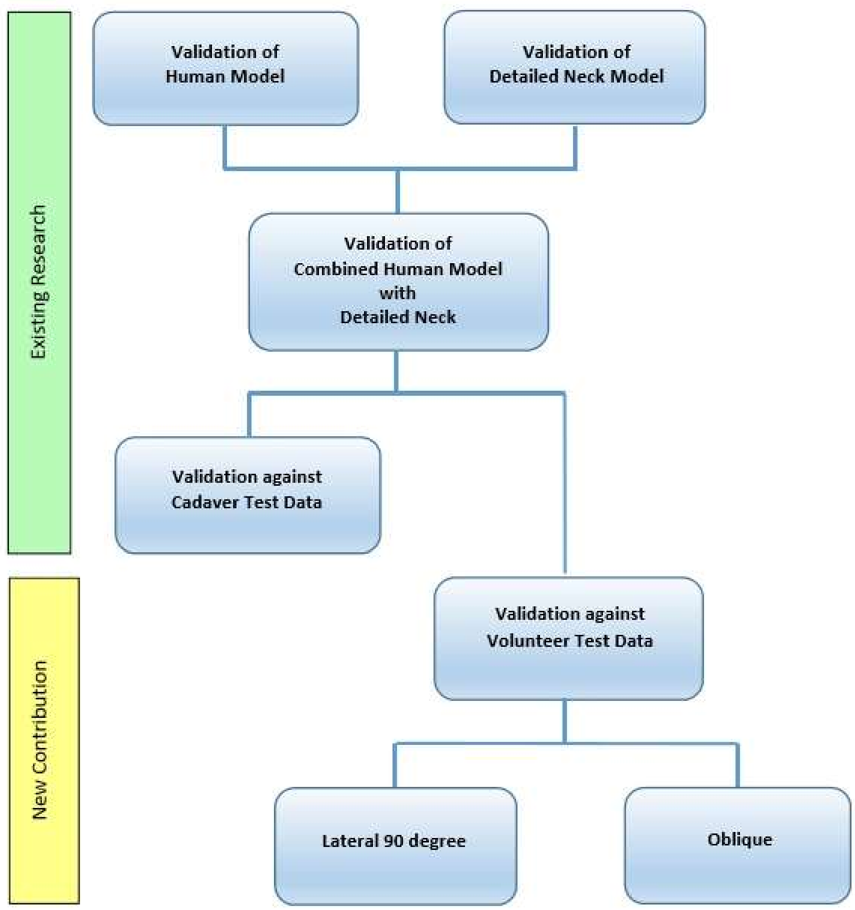


Figure 5.5: Outline of Validation of Human Body Model with Detailed Neck Model

5.3.1 Human Model

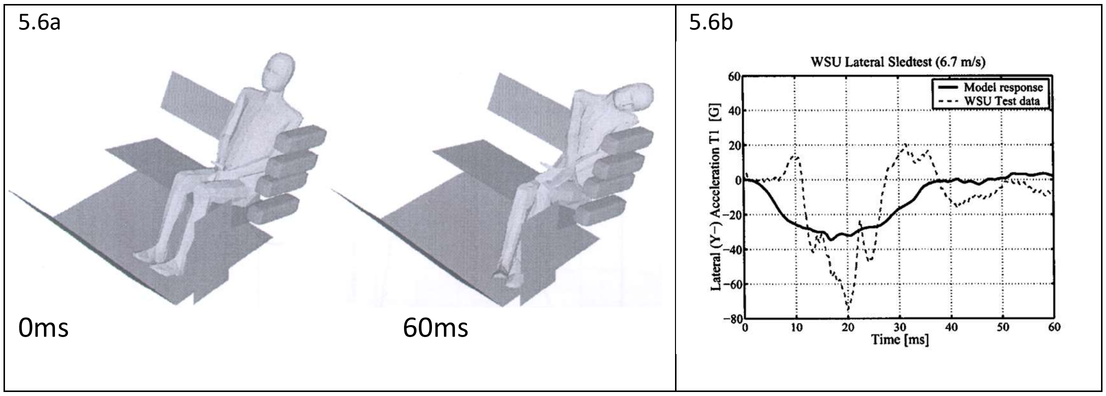
The human model has been validated against both cadaver and volunteer data (Happee et al. 2000) for multiple directions (rear, front and lateral impacts). This is important for this investigation as the analysis of real world crashes undertaken in Chapter 3 showing that low severity neck injury typically occurs during oblique side impacts as well as those of 90°. The human model has been validated both a whole model and as body segments. A summary of these validations are presented in Table 5.3 (Happee et al. 2000). The table shows that the model has only being tested against post mortem human (cadaver) subjects in lateral impacts. This supports the need for testing the combined human model against human volunteer subjects that has been presented in following sections of this thesis.

Table 5.3: Past Validation of Human Model (Happee 2000)
[PMHS- Post Mortem Human Subject]

Test Set-up	Loading Direction					
	Frontal		Rear		Lateral	
	Type	Severity	Type	Severity	Type	Severity
Whole Body Sled	Volunteer	3-15g	Volunteer	4-5g	PMHS	21g, 37g
Spine Quasi-static	Volunteer		PMHS	9-12g		
Thorax Impactor	PMHS	3-10m/s	Volunteer	9-12g	PMHS	3-6m/s
Abdomen Impactor	PMHS	6-9m/s				
Shoulder Impactor					PMHS	4-7m/s
Pelvis Impactor					PMHS	3-10m/s

The most detailed validations of the human model in the lateral direction were undertaken by Happee et al. (2000). The experimental data used was cadaver tests undertaken at Wayne State University presented by Irwin et al. (1993). The experimental setup undertaken by Irwin et al. (1993) has been presented in detail in section 5.4 as it was the same experiment that was used to validate the human model combined with the detail neck model used in this investigation. In the experiment undertaken by Irwin et al. (1993), post mortem human subjects were laterally impacted into a side wall at velocities of either 6.7m/s or 9.1 m/s. Happee et al. (2000) simulated this cadaver experiment using the human model. A sample of the model's kinematics at a velocity of 6.7 m/s has been presented in Figure 5.6a. The subject's kinematics are displayed at a resting position (0ms) and with the body in contact with the wall and the neck laterally bending (60ms). The lateral acceleration for the first thoracic vertebra from the model and experiment has also been presented in Figure 5.6b (Happee et al. 2000). With respect to the human models validation compared to frontal and rearward volunteer sled tests, lateral and rear PMHS sled tests, frontal and lateral PMHS impactor tests, (Happee et al. 2000) describes the human model as satisfying "the available biofidelity requirements in terms of kinetics, chest deflections and accelerations".

Figure 5.6: Lateral Validation of Human Model by Happee et al. (2000).



5.3.2 Neck Model

The development of the detailed neck model is discussed in detail, in the research thesis "Mathematical Head-Neck Models for Acceleration Impacts" by de Jager (1996). The model was developed form a three-dimensional model presented by Deng and Goldsmith (1987) to include simulated muscles to replicate muscle response and resistance during an impact. The detailed neck model been validated for frontal and lateral impacts against the responses of human volunteers (de Jager 1996 & 1998 and van der Horst ,1997).

De Jager (1996 & 1998) tested the models response against volunteer front and lateral impact data presented by Ewing and Thomas (1972). De Jager (1996) describes the model validations for linear and angular accelerations of the head and trajectories of the occipital condyles and the centre of gravity of the head and lateral head rotation as agreeing "excellently". De Jager (1996) identified a limitation in the model that axial rotation of the head was larger than that recorded in the volunteers. This was also found with the lateral rotations of the lowest intervertebral joints. This limitation will be noted and considered in the findings of this investigation as axial rotation of the head and lateral rotation of the lower vertebrae may be observed in the kinematics of occupants in real world side impacts simulations presented in Chapter 6. When comparing muscle activity in the detailed neck model it was found that to accurately replicate the volunteer data the muscle functions needed to be active and not passive. A comparison of the models behaviour with both passive and active muscle functions is presented in Figure 5.7.

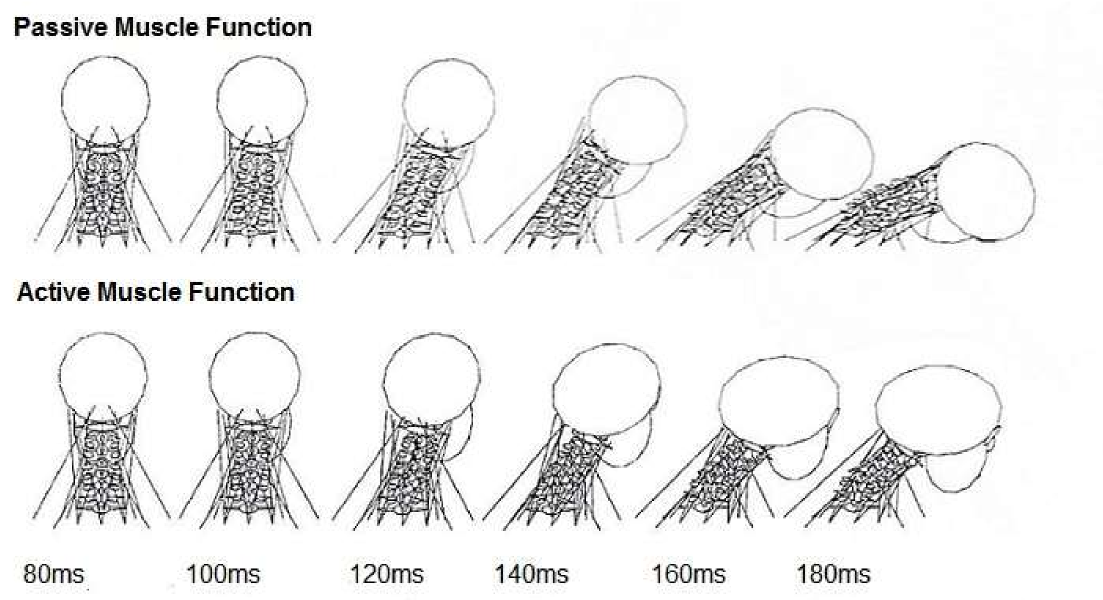


Figure 5.7 Muscle behaviour for passive trial (top) and active trial (bottom) during lateral impact (De Jager 1996)

To address the problems with the unrealistic head rotation findings in the model reported by De Jager (1996 & 1998), the model was further developed van der Horst (1997) so that the muscles curve around the neck during bending. To assess this modification of the model, it was evaluated again with the application of neck muscle activation. The head-neck responses of volunteers exposed to 15g frontal and 7g lateral sled tests were compared to the models output. The accelerations were chosen to correspond to the magnitudes tested of the frontal and lateral HYGE sled experiments undertaken by the Naval Biodynamics Laboratory and presented by Ewing et al. (1977).

In the 15g frontal simulation, the extensor muscles of the neck model were activated at 100% to oppose head and neck flexion. The model's response with this muscle activation was reported to correspond to the volunteer's response (van der Horst et al. 1997), although some differences were observed. The head and neck rotation, in the model were less than the volunteers, as were the displacement of the occypital condyle and the center of gravity of the head. van der Horst et al. (1997) compared a model simulation with the condition of passive muscles to the same simulation with the condition of active muscles. The active condition provided a more human like response. This finding is important, as the crash simulations presented in Chapter 5 are reconstructions of human occupants, and the muscle function of the neck is an important variable to consider in these simulations.

For the 7g lateral simulation, the left side flexor and extensor muscles were activated at 100% to oppose the right sided impact. The model's active muscle response and that of the volunteer's compared well. The trajectories, lateral and axial rotations were all reported to be within the desired range of the volunteers (van der Horst et al., 1997). The research undertaken by van der Horst et al. (1997) supports the choice of using the detailed neck model in the lateral impact research in this study. Again, the need to ensure that the model has active neck muscle functions have been highlighted when the model is used to simulate the motion of live human occupants.

5.3.3 Combined Human Model

The combined human model (human body model combined with the detailed neck model) has previously been used to investigate rear impacts by van der Horst et al. (2001). This investigation compared the response of the model to data from both post mortem and human volunteer subject tests. Simulations of post mortem human tests were undertaken to compare two different stiffness conditions for passive muscles. The response of the model was compared to rear impact experimental data tested at a velocity of 10 km/h. It was found that the models response was more comparable to the experimental data when the model had increased stiffness in the passive muscles.

The response of the model was compared to data from volunteers seated in a standard car seat with a head restraint attached to a sled, tested at a velocity of 9.5km/h (simulated rear impact). The kinematic response of the model was shown to be a good match to that of the volunteers. An interesting finding reported by van der Horst et al. (2001) was that the seating posture of the model has an influence on the response of T1 and the head motion. This finding has relevance to the work undertaken in Chapter 5 that reconstructs occupants in real world side impacts. The level of muscle activation in this simulation was not reported.

5.4 Configuration of Post Mortem Human Subject (Wayne State University) and Computer Simulation

The combined human model, as used by van der Horst et al. (2001) to investigate rear impact has been used in this chapter for comparison against experimental lateral impacts. This was undertaken to determine the suitability of the model to predict the response of the human neck in side impact crashes. In this section, the response of the 50th percentile human male model with detailed neck was compared to a single post mortem human subject (male) of similar stature. The post mortem human subject was previously a part of an investigation into lateral injuries undertaken by Wayne State University, and has been reported by Cavanaugh et al. (1990 & 1993) and Irwin et al. (1993). The human model alone has been previously validated by Happee et al. (2000) against the same data set, although this model did not include the detailed neck.

The post mortem subject's (SIC-07) physical details against the combined human model used in this investigation are as listed in Table 5.4. The subject was chosen as the gender, height and weight was the closest match to the 50th percentile male human model.

Table 5.4: Physical details of the post mortem subject and human model. (Cavanaugh et al., 1990 & 1993), Irwin et al. (1993), MADYMO Human Model Manual 6.0 (2001).

	Post Mortem Human *Subject SIC-07*	Human Model
Gender	Male	Male
Age	67	N/A
Height	1.70	1.74
Weight	74.8	75.7
Cause of Death	Heart Disease	N/A

The human model is 40 mm taller and weighs 0.9 kg more that the post mortem human subject (Table 5.4). The post mortem human subjects were cleared of any anatomical abnormalities prior to testing. In preparation for the experiment, cadavers were un-embalmed and were not tested on until rigor mortis had passed. The subjects were instrumented with accelerometers, pressure transducers and load cells (Cavanaugh et al. 1990 & 1993, Irwin et al. 1993). The subject was placed on a Heidelberg- type seat fixture without any restraint as illustrated in Figure 4.8. This fixture was mounted to a horizontally accelerated sled. There were three variations of impacting walls used by Wayne State University: flat rigid side wall, flat padded side wall and side wall with 6 inch offset. The subject used in this investigation was investigated on a flat-rigid side wall. The apparatus with flat padded side wall is illustrated in Figure 5.9. For the computer simulation the padding was not taken into account as it was removed when subject SIC-07 was tested.

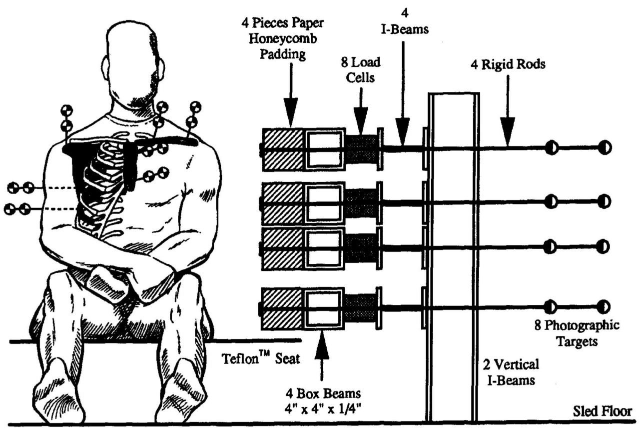
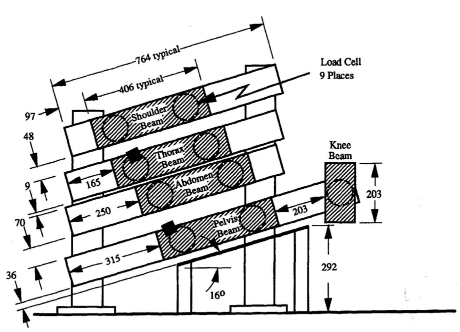
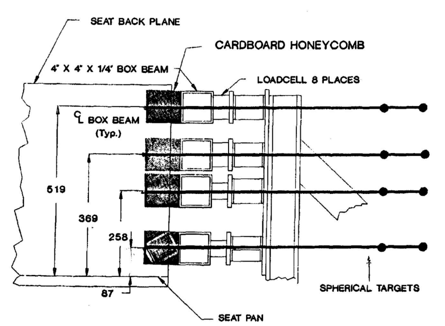


Figure 5.8: Experimental setup of Wayne State University lateral cadaver tests with flat padded side wall (Irwin et al. 1993 adapted from Cavanagh et al.1990).

The Wayne State University seat fixture had a Teflon coated seat with two thin rear beams supporting the posture of the cadaver (not visible in Figure 5.8). These rear beams were intended to be rigid beams, although upon observation of video of the experiment the beams appear to be flexible, this flexibility may be significant for modelling purposes. Five steel beams make up the sidewall. They are positioned at the knee, pelvis, abdomen, thorax and the shoulder. The dimensions of the wall are illustrated in Figures 5.9a and 5.9b.

 a	 b
Figure 5.9a: Lateral view (-Y) of Wayne State University Heidelberg-type sled, Irwin et al. 1993 adapted from Cavanagh et al.1990)	**Figure 5.9b : Frontal view (X) of Heidelberg-type sled, Irwin et al. 1993 adapted from Cavanagh et al.1990)**

The lateral view of the seat fixture, highlighting the incline of the seat and the sidewall is presented Figure 5.9a. The angle of this incline is 16°. The spacing between the beams in the sidewall is presented in Figure 5.9b. These beams are not spaced evenly as they are designed to correspond with key areas of the body. A summary of the measurements of the Heidelberg type seat fixture used by Wayne State University is presented in Table 5.5. These dimensions were used to scale the sled model used in the computer simulations.

Table 5.5: Summary of sled dimensions

Seat Angle	16°
Height of side beam (shoulder, thorax, abdomen ,pelvis)	101.6 mm (4 inches)
Width of side beam (shoulder, thorax, abdomen ,pelvis)	101.6 mm (4 inches)
Length of side beam (shoulder, thorax, abdomen ,pelvis)	406 mm
Distance from seat to center point of pelvis beam	87 mm
Distance from seat to center point of abdomen beam	258 mm
Distance from seat to center point of thorax beam	369 mm
Distance from seat to center point of shoulder beam	519 mm
Distance between seat and pelvis beam	36 mm
Distance between pelvis beam and abdomen beam	70 mm
Distance between abdomen beam and thorax beam	9 mm
Distance between thorax beam and shoulder beam	48 mm

The subject was positioned on the seat fixture approximately 50 centimeters from the sidewall. The sled was accelerated to the desired peak velocity (between 6.7m/s and 10.5m/s) and then rapidly decelerated. This rapid deceleration caused the subjects to continue to translate along the Teflon seat and into the sidewall. The peak sled velocity for SIC-07 was 6.7m/s (Cavanaugh et al. 1990 & 1993, Irwin et al. 1993).

The results for trial SIC-7 have been presented by Cavanaugh et al. 1990 & 1993, Irwin et al. 1993. The acceleration data for the head and first thoracic vertebra (T1) have been used in this study for comparison with the combined human model in the following section. It has been reported that subject SIC-07 received injuries to the shoulder as a left acromion separation and left acromion fracture. Injures to the thorax were 13 left rib fractures and 3 right rib fractures Irwin et al. (1993).

The acceleration data for the head and T1 have been presented against the predicted data for the model in Figures 5.11 and 5.12. The head data has been filtered at 1000 Hz the T1 data and been filtered at 180 Hz.

5.5 Computer Simulation of Combined Human Model - 90° Lateral Simulation- Passive Muscles

The human model was provided by TNO Netherlands with the detailed neck, was already positioned in a seated position. No structural changes were made to this model. Modifications made were to the angle of the trunk, knees and ankles, to fit with the inclined seat of 16°, and an inclined seat back of 100° to fit on the sled. The lower arm angle was modified to give an elbow angle of 90°. This angle differs to the hand position of the cadaver experiment where the wrists were taped together in the lap of the subject. A 90° elbow angle was implemented to simplify the human model to increase the numerical stability of the model (which was a serious concern during simulations). The recommendation for positioning the human model in the Human Models Manual for MADYMO 6.0 (TNO, 2001), is to include a point restraint force of 100N in the z-direction (vertical) above the center of gravity of the head. This is to keep the head upright, and to prevent the neck from bending under the head's weight.

The Heidelberg type sled was modeled using planes for the seat, seatback, floor and knee beam. The side wall was modeled using ellipsoids. The ellipsoids used to build the sled wall were programmed to have a high degree. This resulted in the ellipsoids having four distinctive sides. The model of the sled followed dimensions of the sled reported in Table 5.5. Contacts between the human model and planes and ellipses were defined to replicate contact between the human body and steel sled.

One acceleration and one velocity were applied to the complete model (human model and sled seat). Gravity (9.81 m/s^2) was applied as an acceleration field for the whole model. A velocity of 6.7m/s was applied to the pelvis of the human model. This velocity matches the one used in the Wayne State University subject SIC-07 experiment. The model simulation time was 0.2s calculated at a time step of 1.0 E-5. It is important to note that the start of the computer simulation was the point where the sled motion had stopped and the subject had begun translation.

Upon viewing the video footage of the subject from the Wayne State University study, it became apparent that the subjects head was contacting of the top beam of the side wall. With the initial geometry, the model head did not contact the wall. Hence, the sled model was modified by raising the top ellipsoid of the wall by 50mm to encourage contact between the head and the side wall. The need for this modification suggests that the neck of the model is possibly longer than that of the cadaver.

5.6 Comparison of Results for the 90° Lateral Impact

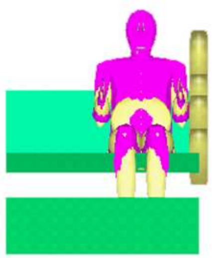
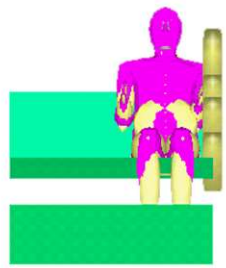
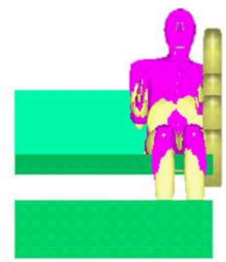
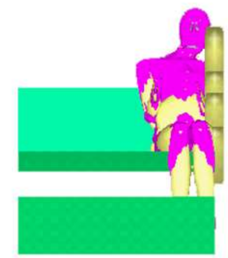
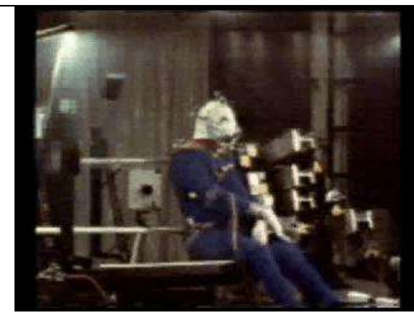
Comparisons of the predictions from the combined human model to those of the Wayne State University cadaver tests are presented in this section. These comparisons are for the kinematic/video data, for the lateral acceleration of the head and for the first thoracic vertebra (T1).

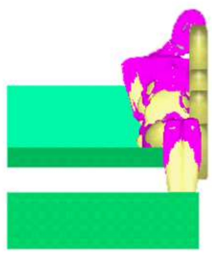
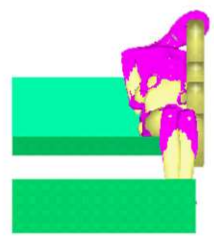
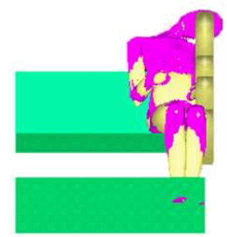
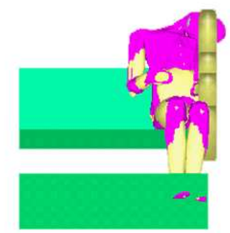
The model's kinematic (video) images compared to the video of the lateral experiment with the human cadaver undertaken by Wayne State University are presented in Table 5.6. This table has 4 rows, with the time presented in the first row at increments of 10 milliseconds. The images from the human model simulation are shown in the second row, and the video images from the experiment are presented in row three. Row 4 has comments related to the key events, when relevant.

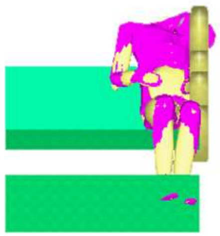
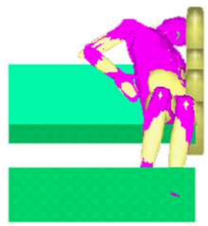
The kinematic images of the human model show that the simulation has a good likeness to the real world experiment, as the same key events occur in both cases. The body translates laterally until shoulder contact occurs with the rigid wall. In both the experiment and the simulation, the translation of the head continues followed by neck bending until the head contacts the wall. The time of the key events highlighted in the video data do show some minor variations between the model and the cadaver test.

From 0 to 20 ms the model and cadaver behave the same, with the arm and body contacting the wall. At 30 ms, the cadaver's neck begins to bend resulting in the head contacting the top of the wall. The model's head continues to laterally translate from 30ms with neck bending beginning at 40ms. This neck bending continues until 70ms when the models head finally contacts the wall. Following head contact, both the cadaver and the model begin to rotate around the head and pelvis which are the contact points with the wall. This rotation occurs at 50ms for the cadaver, and at 90ms for the model. Despite the timing differences, the good likeness between the kinematic results of the combined human model and the human cadaver suggests that the model may be a useful tool to investigate the kinematics of occupants in lateral impacts.

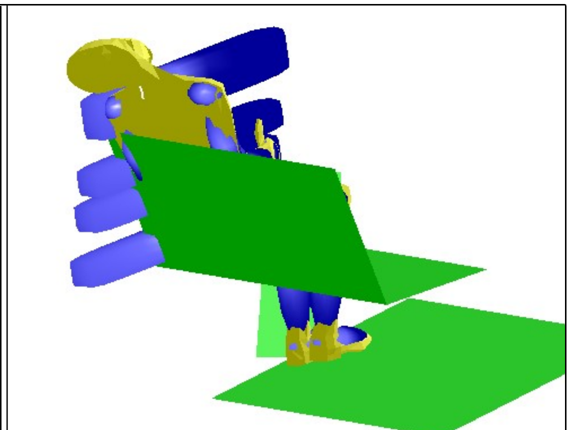
Table 5.6: Kinematic results from 90° lateral impact- cadaver compared to combined human model

Time: 00ms	Time: 10ms	Time: 20ms	Time: 30ms
			
			
	Model/ Cadaver: Arm contacts wall, body contacts wall	Model/Cadaver: Head lateral translation begins	Model: Head translation continues (beam is tilted so head is yet to contact) Cadaver: Neck bending begins

Time:40ms	Time:50ms	Time:60ms	Time:70ms
			
			
Model: Neck bending begins (no evidence of head rotation) Cadaver: Head contact with wall	Cadaver: Body begins to rotate around head and pelvis in contact with the wall		Model: Head contact with wall

Time:80ms	Time:90ms	Time:100ms
		
		
	Model: Body begins to rotate around head and pelvis in contact with the wall	

The timing difference between the model simulation and the cadaver experiment, may be attributed to the difference in the initial seating and body position prior to impact. Previous research by Van der Horst et al. (2001) showed that response of T1 and the head motion was influenced by the seating posture of the model. The pre-impact posture of the cadaver shows the slightly rounder shoulders (kyphosis) with slight neck flexion resulting in a slightly protruding chin, assisted by the hands being bound in the lap of the cadaver. This posture has an influence on the point of impact on the beam, with the cadaver head contacting the front half of the beam (Figure 5.10a). On the other hand, the human model has been pre-programmed to have the head held upright. The model arms are bent at 90° at the elbow, to ensure the numerical stability in the model. This initial position influences the impact point of the head in that the model's head contacts the rear half of the beam (Figure 5.10b). The delay in head contact occurs as the pelvis and shoulder rotates about the wall immediately following body contact and just prior to head contact. In contrast the cadaver trial the body rotation occurs following the head contact.

	
Figure 5.10a Rear view- human cadaver, head contact. Neck is flexed at contact.	**Figure 5.10b Rear view- Human model, head contact.**

Neck length may also be an attributing factor to the timing differences between the model and cadaver's head, and the need to raise the wall to ensure head contact in the model trial. While anatomical measurements are not available, the cadaver's neck appears to be longer than that of the neck pre-programmed in the model.

The lateral (Y- direction) acceleration history of the first thoracic vertebra for the human model (red) and the cadaver (blue) is presented in Figure 5.11. The two traces follow very similar trends. Both traces show an initial positive peak occurring just before 5ms with the cadaver peak at a greater magnitude.

Following this first initial peak, the acceleration for both the human model and cadaver decrease rapidly. The peak deceleration for the human model is 809 m/s^2 at 19 ms compared with slightly smaller peak for the cadaver of 714 m/s^2 occurring slightly earlier around 15 ms. Before peak deceleration occurs there is a slight hesitation seen in both traces. For the human model, it occurs at 10ms, and for the cadaver, it occurs at approximately 8 ms. This hesitation occurs when the body contacts the arm that is already in contact with the wall. This change in deceleration is more pronounced in the cadaver.

Finally, the deceleration crosses through zero. This occurs at 22 ms for the cadaver and at 27ms for the cadaver. This is followed by a peak acceleration of 174 m/s^2 for the cadaver and a rapid deceleration crossing through zero to 37 m/s^2. This deceleration coincides with the head contact with the wall. The acceleration for the model, peaks at a 68m/s^2, followed by minor deceleration peaking at 25m/s^2.

The likeness between the two T1 acceleration traces suggests that the delay in the head contact by the model, observed in the kinematic data is a result of the structure of the model above T1. This also supports the suggestion that the initial head and neck position may have an influence on the predicted acceleration results.

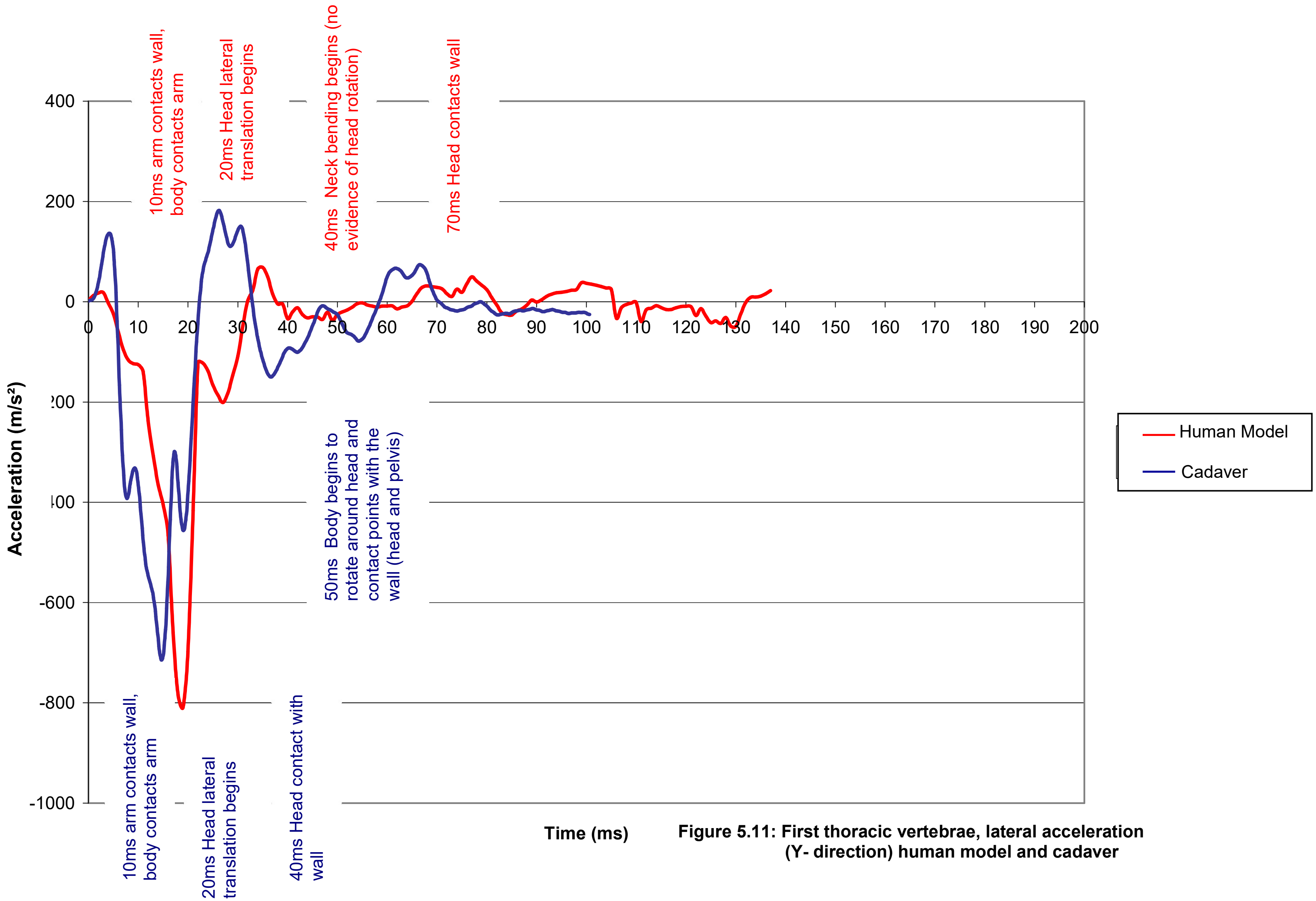


Figure 5.11: First thoracic vertebrae, lateral acceleration (Y- direction) human model and cadaver

The head lateral (Y direction) acceleration data for the cadaver and model is presented in Figure 4.12. The combined human model is presented in blue, with the Wayne State University human cadaver data presented in red. The cadaver shows initial acceleration with three peaks of 282 m/s^2 at 13 ms, 165 m/s^2 at 23ms and 182 m/s^2 at 28 ms. Following the third acceleration peak there is deceleration to a peak of -451 m/s^2 at 56 ms coinciding with head contact. Then the head acceleration returns to 0 around 80ms.

The human model's head acceleration begins with a peak of -102 at m/s^2 at 25 ms. This peak coincides with lateral translation of the head. Following this peak, the trace passes through zero, reaching a peak of 52 m/s^2 at 5 6ms, coinciding with lateral bending of the head. This peak is followed by rapid deceleration peak at -380 m/s^2 at 75 ms, as a result of head contact with the wall. Acceleration then returns to zero at around 80 ms.

The difference in acceleration between the model and the experimental data is a result of the head lateral translation and lateral bending being two separate events. During lateral translation the head decelerates, and during lateral bending, the head accelerates. The total time of these two events takes 70ms. During the cadaver trial, the head's lateral translation and lateral bending occur with head acceleration. These key events take a shorter time of 37 seconds.

The difference in head acceleration suggests a difference in the flexibility in the neck between the model and the cadaver. The passive muscle functions in the model appear to be less flexible than that of the cadaver neck. Using the active muscle functions available in the human model may address the issues of neck flexibility. In the following section the model with active neck muscles is compared to human volunteer subjects.

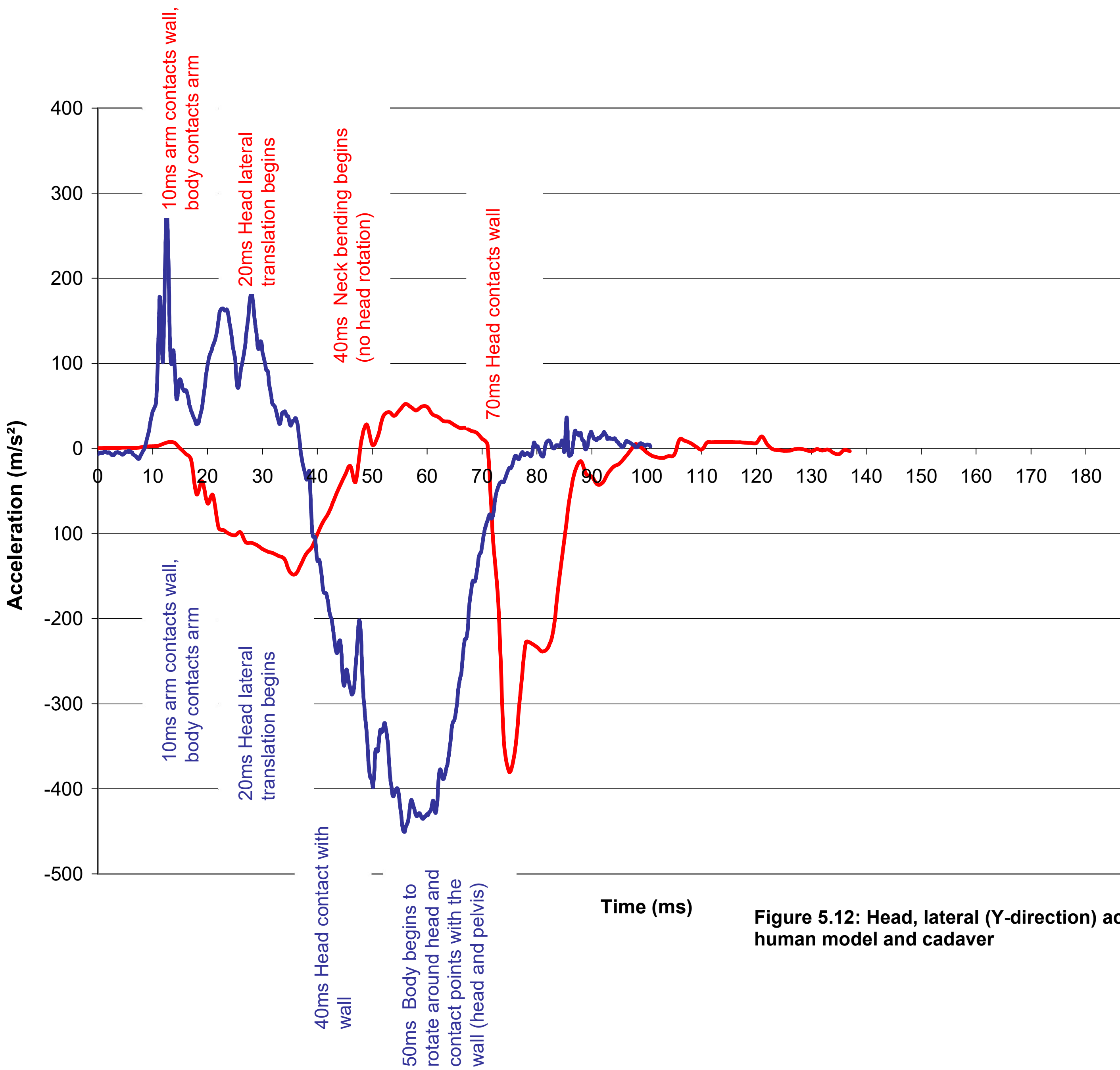


Figure 5.12: Head, lateral (Y-direction) ac human model and cadaver

5.7 Comparison of human model for lateral and oblique impacts using human subject data

The real world accident data in Chapter 3 indicates that low severity neck injury occurs in both near and far side impacts. These impacts are at angles predominantly between 0 and 90° with respect to the front of the vehicle. van der Horst et al. (2001) shows that the combined human model has only been validated for rear impacts. This section presents a comparison of the combined human model, for 45° oblique and 90° lateral impacts using human volunteer data. These simulations were undertaken to supplement the previous comparison against cadaver data, where the combined human model was tested with passive muscle functions. One of the features of the combined human model is that it can be set to simulate active muscle function within the neck. This feature also includes a delay period before the muscles activate, to simulate a more human-like response.

In this section the model is used with active muscle functions to simulate a human volunteer as in the experimental data. Also in this chapter the model is used to simulate formerly untested side impact angle of 45°. It is important to understand the model's behaviour at this impact angle, because in the following chapter two real world crashes are tested at a comparable impact angle.

The volunteer data used in this chapter for validation, is from the National Crash Survival Data Bank collected by the National Biodynamics Laboratory (NBDL). This data has previously been presented in research undertaken by Wismans et al. (1986), Ewing et al. (1977), Guccione et al. (2000), Ewing et al. (1975 & 1978).

5.8 Configuration of NBDL Volunteer Tests

The National Biodynamics Laboratory houses a database of acceleration tests using military volunteer subjects. The database includes horizontal sled tests in 90° (lateral) and 45° (oblique) directions. The subjects included in the database were young adult male volunteers. The anthropometry of each subject is listed in Wismans et al. (1986) and Ewing et al (1975 & 1978).

The subjects were seated on a HYGE accelerator (sled) with a V shaped harness shoulder belt configuration and lap belt. This belt is comparable to a military harness used in aircraft at the time of testing. The experimental sled and belt configuration can be seen in Figures 5.13 and 5.14. In the lateral impacts, the impacting wall was padded.

An additional belt was used in the lateral and oblique tests to limit the load on the subject's right shoulder. This belt was positioned horizontally across the chest and under the subject's right arm (the left arm was under the belt), this belt is illustrated in Figure 5.13. This figure also includes the padded wall in contact with the subject's arm.

Figure 5.13: NBDL experimental set up

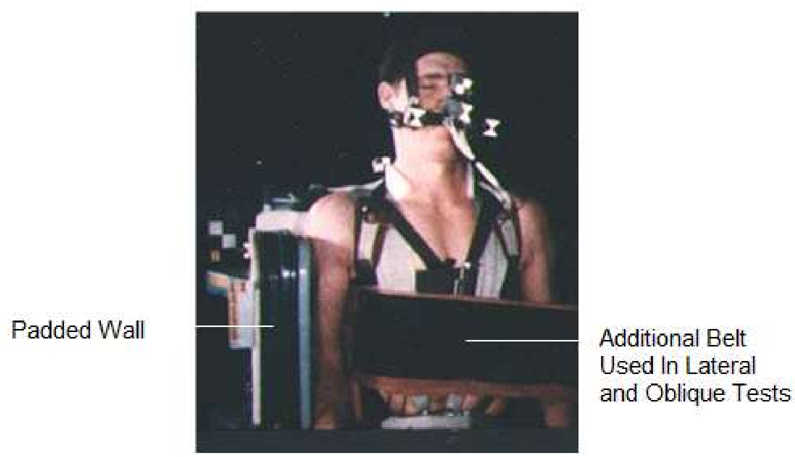


Figure 5.14 Detail of NBDL lateral impact belt configuration and padded wall.

The sled acceleration trace for the 90° lateral and the 45° oblique experiments is presented in Appendix 4. These acceleration traces were presented by Wismans et al. (1986). During the volunteer experiments the acceleration pulse was applied in one direction along the sled tracks. To achieve the different impact directions, the sled and occupant was rotated 45° or 90°.

5.9 Configuration of MADYMO model for simulation of 90° lateral volunteer test

The combined human model was positioned in a MADYMO 5.4.1 model of the NBDL sled, discussed in the previous section. The sled (Figure 5.15) was created using planes and ellipses. Planes were used to construct the seat base, seat back and the far side wall (on subject's left). The contacted side wall was modelled with two ellipses, one for the upper body and one for the lower body. With the original sled geometry, the simulation was stopped, terminating with numerical instability errors. To address this, the lower ellipsoid was extended to be longer than that of the real world sled to promote clean leg contact with the wall.

The v-shaped military style harness was created using finite element lap and shoulder belt pieces. This was to replicate the type of harness that the subjects were wearing in the volunteer trials undertaken by Wismans et al. (1986). The shoulder belts were reorientated about the end point to create the v-shape of the harness. The lap belt was positioned across the subject's lower torso. Belt elements attach these belts to the inertial space. A pre-simulation was undertaken to position the harness. In both the lateral (90°) and oblique (45°) simulations, a belt was fitted horizontally across the chest and was anchored to the left side, as with the experiment.

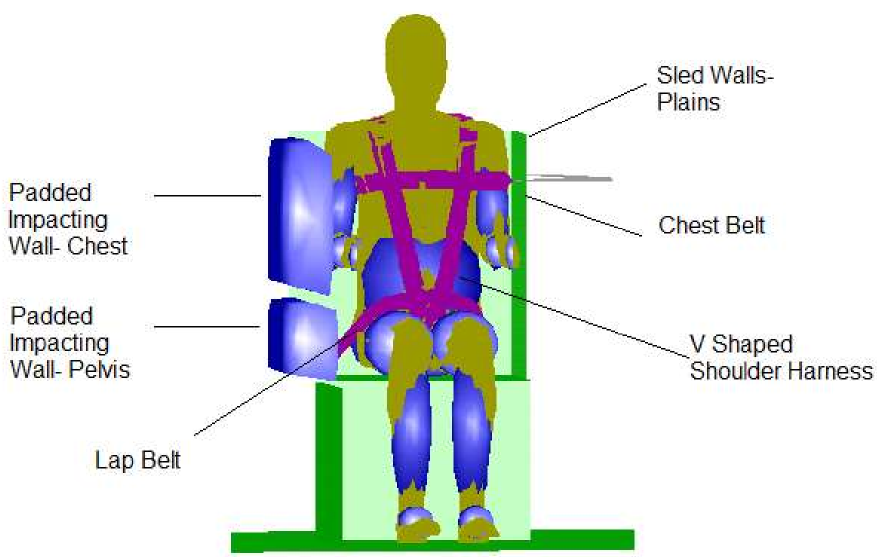


Figure 5.15 MADYMO 5.4.1. Computer simulation of NBDL lateral sled test

The sled acceleration trace for the 90° lateral and the 45° oblique trial (Appendix 5) presented by Wismans et al. (1986) was applied to the pelvis of the human model.

Both simulations were run for 250ms, this time allowed for the model to impact with the wall, rebound off the wall and to settle. The muscle functions for both simulations were activated at 100% for the left and right flexors, extensors and sternocleidomastoid muscles. A delay period to represent the volunteer's reaction time was 50ms as tested by de Jager (1996). The 25ms delay period recommended by van der Horst et al. (2002) was not used. With this short delay time, the head/neck began to extend while the rest of the model was stationary, suggesting that the muscles were activating too soon.

5.10 Comparison of results between combined human model with NBDL volunteers

The results for the 90° lateral and oblique simulation have been presented in this section. Two sets of data have been presented for each simulation. The first data set shows the kinematic images of the simulation. For the 90° lateral simulation the results are compared to three images (NBDL, 2002) of the NBDL lateral volunteer test. The oblique simulation results are presented alone, as there are no images available. The second data set is the acceleration histories for the first thoracic vertebra. This predicted data has been presented against the "average" lateral acceleration data (of the first thoracic vertebra for all subjects tested) in the NBDL tests. The predicted head acceleration data has been presented only for the model, as the matching data for the human volunteers was not available.

The kinematic image results for the 90° lateral impact trial, from the combined human model are presented in Table 5.7. This table has 3 rows of data. The first row is the time in milliseconds. The second row is the kinematic image data and the thirds row is a description of the key events.

The kinematic images of the volunteer trial have not been compared to the model in Table 5.7. Two images of an unknown trial have been presented in figure 5.16. These images have been presented to provide a guide as to how the model compares to a human volunteer. These images illustrate the key events that occurred during the volunteer tests.

The kinematic images of the combined human model in the Table 5.7 show that at 20ms the arm contacts the wall and at 40 ms the pelvis contacts the wall. At 60ms, lateral bending of the neck begins coupled with head rotation, this continues until the end of the simulation at 120 ms.

Only 2 frames are available illustrating the volunteer motion following contact with the wall (Figure 5.16), these are timed at 0 ms and 100 ms (normalised for body contact with wall at 0 ms). At 0 ms, it is shown that arm is in contact with the wall and the head is beginning to laterally translate. The images show that at 100 ms the human subjects head is rotating coupled with lateral bending. It is also evident in this image that the subject's shoulder and scapula has elevated. These key events are consistent with the kinematic/video footage of the human model shown in Table 5.7.

When comparing the kinematic images from the model to the experimental data the same key events are visible. These events occur at similar time in both trials. During the volunteer trial (illustrated in Figure 5.16) at 0 ms, arm contact occurs. This is also the point where lateral head translation begins. At 100 ms shoulder elevation is visible. At this point the neck is laterally bent and lateral head rotation has occurred. In the human model simulation similar key events are visible. Following arm contact at 0 ms, the head begins to laterally translate. Also following arm contact the shoulder elevates. Lateral neck bending and head rotation begins at 60ms and is completed between 100 ms and 120 ms.

Table 5.7: Kinematic results of simulation of NBDL 90° lateral human volunteer

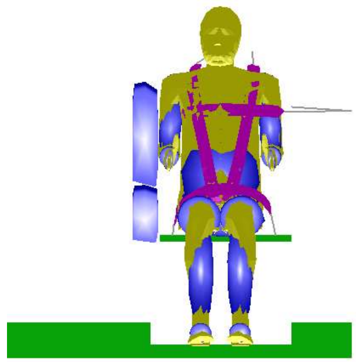
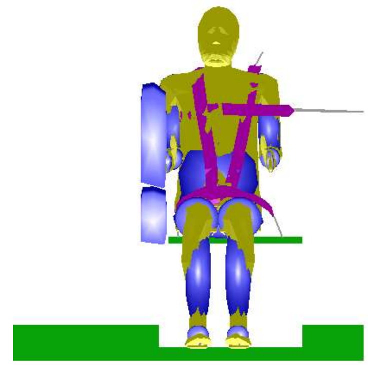
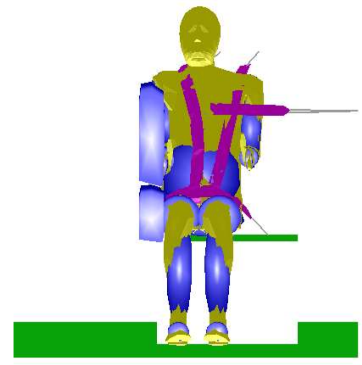
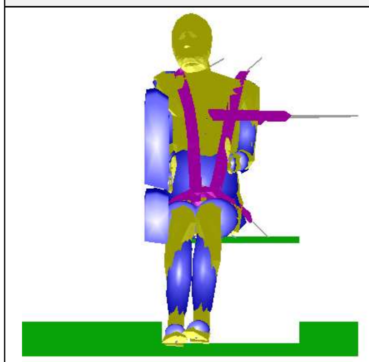
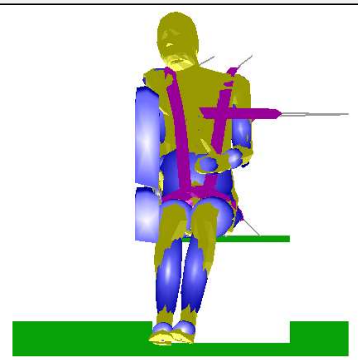
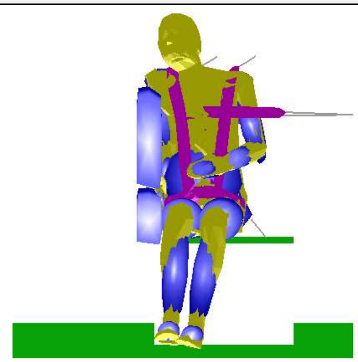
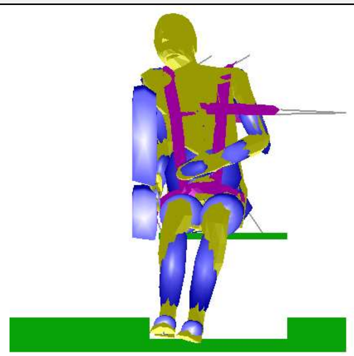
Time: 00 ms	Time: 20 ms	Time: 40 ms	
			
	20ms: arm contacts sled wall, slight lateral neck extension, slight lateral neck translation		
Time: 60 ms	**Time: 80 ms**	**Time: 100 ms**	**Time: 120 ms**
			
60- 120: body continues to contact wall as result shoulder (scapula) elevates, pelvis begins to rotate neck lateral bending is coupled with head rotation, body continues to contact wall as a result shoulder (scapula) elevates, pelvis begins to rotate neck lateral bending is coupled with head rotation			

Figure 5.16: Kinematic results of NBDL 90° lateral human volunteer

0 ms: (normalised)	100 ms: (normalised)
	
Lateral Translation of head begins	Lateral Bending of the head coupled with head rotation, shoulder elevation

The lateral acceleration of the first thoracic vertebra for the human model and the volunteer (NBDL) for the 90° trial is presented in Figure 5.17. The human model is presented in red and the human volunteer is presented in blue. The acceleration traces illustrate a very close likeness between the model and the human volunteer. It is important to note that there is a difference between the two traces in the time when acceleration changes from zero. This is a result an estimate being made of the initial position of the volunteer in the experimental trials to reconstruct for the combined human model. This difference seems to suggest that the volunteer was positioned closer to the wall that the human model.

The human volunteer trace shows that rapid T1 deceleration begins, slowing slightly at 40ms and peaking at -140 m/s^2 at 60 ms. Following peak deceleration the trace returns to 0 at 140ms, prior to reaching this point there are two slight increases in the rate of deceleration, appearing as a wave pattern on the trace at 80 ms (-80ms^2) and 95 ms (-60 ms^2).

The T1 acceleration from the human data is comparable to the T1 acceleration from the computer simulation using the combined human model. There is also rapid deceleration peaking at -152 m/s^2 at 41ms. This occurs at a similar time as the peak deceleration in the volunteer trials. Following this peak deceleration there is also a slight wave pattern with the T1 deceleration slows to -111 m/s^2 at 51ms but briefly increases until 57 ms at -131 m/s^2. At this point the deceleration slows until 85 ms when it returns close to 0.

The difference in the wave like deceleration being more pronounced in the pattern between the human trials than the model, may be described by the shoulder/scapula elevation that is visible following torso contact with the wall. This shoulder elevation is visible in the kinematic images (Table 5.7 and Figure 5.16) of both trials, it is likely that the shoulder in the human model has a greater stiffness than the human volunteer resulting in a more rapid rebound. Another likely cause of the difference in the wave like deceleration could be related the foam covering impacted wall. The contact functions used to define the contact between the human model and the foam wall were general values for foam derived from the MADYMO library of sample impacts. General values were used as the exact values regarding the foam used in the experiment were not available in the literature.

The lateral head acceleration for the human model for the 90° is presented in Figure 5.18. The head acceleration for the NBDL trial was not available for comparison. Head acceleration remains at 0 until 15ms. This is followed by rapid deceleration peaking at 97 ms^2 at 50 ms. It is important to note at 40ms that the data becomes slightly noisy. This coincides with pelvis contact with the wall. The head deceleration diminishes at a constant rate until the end of the trial.

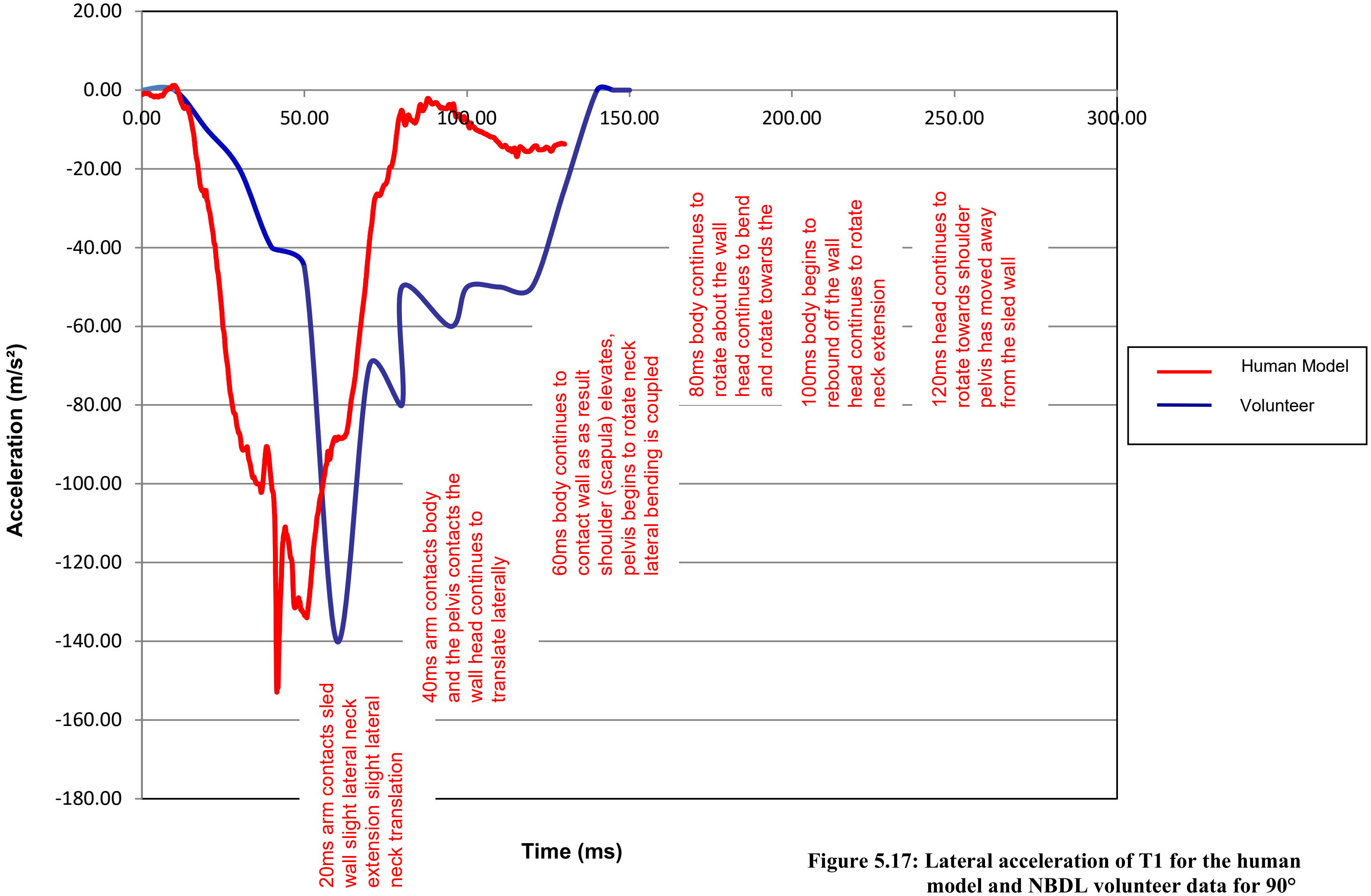


Figure 5.17: Lateral acceleration of T1 for the human model and NBDL volunteer data for 90° trial

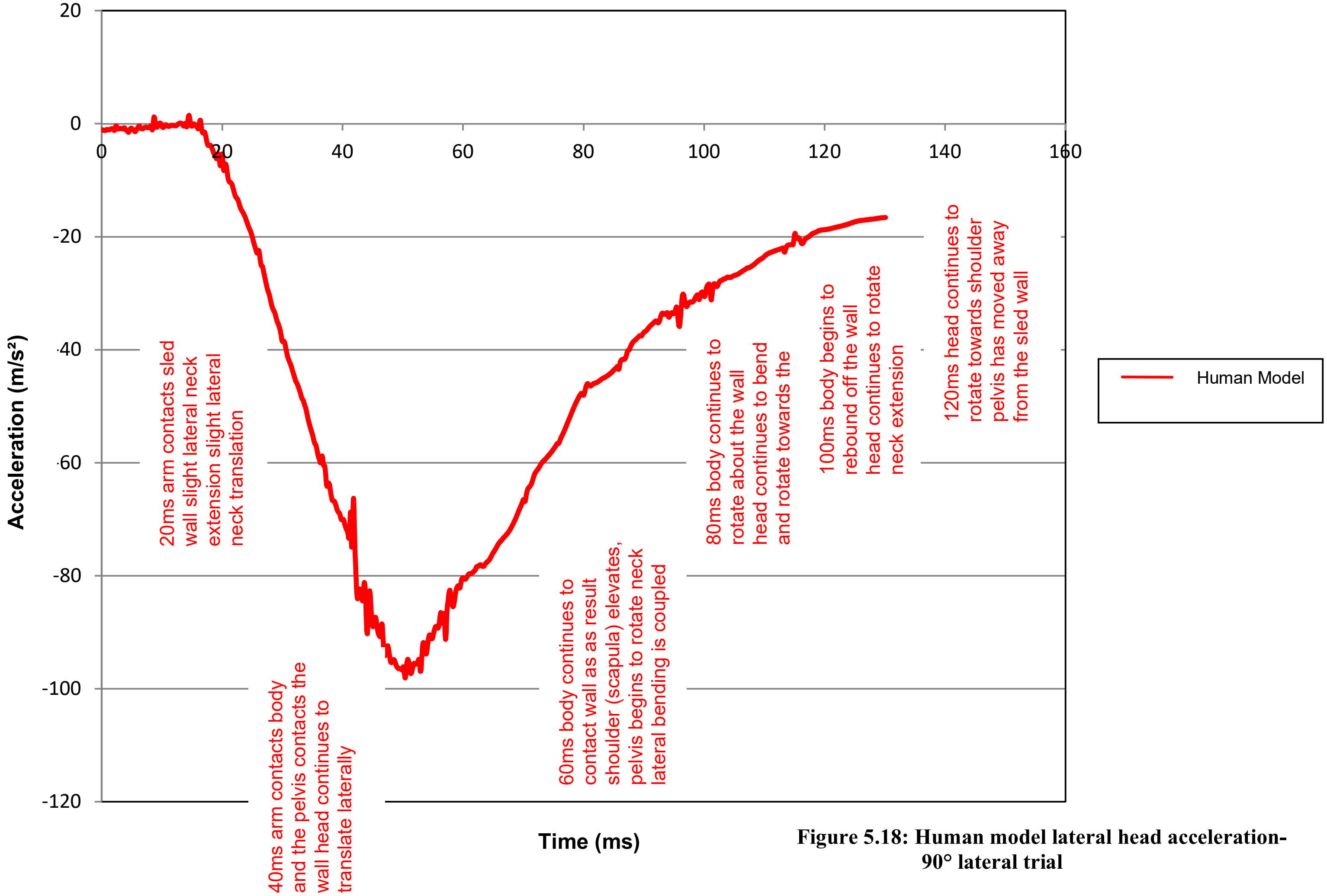
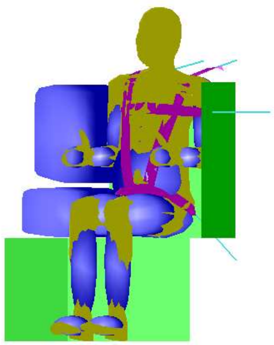
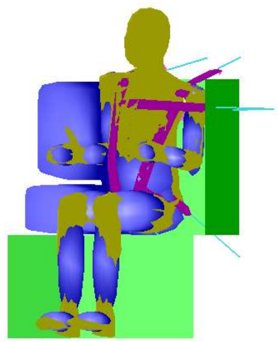
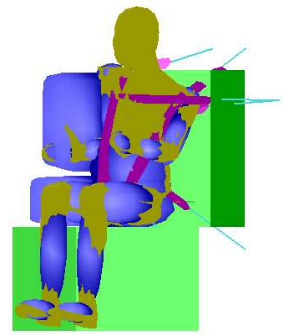
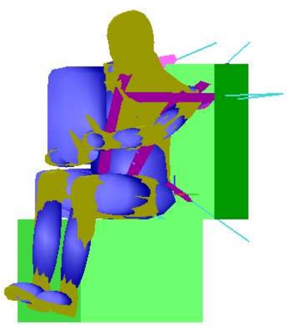
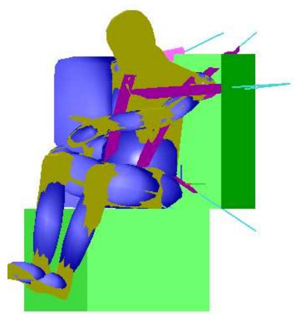
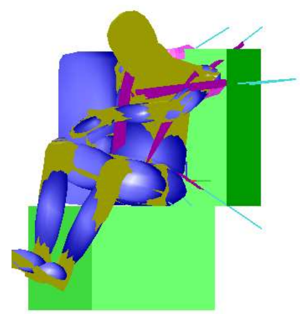


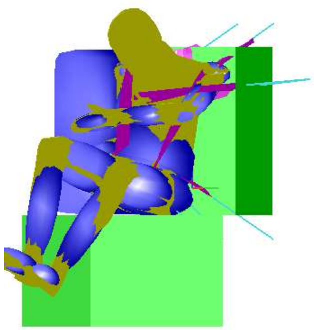
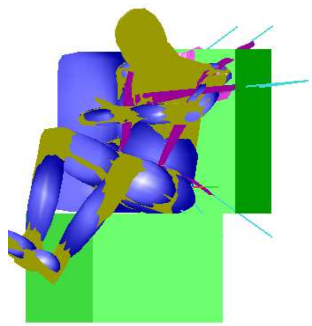
Figure 5.18: Human model lateral head acceleration-90° lateral trial

Table 5.8 contains the kinematic images for the human model in the 45° oblique impact trial. This computer simulation was undertaken to replicate NBDL volunteer experimental trials, where the subjects were exposed to a 45° lateral impact. This table has 3 rows of data, the first row is the time in milliseconds. The second row is the kinematic image data and the thirds row is a description of the key events. There are no images available of the NBDL trials to compare the kinematics events.

Prior to body contact, the body has translated towards the wall in a 45° direction. During this time, the head translates with the body. At 0 ms the body contacts the arm and at 30 ms the hip contacts the wall. Also at this time, the shoulder that is in contact with the wall elevates, and the opposite shoulder begins to rotate about the contact point, and neck flexion begins. At 40 ms the hips begins to rotate about the point of contact. Head rotation begins at 60 ms coupled with rapid neck flexion. This continues until the end of the trial. When comparing the kinematic images for the 45° oblique trial in Table 5.8 to the kinematic images of the 90° lateral trial in Table 5.7 the images from the oblique trial illustrate a combination of both frontal and lateral motion.

Table 5.8: Kinematic results of simulation of NBDL oblique human volunteer

Time: 00 ms	Time: 20 ms	Time: 40 ms
		
10ms: arm contacts body	30ms: hip contacts the wall, right shoulder begins to rotate towards the wall, head flexion begins	40ms: the outside hip begins to rotate about the wall
Time: 60 ms	**Time: 80 ms**	**Time: 100 ms**
		
60-100ms: Head rotation begins, coupled with head flexion		

Time: 120 ms	**Time: 140 ms**
	
120-140: Head rotation continues, coupled with head flexion	

The results presented in Figure 5.19 are the lateral acceleration of the first thoracic vertebra for the NBDL human subject (blue) against the combined human model (red) in an oblique impact. The T1 acceleration for the NBDL subject rapidly decreases to with a peak of -210m/s² at 50ms.The deceleration slows with a wave pattern at 80ms at -80 m/s² followed by a slight increase (100 m/s², 95ms) followed with another decrease at -80 m/s² at 100ms and peaking at 150 m/s² at 120ms.

The lateral acceleration trace for the combined human model is more noisy than the human volunteer data. There is an initial deceleration that peaks at -91ms² at 10ms. This deceleration slows at 17ms to -20ms². Following this, the deceleration increases to -142ms² at 30ms at this point the deceleration slows again reaching zero at 42ms. Acceleration then begins, peaking at 70ms² at 51ms. This acceleration coincides with the torso and pelvis beginning to rotate about the wall and is likely to be a result of the horizontal chest belt providing resistance against the rotation. This acceleration slows returning to zero at 65ms. At this point the data becomes extremely noisy at the body continues to rotate about the wall and interacts with the belt system.

One of the key differences between the volunteer T1 acceleration and the human model in the oblique trial is the magnitude of the initial acceleration. The human model has a peak deceleration of -142ms² and the human volunteer data has a greater peak deceleration of -210ms². Another difference is the human model trial has an acceleration peak that is not present in the human volunteer trial. It is evident from the human model trace that the belts in the harness are having an influence of the T1 acceleration results due to the noisy data throughout the trial. It is likely that at the point the model's T1 has begun to accelerate, the horizontal chest belt is resisting the rotation of the torso of model following contact with the wall. As the specific variables of the harness used in the volunteer trials were not available the harness of the model was based on belt parts and variables from trials in the MADYMO library. This is a likely cause of the variation in the way that the harness interacts with the human model compared to the human volunteer.

The lateral head acceleration for the human model for the oblique is presented in Figure 5.20. The head acceleration for the NBDL trial was not available for comparison. At 0ms there is initial head acceleration of 10ms². This is unexpected and may be a result of the active neck muscles. This is followed by rapid deceleration that peaks at 72ms² at 29ms. This deceleration slows until 40ms when the data becomes noisy. The head deceleration briefly passes through 0 at 43ms and again at 65ms, followed by a peak of 10ms². This is followed by a deceleration peaking at 37ms² at 100ms. This deceleration slows until the end of the trial. As there is no data available for the volunteer head acceleration, this was presented for reporting purposes only.

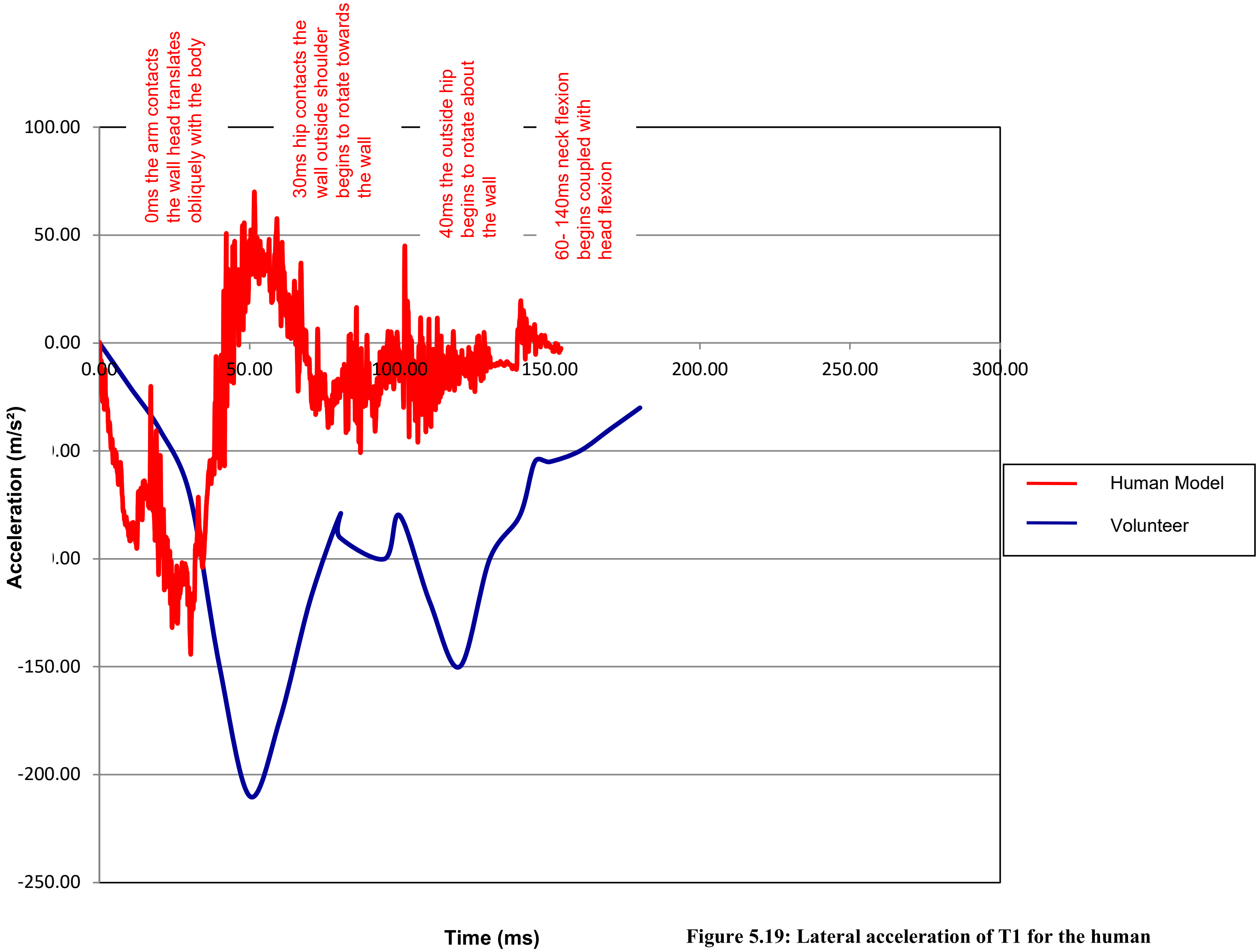


Figure 5.19: Lateral acceleration of T1 for the human model and NBDL volunteer data for oblique trial

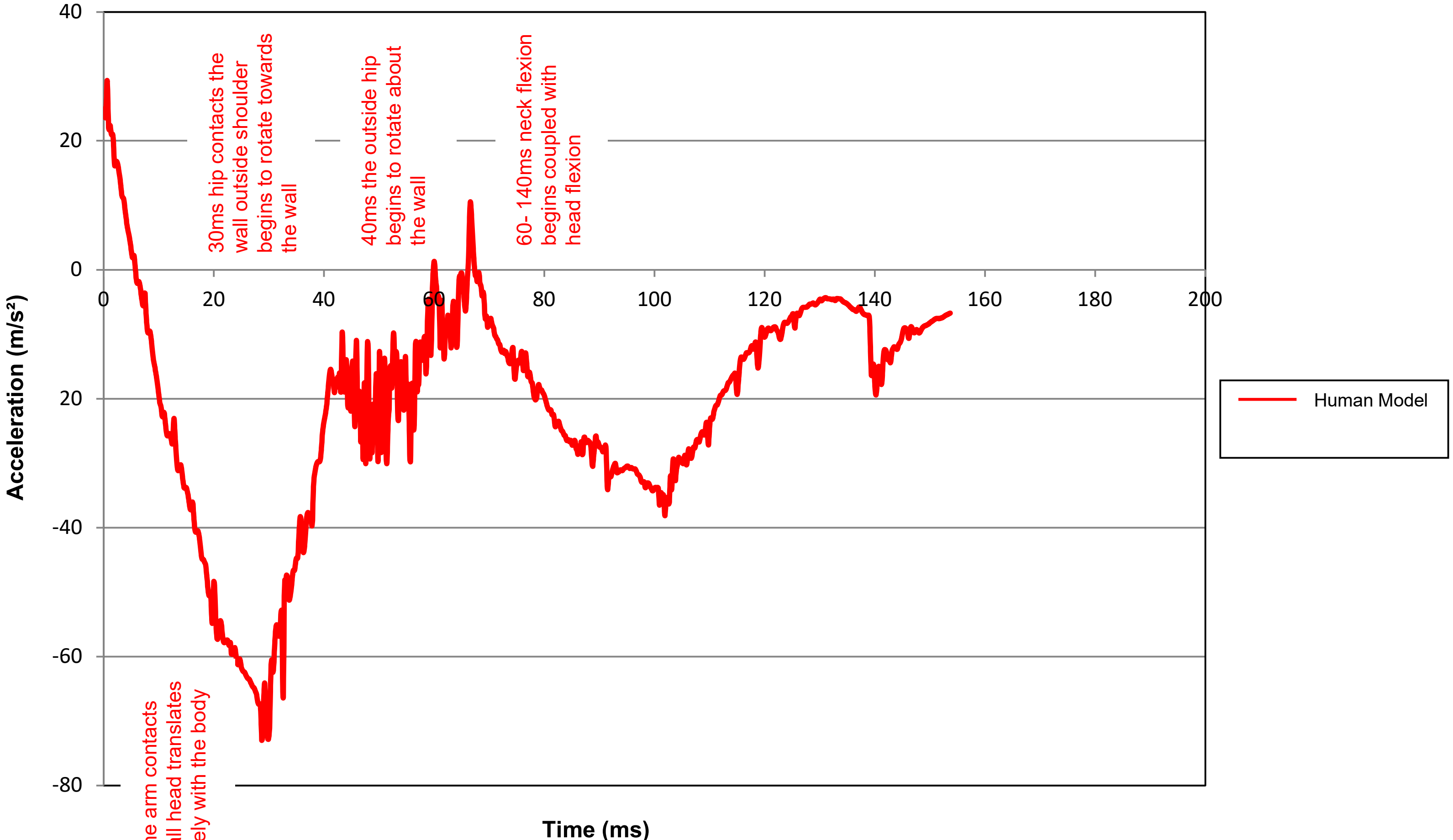


Figure 5.20: Lateral acceleration of the head for the human model - oblique trial

5.11 Summary of findings

The human body model in MADYMO is as commercially available product, created to predict human response in vehicle crashes. In this research, the human model has been combined with the detailed neck model. Prior to this piece of research, the human body model on its own had been validated for front, side and rear impacts at varying impact magnitudes. The detailed neck model had been validated for front and rear impacts.

The real world crash data presented in Chapter 3 showed that side impact crashes also frequently occur at impact angles of less than 90°. Due to this fact, in this chapter the combined human model has been compared to lateral 90° and oblique 45° impacts.

The data used to for these comparisons included head and first thoracic vertebra (T1) and when available kinematic (video) data. This data came from two sources. Post mortem human subject (cadaver) data used in side impact sled test at Wayne State University was used to validate the model with passive muscle functions in a 90° lateral simulation. Human volunteer data of sled tests undertaken by the Naval Biodynamics Laboratory (USA), was used to validate the model for the 90° lateral and oblique simulations with active muscle functions.

5.11.1 Comparison with cadaver data (passive muscle functions)

When the combined human model was compared to the Wayne State University data the same key events were observed in the kinematic data in a lateral 90° impact. When comparing the lateral acceleration data for the first thoracic vertebra for these trials, there is a good likeness between the trials. In these trials the two traces following a similar path with comparable peak values. The peak first thoracic lateral acceleration for the cadaver was 714 m/s^2, with the peak acceleration for the model 809 m/s^2. In spite of the good likeness between the cadaver and the human model for the kinematic and T1 data, there is variation in the lateral head acceleration data. This data shows that the magnitude of the acceleration is comparable between the two trials although there was an additional peak in the acceleration data for the human model. This finding suggests that the difference in the flexibility of the neck between the model and the cadaver, with the passive muscle functions in the model less flexible than that of the cadaver neck. The peak head lateral acceleration for the cadaver was -380 ms^2, with the peak acceleration for the model -451 ms^2.

5.11.2 Comparison with volunteers (active muscle functions)

When the simulation results from the human model was compared to the NBDL 90° lateral tests the same key events were observed during in the kinematic images available. This is the same for the first thoracic lateral acceleration, where both trials show very similar traces. The peak T1 acceleration for the volunteer trials was -152 m/s^2 and for the human model -140 m/s^2. The lateral head acceleration data was not available for the NBDL trial. The peak head lateral acceleration for the model was 97 m/s^2.

The kinematic results from the comparisons between the human model and the NBDL 45° oblique simulation also show similar events in both trials. In spite of this there are differences in the lateral acceleration of the first thoracic vertebra trace. There is more variability with the model data, than the volunteer data. This is most likely due to the interaction between the seatbelt and the combined human model. The volunteer peak lateral T1 acceleration was higher at -210 m/s^2 than the human model at -142 m/s^2. The peak head lateral acceleration for the model was 72 ms.

The comparisons show there is good likeness between the human body model and the post mortem human subject and volunteer trials for the kinematic data and the peak acceleration values. These results for makes it a suitable tool to use to evaluate neck mechanism in side impacts of varying lateral/side impact angles. The finding also shows the need to utilise the models active muscle functions when simulating lateral impacts with live human subjects, such as real world impacts.

The findings in this chapter suggest that the model is a suitable tool for further investigation into head and neck kinematics in side impact. The results from the model are the greatest likeness to human cadaver/ human volunteers in 90° simulations.

5.12 Summary Table of Acceleration Data

Table 5.9 presents a summary of the model output data from the simulations undertaken in this chapter. The results have been discussed in the respective results sections in this chapter.

Table 5.9: Summary of Acceleration Data

Variable	Human Cadaver	Model
90°Lateral Cadaver		
First Thoracic Vertebrae		
Initial Peak Deceleration	714 ms^2	809 ms^2
Time of Peak Deceleration	15 milliseconds	19 milliseconds
Arm Contact Deceleration Delay	8 milliseconds	10 milliseconds
Time Acceleration crosses zero	22 milliseconds	27 milliseconds
Peak Acceleration	174 ms^2	68 ms^2
Head		
First Peak Acceleration	282m/s^2	102 m/s^2
Time of First Peak Acceleration	13ms	25ms
Second Peak Acceleration	165m/s^2	52 m/s^2
Time of Second Peak Acceleration	23ms	56ms
Third Peak Acceleration	182m/s^2	no third peak
Time of Third Peak Acceleration	28ms	no third peak
Deceleration	-451m/s^2	-380m/s^2
Time of Deceleration	56ms	75ms
Time Acceleration returns to zero	80ms	80ms
90°Lateral Volunteer		
First Thoracic Vertebrae		
First Peak Deceleration	-140ms^2	152 ms^2
Time of First Peak Deceleration	60ms	41ms
Time Return to Zero	140 ms	-
Second Peak of Deceleration	-80 ms^2	-111ms^2
Time of Second Peak Deceleration	80ms	51ms
Third Peak Deceleration	-60 ms^2	-131 ms^2
Time of Third Peak Deceleration	95ms	57ms
Time Acceleration returns to zero	-	85ms
45°Lateral Volunteer		
First Thoracic Vertebrae		
First Peak Deceleration	-210m/s^2	-91m/s^2
Time of First Peak Deceleration	50ms	10ms
Second Peak of Deceleration	-80 m/s^2	-20 m/s^2
Time of Second Peak Deceleration	80ms	17ms
Third Peak Deceleration	-100 m/s^2	-142m/s^2
Time of Third Peak Deceleration	95ms	30ms
Fourth Peak Deceleration	-80 m/s^2	0
Time of Fourth Peak Deceleration	100ms	65ms
Fifth Peak Deceleration	-150m/s^2	-
Time of Fifth Peak Deceleration	120ms	-

Chapter VI
Crash Simulations of Side Impact Cases Resulting in Low Severity Neck Injury

6.1 Introduction

To further understand LSNI from side impact, an investigation into occupant kinematics and head and T1 acceleration was undertaken and is presented in this chapter. The investigation used computational crash reconstructions of real world crashes. Three different crash configurations were simulated, with two cases for each configuration, one case where an occupant received a LSNI, and one where the occupant did not receive a neck injury (used for comparison). These simulations were taken from the Monash University Accident Research Centre database, previously discussed in Chapter 2. Each case was modelled using two methods. Firstly, *Human, Vehicle, Environment (HVE)* crash reconstruction software was used to establish the acceleration pulse of the crash. This acceleration pulse was then used in MADYMO 5.4.1, including the combined human model seated in a vehicle model. This second computer simulation was undertaken to produce detailed predictions describing the occupant's kinematics during the crash.

The body of knowledge regarding LSNI that has been presented in the earlier chapters has been drawn on to undertake the computer simulations of crashes presented in this chapter. Figure 5.1 illustrates how the findings in the earlier chapters have supported the development of the simulations. The findings from the literature review in Chapter 2 and the analysis of crash data presented in Chapter 3 were used to determine the common factors to be included in the cases. The common factors included in the selected cases are presented in section 6.3.1. The combined human model was validated and assessed for its suitability to be included as the occupant model in the crash simulations (Happee 2000). This previous work is progressed in this chapter to add to the body of knowledge on LSNI from side impacts. This includes determining the crash acceleration for each of the chosen crash cases simulated. The acceleration was determined using HVE (Human, Vehicle, Environment). The crash simulations and the findings from those simulations also contribute to the body of knowledge for LSNI from side impact. The methodology of reconstructing the real- world crashes using the combined human model has not been used before and is unique to this thesis. These simulations undertaken, investigated three different crash angles 90°, near side, 90° far side and 45° (oblique) near side. These crash angles were chosen as they represent the angles that were shown to be most common amongst the sample group with LSNI, in the crash data. The results from the crash simulations undertaken in this chapter have been compared to the literature in Chapter 2 regarding the mechanism of LSNI.

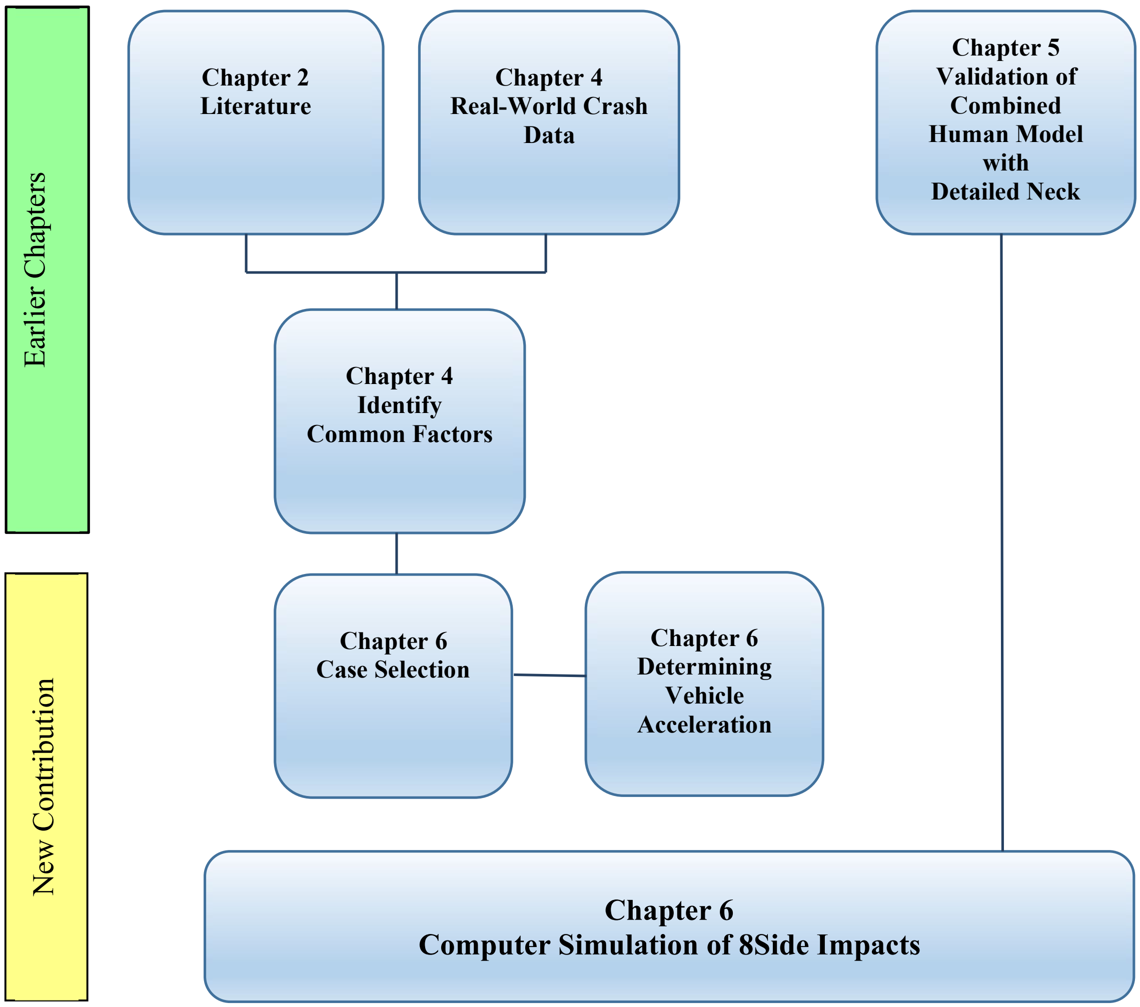


Figure 6.1: Relationship between chapters and body of knowledge

6.2 Experimental Setup of Computer Simulations of Six Side Impact Crashes

To analyse the mechanism of LSNI from side impact a reconstruction of real world crashes has been reported in this chapter. The reconstructions were undertaken using HVE Human, Vehicle, Environment and MADYMO software packages. The real-world crashes were cases from the Monash University Accident Centre crash database. The reconstruction of the cases has been undertaken in three parts, case selection, crash reconstruction using HVE to predict the crash pulse and reconstruction of occupant mechanics undertaken using

MADYMO 5.4.1. The MADYMO model included a vehicle model and the combined human model. The methodology of using HVE to determine crash acceleration (crash pulse), and programming that data into MADYMO to reconstruct real-world crashes has also been used by Hasija et al. (2007, 2009). An outline of this research has been presented in section 6.2.2. Franklyn et al. (2003, 2005a, 2005b) also used HVE to determine the crash acceleration (crash pulse) for real- world crashes to investigate head injury. Although, the research methodology undertaken by Franklyn (2003, 2005a, 2005b) differs in part to the work presented in this thesis as the crash acceleration obtained in HVE, is used in crash tests. An outline of this research has also been presented in section 6.2.2.

6.2.1 Case Selection

Six cases were selected from the Monash University Accident Research Centre database. In three of these cases an occupant received a LSNI and in three of the cases the occupant did not. The cases with an occupant receiving a LSNI were chosen as they displayed the typical characteristics shown in Chapter 4. From this data it was shown that LSNI from a side impact were more likely to occur at impacts between 0 and 90°. Due to this finding the crash configurations chosen to model in this study are near side 90° impact, far side 90° impact and a near side oblique impact of 45°.

Although, the data showing that female occupants are more likely to be injured than males, only cases with male occupants were considered. This is because the human model used in the study has the geometry of a 50th percentile male. Only cases with drivers were modelled as the data in Chapter 4 shows that males are most likely to be injured when they are drivers. A summary of the case data (Tables 6.4, 6.6 & 6.8) has been presented in the following sections with the results for each impact angle. These tables show the occupant age, weight, seat belt usage and airbag deployment. The case data also includes information regarding injuries received, the vehicle details, and impact details. It also presents the two vehicles for each crash configuration presented against each other so that these details can be readily compared.

6.2.2 Computational Simulation Software Package- Human, Vehicle, Environment (HVE)

Human Vehicle Environment (HVE) is accident reconstruction software that can be used to predict the acceleration pulse for vehicles involved in a crash. The software simulates the trajectories of the vehicles allowing the outcome of the crash to be predicted. The accuracy of the software is presented through validation of the HVE software is presented in Day et al. (2004, 2001, 1996, 1990, 1989). HVE has been validated for both front and side impacts at both 90° and oblique angles. This makes the software suitable to reconstruct real-world crashes at the range of different side impact angles that have were shown in Chapter 4 to result in a high incidence of LSNI. To reconstruct a case, the occupants and impacting objects (vehicles and barriers) were chosen from the library and placed in a crash environment, that are available to be selected within the software. The human component of this software was not used in this study as it was not required to obtain an acceleration pulse.

The methodology of using HVE to determine crash acceleration (crash pulse) has been used previously by researchers to investigate head injury. Franklyn et al. (2003) reconstructed real world crashes to investigate head injury. This study used real world data to predict the injury outcomes using computer models. The real-world data was collected by the Monash University Accident Research Center Investigations Team. The method of collection of the real-world data is the same as outlined in section this research thesis. HVE simulations [the real world. Where the methodology used by Franklyn et al. (2003), crash tests were undertaken using the predicted data from the HVE models to compare the head injury results from the crash test to the real world medical data. This methodology was also used by Franklyn et al. (2005a, 2005b). The investigation undertaken by Franklyn et al. (2003) included reconstructing two lateral impacts. One crash configuration was a vehicle to vehicle crash where no occupants received a head injury (AIS0) and the other a single vehicle crash with a pole, where the driver received an AIS5 head injury. Four real world crashes were investigated by Franklyn (2005a, 2005b) which had resulted in the following head injuries, 1. AIS1, 2. AIS4, 3. AIS5, 4. AIS multiple injuries. When undertaking this work, Franklyn et al. (2003, 2005a, 2005b) found that accuracy of the crash test results were dependent on the prior computer simulations undertaken in HVE. It was identified that the stiffness values in the HVE software influenced the results of delta V as only area stiffness could be programmed (such as front, back and side etc). Point to point stiffness values could not be included in the HVE simulations undertaken by Franklyn et al. (2003, 2005a, 2005b), to improve on the model accuracy.

Hasija et al. (2007, 2009) also used real world and computational modelling of crashes to investigate head injury. HVE software was used to determine the crash parameter of crash pulse and this was then programmed into a MADYMO to reconstruct the occupant mechanics, this is the same methodology used in this research thesis to investigate neck injury. The crash configurations investigated by Hasija et al. (2007, 2009) differ to the crash direction investigated thesis in that they are frontal impacts and frontal offset impacts. One of the key findings in the work undertaken by Hasija (2007) was that in the "no brain injury case", the output changed significantly based as the variability of the assumed parameters, entered in place of the unknown parameters from the real-world crashes. This finding shows the accuracy of results from computer simulations of real world crashes improves as more measured parameters are available from the crash investigation to be programmed into the simulation. Hasija et al. (2009) progressed the work presented by Hasija et al. (2009) by optimizing the methodology . Hasija et al. (2009) found that using HVE and MADYMO combined to reconstruct two real world crashes was a sound methodology, when data wasn't available to undertaken the reconstructions "deterministically". It was also reported by Hasija et al. (2009) that this type of computerised reconstruction method is a benefit, as it can provide understanding as to how the occupant moves inside a vehicle in a crash. The specifications of the simulations are presented in the following sections.

6.2.3 Vehicle Models

HVE has a vehicle library consisting of a range of vehicles (vans, truck and included) as well as a series of barriers. In the crashes chosen for further investigation, the case vehicle was the Holden Commodore. This vehicle was not available in the HVE library at the time this research was undertaken. The vehicles available in the HVE vehicle library were models available in the U.S.A, due to this a model of vehicle was chosen to most closely matched the geometry of the Holden Commodore. Based on geometry, the vehicle that most closely replicated the Holden Commodore was the Opel Omega. Table 6.1 shows a comparison between the measurements of the vehicle.

	Holden Commodore VT	Opel Omega	Variation in Measurements
Length mm	4884	4914.9	+30.9
Width mm	1842	1788.2	-53
Height mm	1450	1465.6	+15.6
Wheelbase mm	2788	2730	-58.0

Table 6.1 Comparison of Specification between the Holden Commodore and Opel Omega

The HVE software allows for modifications to be made to the structure of the vehicle. These modifications are listed in Table 6.2. The only modification made to the Opel Omega base model was adjusting the vehicle mass to match the mass of the Holden Commodore. For the rest of the listed variables, the default variables in the model were used.

Table 6.2: Vehicle Parameters available for modification in HVE

Vehicle Parameters	
Sprung Mass	
	Inertias (including mass)
	Centre of Gravity
Unsprung Mass	
	Wheel Location
	Tire Data
	Suspension Data
Exterior	
	Exterior dimensions
	Stiffness Coefficients

6.2.4 Crash Scenario

A library of environments is available in the HVE software, ranging from minor roads to highways and open proving grounds. The crash scenario is created by placing the vehicles in an environment. A list of the factors defining the crash scenario that can be modified is provided in Table 5.3. In this study the vehicle initial positions, initial speeds and driver controls such as steering and breaking are defined were modified to ensure that the collision deformation characteristic (CDC) for both vehicles calculated by the software matched the measured code for the real world cases.

Table 6.3: Crash Scenarios Available for Modification

Crash Scenario Parameters
Vehicle Initial Position
Vehicle Initial Velocity
Driver Controls, Steering
Driver Controls Braking
Drivers Controls Throttle
Wheels, Tire Blow out
Wheels, Displacement
Accelerometers

6.2.5 Vehicle Accelerations- Output from HVE

The lateral and frontal vehicle accelerations have been presented in Figures 6.2 to 6.4. The LSNI frontal X acceleration is the dashed red line, the LSNI lateral Y acceleration is the solid red line, the control frontal X acceleration is the dashed blue line and the control lateral Y acceleration is the solid blue line.

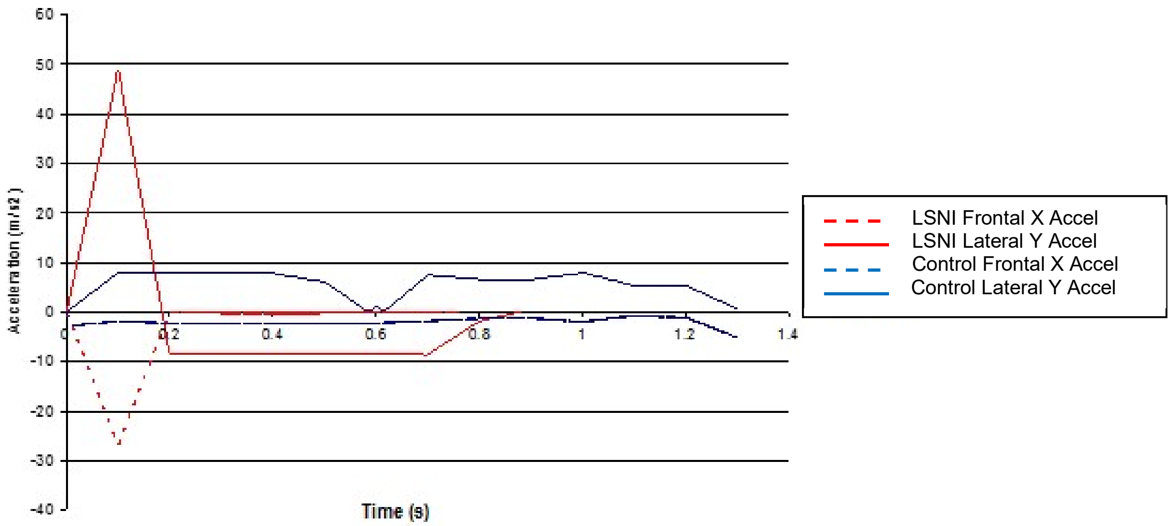


Figure 6.2: 90° near-side acceleration

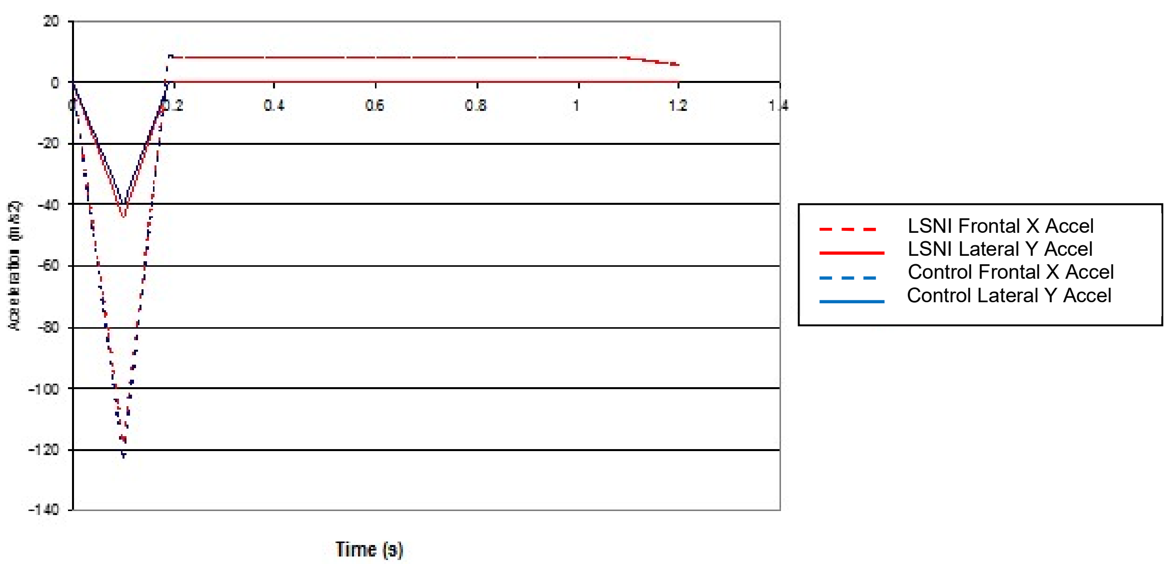


Figure 6.3: 90° far-side acceleration

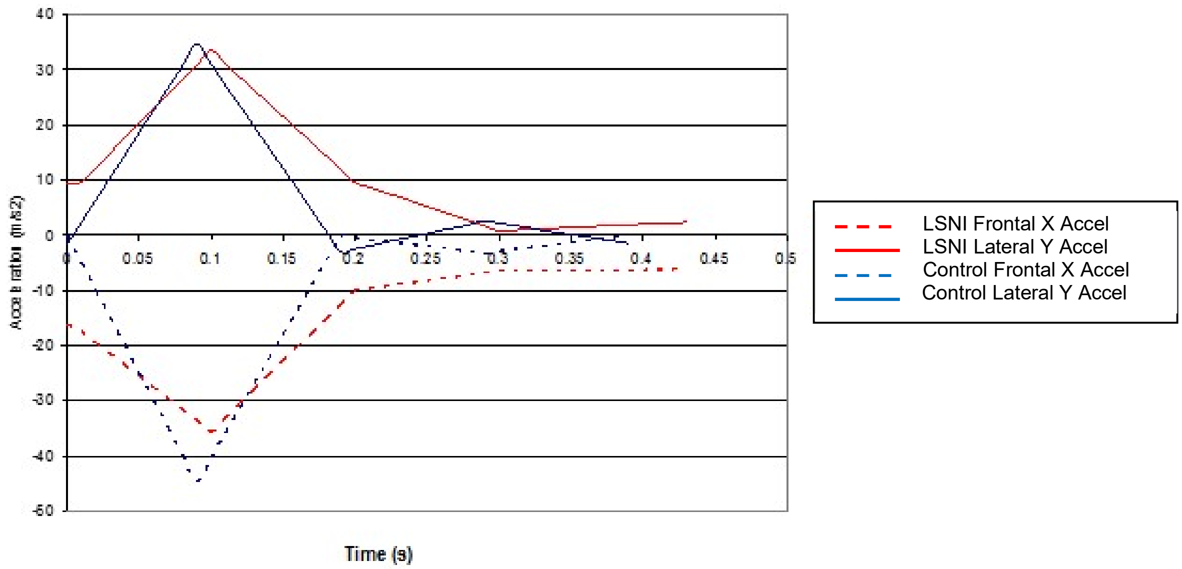
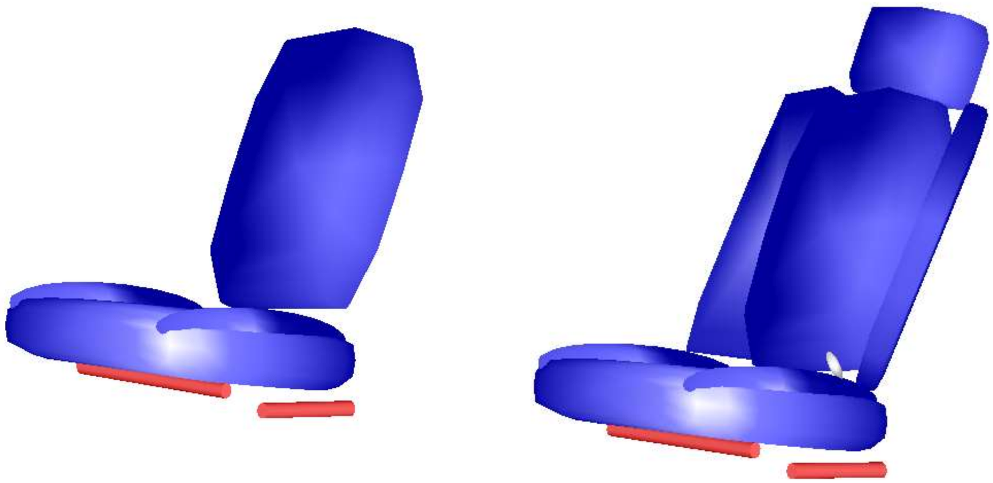


Figure 6.4: 45° (oblique) nearside vehicle acceleration

6.2.6 Computer Reconstruction of Real World Crashes Using MADYMO

Two existing MADYMO models an occupant model (combined human model) and a vehicle model were modified and combined to undertake a computer reconstruction of the real world crashes shown in (Tables 6.4, 6.6 & 6.8). This is occupant model is the same model discussed in detail Chapter 4 and used in in the comparisons between volunteer and cadaver lateral impact and the models output.

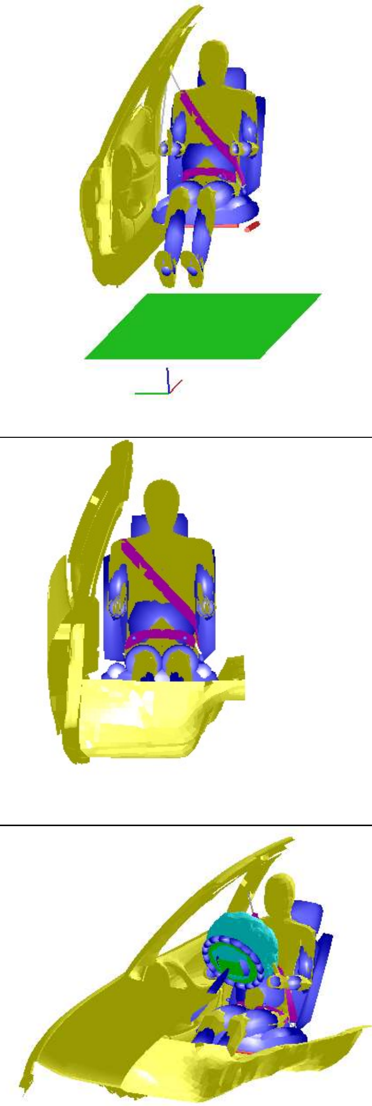
The second model was a vehicle model of a VT Commodore provided by General Motors Holden. The initial MADYMO vehicle model included an ellipsoid seat, facet surface vehicle body and interior. The model was modified to include a finite element lap and sash belts, head restraint (an ellipsoid with squared edges), steering wheel (planes and ellipses) and, in the oblique cases, an airbag (finite element). A head restraint and seat side wings were not included as a part of the seat in the original model and were added for the purpose of this piece of research. The geometry of the head restraint and the side wings were measured from a Holden Commodore vehicle. These were modelled using ellipsoids with squared edges. The head restraint was angled to match the angle of the seatback and was positioned to be at the maximum height of the commodore head restraint, to represent optimal position of the head restraint (Figure 6.5). To reduce the model's run time, all non-essential parts of the vehicle exterior and interior were removed.



Original Seat Model   Modified Seat Model

Figure 6.5 Modifications made to seat model

The near side model for 90° impact (Figure 6.6a) is the simplest of the crash models consisting of seat, door and occupant with lap/sash belt. The interior and exterior structures that did not come into contact with the occupant were removed to reduce the calculation time of the model. The far side model for 90° impact (Figure 6.6b) included an additional faceted console. The near side oblique (Figure 6.6c) is the most detailed with an additional facet console, bonnet and dashboard. The dashboard provided the geometry to position the steering wheel and airbag.



a.

b.

c.

Figure 6.6a: Near-side model- 90° impact
Model includes combined human model, seat with side wing and head restraint, lap/sash belt and vehicle door.
Figure 6.6b: Far-side model- 90° impact
Model includes combined human body model with, seat with side wing and head restraint, lap/sash belt, vehicle door, floor and console.
Figure 6.6c: Near-side oblique- 45° impact

Model includes combined human body model, seat with side wing and head restraint, lap/sash belt, vehicle door, floor, console, dashboard attached to bonnet, steering wheel and airbag.

The steering wheel, airbag and lap sash belts were existing models in the MADYMO 5.4.1 library of applications/models. The lap-sash belt in the vehicle was fitted to the combined human model using a pre-simulation. Both the lap and sash belt consists of a finite element belt section, with MADYMO belt elements at either end. The attachment points for the belt elements were not changed from the original application model. The pre-tensioner from the existing application model was included in all of the side impact reconstructions.

The occupant model was positioned in the seat with a pre-simulation that involved dropping the body into the seat until it settled, this point was taken as the occupant's seating position. The muscle activation level of the neck muscles was the same in the crash simulation models at all impact angles. The left and right flexors, extensors and sternocleidomastoid muscles were activated at 100% with the delay period of 50 ms as tested by De Jager et al. (1996). This activation level used was the same as that used in the lateral and oblique comparisons in Chapter 4 which were found to provide comparable results to the human volunteers. The occupant's arms were positioned to be 90° at the elbow and 90° at the shoulder to compare to a driver holding a steering wheel in a neutral position.

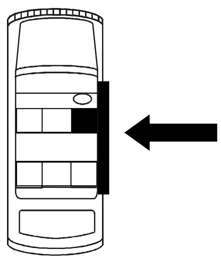
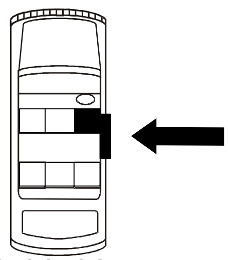
6.3 90° Near-side Impact Results

The case information for the 90° nearside impact case is presented in Table 6.4. The table contains two sets of case data. The first set of data is for the LSNI case and the second set is the data for the control case with no LSNI. The occupants in the lateral near side cases were closely matched for age, height and weight. The driver of the LSNI case was 28 years old, with a height of 175 cm and a weight of 75 kg. The driver of the control case was 24 years old with a height of 176 cm and a weight of 76 kg. The CDC (collision deformation characteristic– described in Appendix 2) was calculated by a crash investigator using the vehicle crush measurements. The calculated CDC for the LSNI case was 03rpew03 and for the control case was 03rpen03. Each vehicle had a driver airbag fitted in the steering wheel which did not deploy. The driver in the LSNI case received injuries to a maximum of AIS 2 with a bruised right kidney. The other AIS 1 injuries received were a bruised right chest and a laceration to the right side of the head. The driver of the control vehicle received no injuries.

The impacting vehicle also varied between the two cases. The mass of the impacting vehicle in the LSNI case was larger than the case vehicle as it was a Mitsubishi Van with the reported mass of 2205kg (Identicar, 2001). This mass is 738kg greater than that of the case vehicle. The impacting vehicle in the control case was a 1985 Toyota Corona with a lighter mass (272kg) than the case vehicle is at 1075kg. This variation is consistent with the findings of von Koch (1995) who report that as difference in mass between the striking and struck (case) vehicle increases, there is an increase in the risk of the occupant receiving a LSNI.

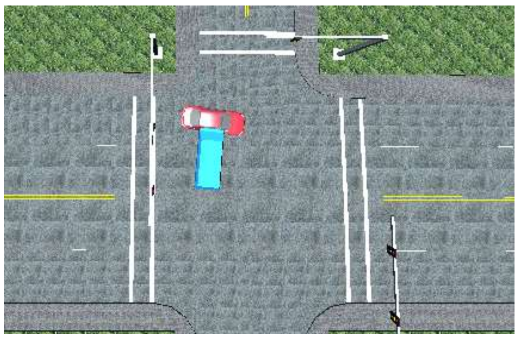
Table 6.4: Case data 90° near-side impact

<table>
<tr><th colspan="5">90° near-side impact</th></tr>
<tr><th></th><th colspan="2">Low Severity Neck Injury Case</th><th colspan="2">Control Case</th></tr>
<tr><td></td><td colspan="2"></td><td colspan="2"></td></tr>
<tr><td>Personal Details</td><td colspan="2">Male
28 year old
175 cm, 75 kg
Seat belt worn
Driver
Airbag Not Fitted</td><td colspan="2">Male
24 year old
176 cm, 76 kg
Seat belt worn
Driver
Airbag Not Fitted</td></tr>
<tr><td>Vehicle Make and Model

Direction of Impact</td><td colspan="2">1996 Holden Commodore
VS Executive Wagon

3 o'clock impact</td><td colspan="2">1995 Holden Commodore
VS Sedan

3 o'clock impact</td></tr>
<tr><td>Calculated CDC</td><td colspan="2">03rpew03</td><td colspan="2">03rpen03</td></tr>
<tr><td>Delta- V</td><td colspan="2">Unknown</td><td colspan="2">12kph</td></tr>
<tr><td>EBS</td><td colspan="2">27kph</td><td colspan="2">19kph</td></tr>
<tr><td>Injuries</td><td colspan="2">AIS 1- Neck Strain
AIS 2- Bruised Right Kidney
AIS 1- Bruised Right Chest
AIS 1- Laceration Right Side Head</td><td colspan="2">Not Injured</td></tr>
<tr><td>Description of Crash</td><td colspan="2">Turning left on green light, vehicle B has run a red light and has been hit on the driver's side</td><td colspan="2">Turning right at a roundabout, failed to see vehicle B, was hit on driver's side as moving into the roundabout.</td></tr>
<tr><td>Impacting vehicle</td><td colspan="2">Mitsubishi Van</td><td colspan="2">1985 Toyota Corona</td></tr>
<tr><td>Vehicle Mass</td><td>Vehicle A
1467kg</td><td>Vehicle B
Unknown
Reported in specs as 2205kg</td><td>Vehicle A
1347kg</td><td>Vehicle B
1075kg</td></tr>
</table>

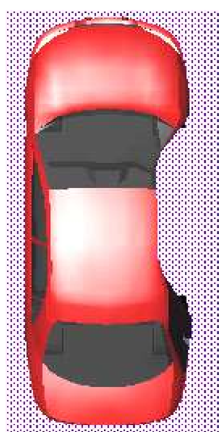
The results from the reconstructions undertaken in HVE are presented in the following section. The images in Figure 6.7 illustrate the point of impact between the two vehicles for each case. The image a shows the Mitsubishi Van (blue), impacting the case vehicle Holden Commodore (red). Image b shows the Toyota Corona (blue) impacting the case vehicle Holden Commodore (red).

Figure 6.7: Point of impact (HVE), 90° near-side impact

90° near-side LSNI	90° near-side control
 a.	 b.

The amount of vehicle crush predicted by HVE for the low severity neck and the control case are presented in Figure 6.8. The CDC predicted by the HVE software is also presented in this figure. The calculated and predicted CDC for the 90° control case was calculated to be the same as 03rpew03. There is a slight difference in the calculated CDC and the predicted CDC for the LSNI case with 03rpen03 as the calculated value and 03rpew02 as the HVE predicted value.

Figure 6.8: Vehicle crush 90° near-side impact

a. LSNI	b. Control
	
03rpew03	03rpew02

The vehicle lateral and frontal accelerations for the 90° near-side cases calculated using HVE were presented in Figure 6.2. These accelerations were used in the MADYMO reconstructions of the near side crashes. The results of the MADYMO crash reconstructions are presented in the following section.

6.3.1 90° Near-side Impact Kinematic Data

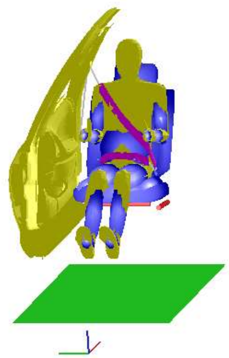
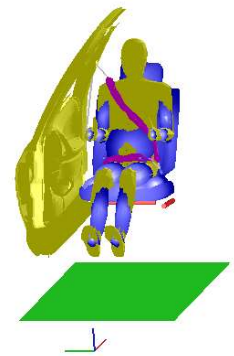
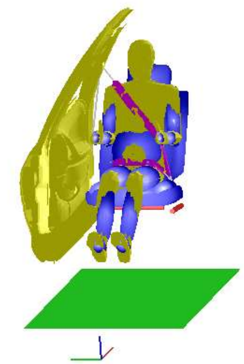
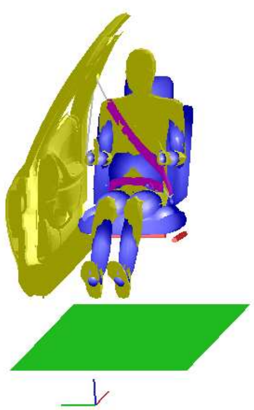
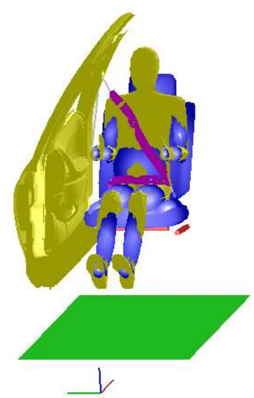
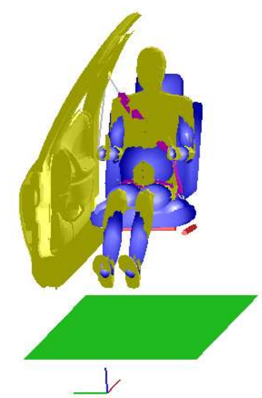
The kinematic images for the 90° near side impact are illustrated in Table 6.5. The images are presented in two rows. The first row contains the low severity neck injury data, the second row contains the data for the control case. Images are displayed at 20 ms intervals. In the LSNI case, the body begins to laterally translate between 60 ms and 80 ms. At 120 ms the lap belt begins to restrain the pelvis while the torso continues to laterally translate. The control case differs with these events occurring later. For the first 100 ms, there is no lateral movement of the body. From 120 ms to 200 ms lateral translation of the torso and head occurs.

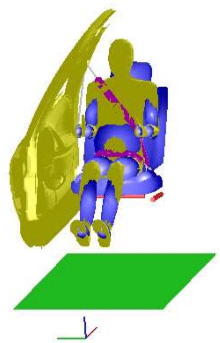
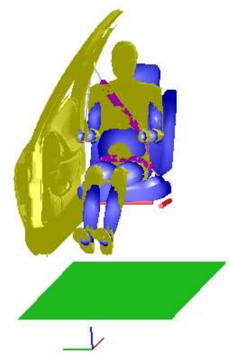
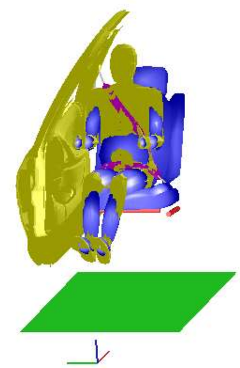
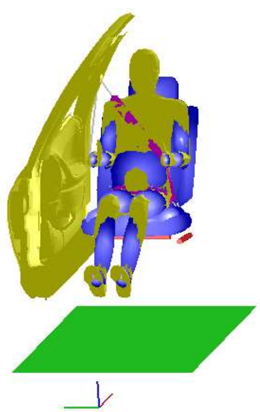
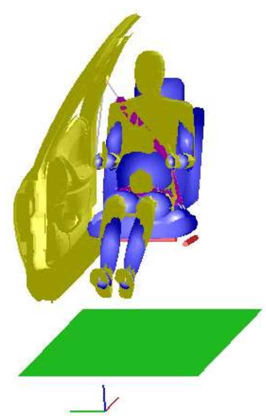
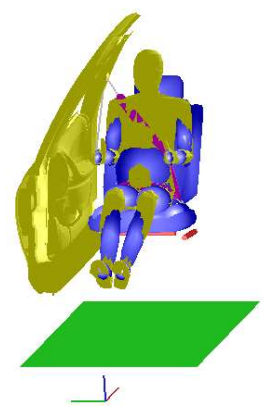
In the LSNI case, as the lap belt restrains the pelvis and the torso continues to translate (120ms), the sash belt contacts the neck and remains in contact until the end. This contact is followed the lateral bending of the head. This differs for the control case where following the continued lateral bending of the torso (200 ms) there is arm contact with the door. It is at 240 ms that the sash belt contacts the neck. It is not until the sash belt has contacted the neck in the LSNI case at 160 ms, that the shoulder contacting the door and at 180 ms the pelvis contacts the door. At 200 ms, the head begins to rotate about the z axis, this continues until 260 ms. In the control case, the torso and pelvis do not contact the door, this is an indicator that the control trial is a less severe trial than the LSNI trial. In this trial, there is minimal head rotation in the z axis. Additional views of the kinematic data have been provided in Appendix 5. These views show the motion of the human model with the facet skin layer removed so that the skull and vertebrae are visible, images have been provided for both trials.

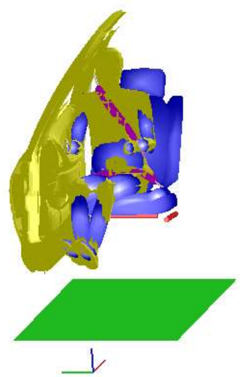
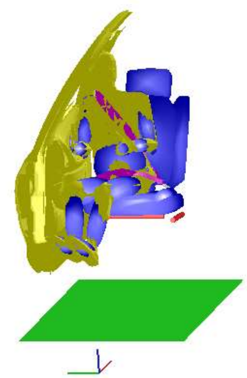
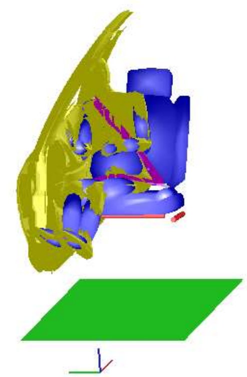
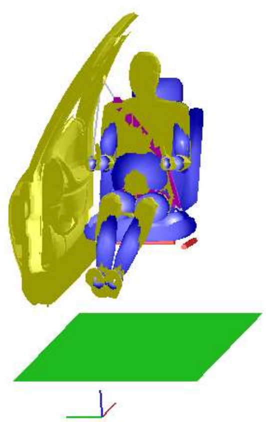
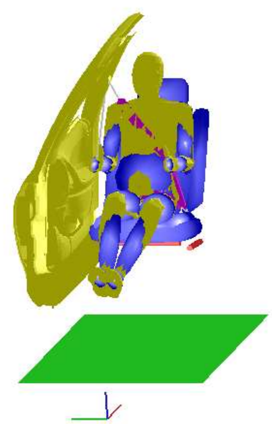
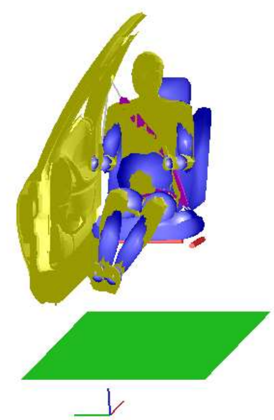
The key finding from the kinematic data is the interaction between the seat belts and occupants in both near side trials. While the lap belt provides some support anchoring the pelvis, the torso moves laterally resulting in contact between the sash belt and the neck. This has been reported in previous literature by Horsch et al. (1979) and Kallieris et al. (1991). As the sash belt contacted the neck in both the LSNI and control case, further investigation is required to determine its influence on injury. It is also important to note that the head restraint provided no protection against neck injury in 90° (lateral) near-side impacts, as there was no head contact with the head restraint in either trial.

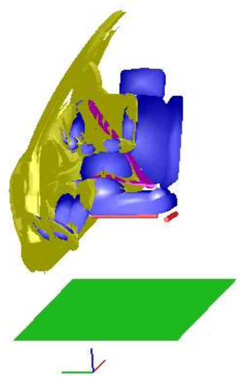
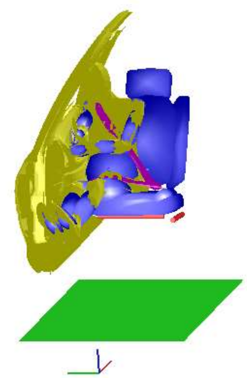
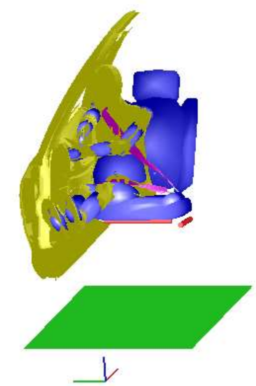
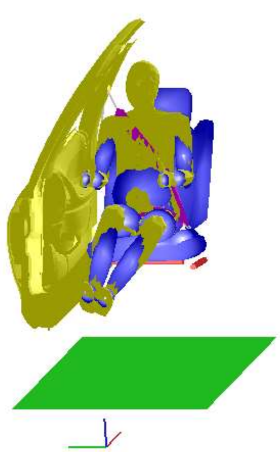
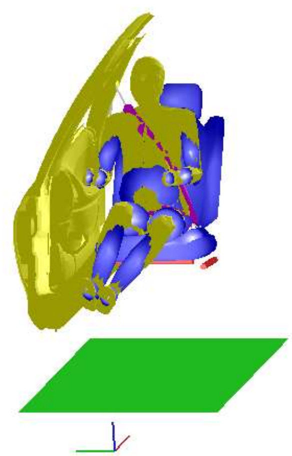
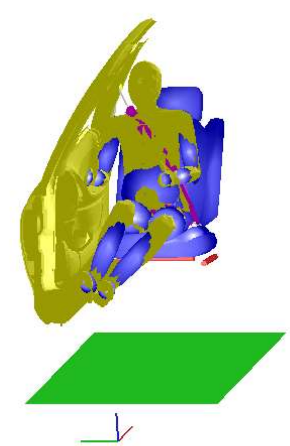
Numerically there is very little variation in the CDC for either case, LSNI case with 03rpew03 compared to the control case 03rpen03. Putting this difference into real-world terms may explain why during the LSNI case that the neck was observed contacting the sash belt earlier in the trial. The 'w' in the CDC represents the wide damage area compared to the ‘n’ which represents a narrow damage area. This difference can be attributed to the geometry and the mass of the impacting vehicle. The impacting object in the LSNI case is a Mitsubishi van with a mass reported in the specs as 2205kg (Identicar 2001). This compares to the impacting vehicle in the control case that is an older 1985 Toyota Corona with a vehicle mass of 1075 kg.

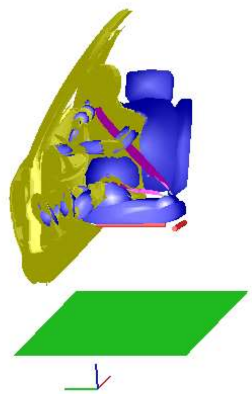
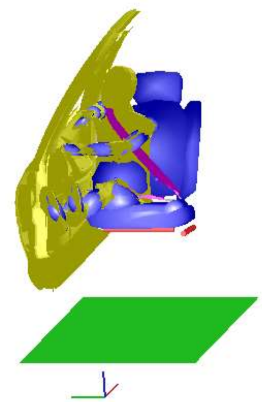
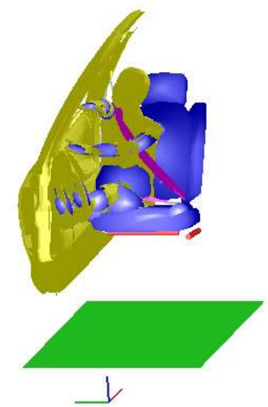
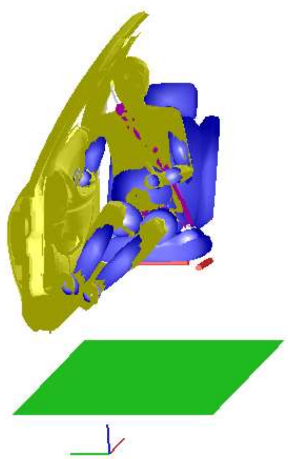
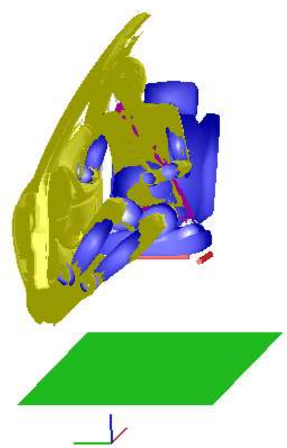
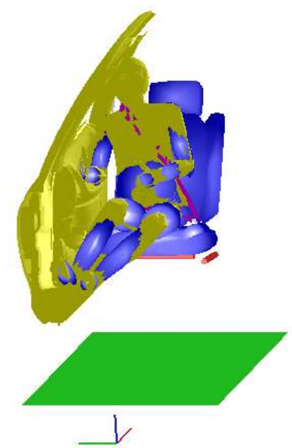
Table 6.5: Kinematic images (MADYMO 5.4.1), 90° near-side impact

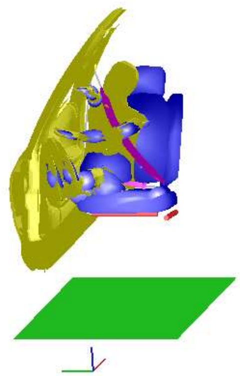
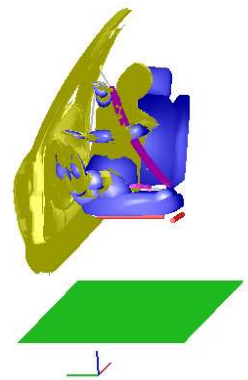
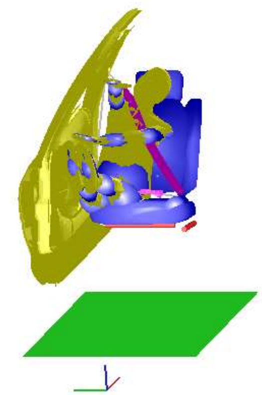
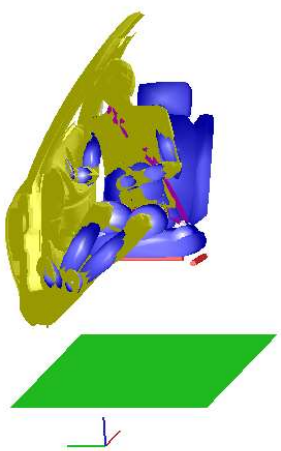
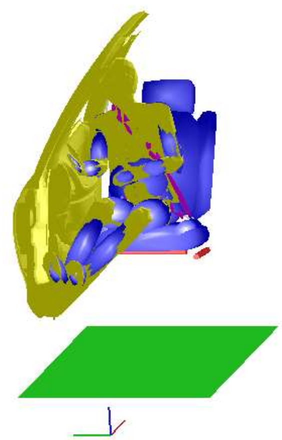
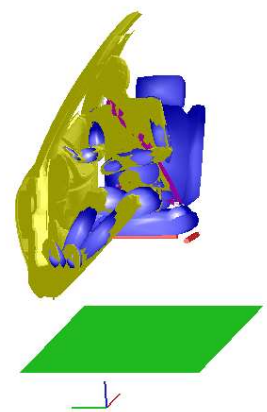
	Time:00 ms	Time:20 ms	Time:40 ms
Low Severity Neck Injury Case			
Control			

	Time:60 ms	Time:80 ms	Time:100 ms
Low Severity Neck Injury Case			
Control			

	Time:120 ms	Time:140 ms	Time:160 ms
Low Severity Neck Injury Case			
Control			

	Time:180 ms	Time:200 ms	Time:220 ms
Low Severity Neck Injury Case			
Control			

	Time:240 ms	Time:260 ms	Time:280 ms
Low Severity Neck Injury Case			
Control			

	Time:300 ms	Time:320 ms	Time:340 ms
Low Severity Neck Injury Case			
Control			

6.3.2 90° Near-side Acceleration

The lateral accelerations for the 90° near-side case show that the sash belt contact with the neck has the greatest influence on the lateral acceleration of the head and first thoracic vertebra. The lateral acceleration of first thoracic vertebra for the 90° nearside control case (blue line, Figure 6.9), remains close to 0 until 218 ms and has an initial peak of 59 m/s^2 then returns around 0. The acceleration remains at 0 until reaching a peak of 39 m/s^2 at 264 ms. Following this event, the acceleration again returns to 0 until the end. For the LSNI case (red line, Figure 6.9), there is constant rate of deceleration to 110ms at a peak of 56 m/s^2. From this point, the data becomes very noisy. When comparing the acceleration data to the kinematic data in the previous section it can be seen that this occurs due to contact between the sash belt and the neck. The peak lateral acceleration observed following the initial contact is 290 m/s^2 at 153 ms.

The head acceleration in the y-direction, for the 90° near-side case (Figure 6.10) has been presented in Figure 6.10. The lateral head acceleration in the control case (blue line) remains constant until 264 ms where there is rapid deceleration to -50 m/s^2 followed by an increase in acceleration to 288 m/s^2 at 268 ms. This rapid acceleration occurs at the time the sash belt is, again, in contact with the neck. It is important to note that the accuracy of the peak values may not be relied on, due to the neck contact with the sash belt not yet being validated. The lateral head acceleration for the LSNI case (red) shows that the head decelerates slowly reaching a peak of -48 m/s^2 at 117 ms. At this point, the head deceleration slows reaching 0 m/s^2 at 129ms. This is followed by head acceleration peaking at 48 m/s^2 at 137 ms at this point the acceleration slows until 142 ms. Following this the acceleration increases reaching a maximum peak of 128 m/s^2 at 190 ms, the acceleration returns to 0 at 224 ms.

The variation in acceleration data can be attributed to the difference in the severity of impact for the two cases, resulting from the variation in the impacting vehicle masses. The case data previously presented in Table 6.4 shows that with the LSNI case there is a greater variation between vehicle mass for the impacting vehicle and the case vehicle (738kg). For the control case the impacting vehicle's mass was less than the case vehicle, with a difference of (272kg). This difference suggests that the LSNI case, experienced a more severe impact than the control case. This is also supported by the key events such as sash belt contact with neck occurring earlier in the LSNI case and the predicted depth of crush on the vehicle visible in Table 6.5.

The head and neck kinematics of the combined human model observed in the near-side 90° impact trials, are consistent with the findings reported in literature. Both Kallieris et al.

(1991) and Wismans et al. (1983) report lateral head displacement followed by head rotation in the z-axis in both volunteer and cadaver tests. The greater head rotation observed in the LSNI case compared to the control case suggests that it plays a role in the mechanism of injury. The occurrence of the sash belt contacting the neck during the near side impacts has been widely reported in the literature. Kallieris et al. (1991) and Horsch et al. (1979) report neck loading from the sash belt contacting the neck. Horsch et al. (1979) reported the loading of the neck depended upon belt anchor height, anchor for-aft position, subjects seating position and height. The influence of these factors on the neck loading from the belt against the neck were not investigated in these reconstructions.

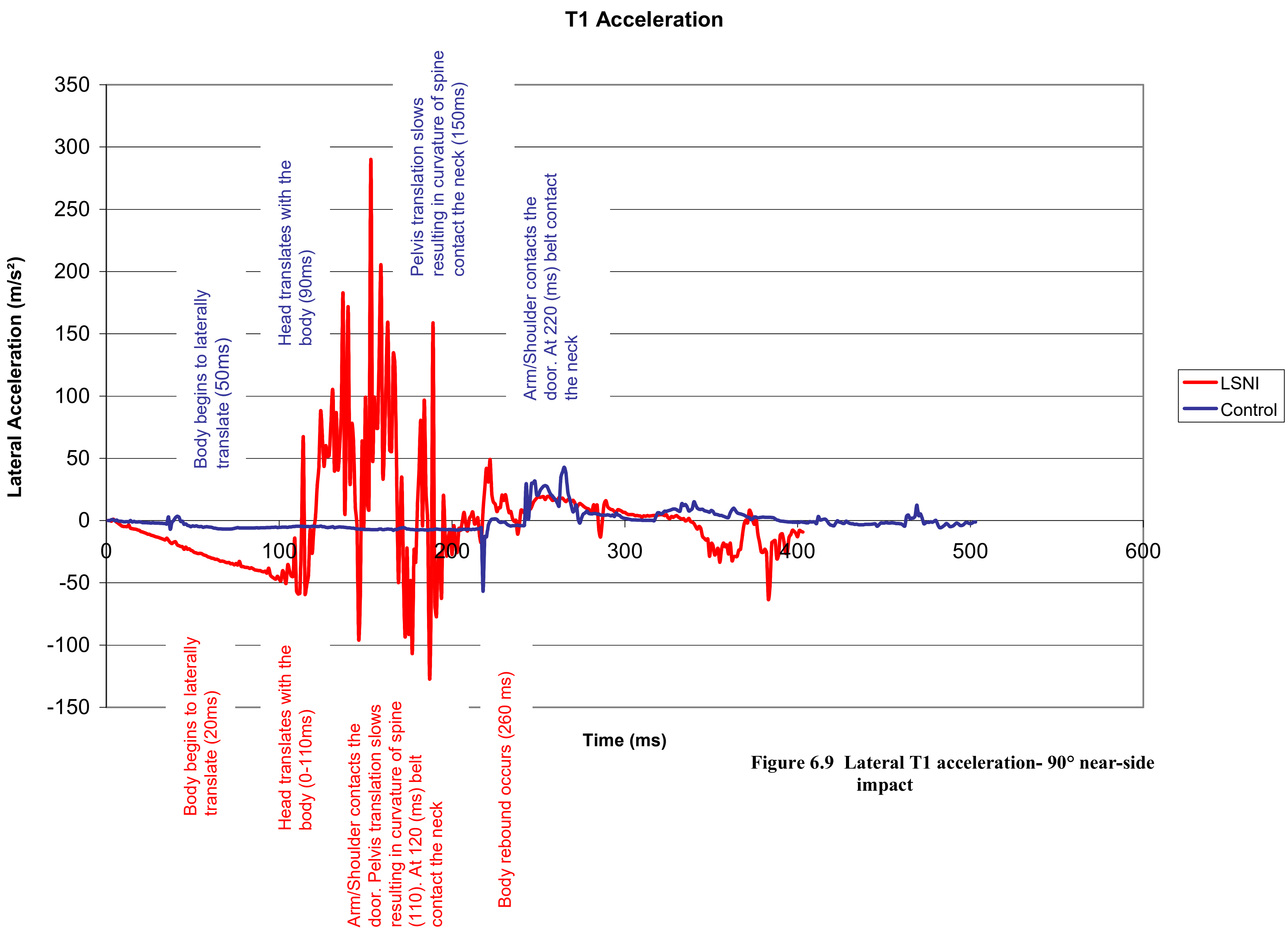


Figure 6.9 Lateral T1 acceleration- 90° near-side impact

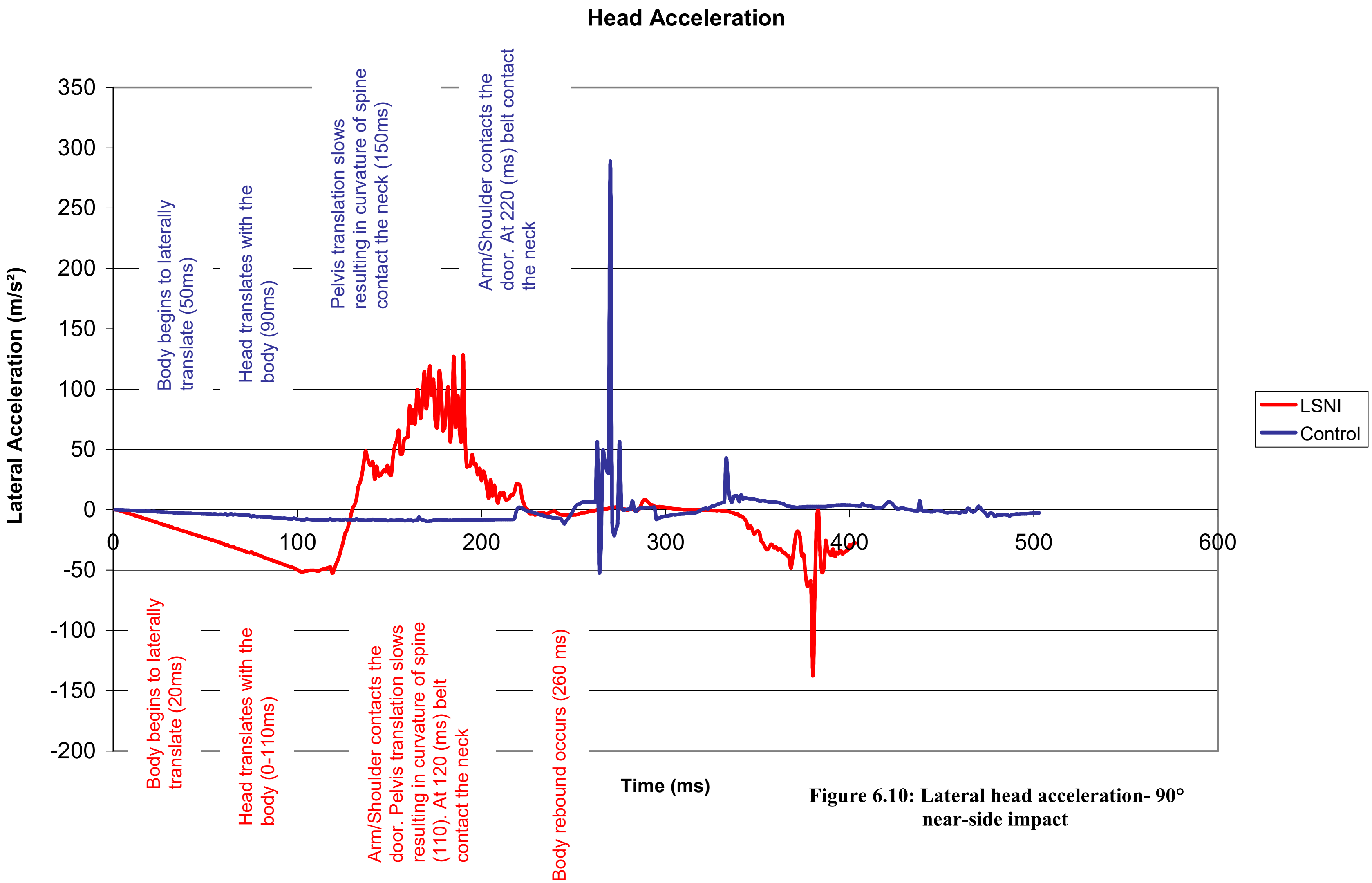


Figure 6.10: Lateral head acceleration- 90° near-side impact

6.4 90° Far-side Results

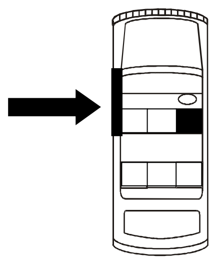
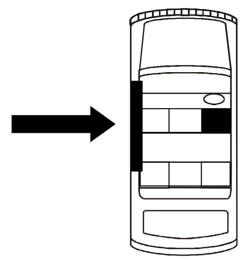
The case information for the 90° far-side impact cases is presented in Table 6.6. The table contains two sets for case data. The first set of data is for the LSNI case and the second set is the data for the control case with no LSNI. The driver of the LSNI case was 23 years old, with a height of 185 cm and a weight of 88 kg. The driver of the control case was 58 years old with a height of 183 cm and a weight of 102 kg. While age has been identified in literature (Morris et al. (1996), Teaming et al. (1998) as an influencing factor for LSNI the difference in the occupants age cannot be considered when interpreting the model's results. This is because the human model does not have the scope to introduce the anatomical effects of ageing.

The case vehicle had a driver airbag fitted in the steering wheel which did not deploy in the 90° impact, the control case had no airbag fitted. The driver in the LSNI case received injuries to a maximum of AIS 1 with lacerations to the left side of the scalp, contusions to the left hip, contusions to the right shoulder, contusions to the left shoulder and contusion at the left ribs. The driver of the control vehicle received no injuries. The impacting vehicle also varied between the two cases.

The impacting vehicle in the LSNI case was of a greater mass than the case vehicle as it was a 4 wheel drive (SUV) with a bull bar, with the reported mass of 1903kg. The impacting vehicle in the control case was a Suzuki Vitara with a smaller mass than that of the case vehicle at 1039kg. As with the nearside case, the variation in mass between the impacting vehicle and the case needs to be considered when interpreting the results. For the LSNI case the striking vehicle has a 275 kg greater mass of than that of the struck (case) vehicle. In the control case the struck vehicle has a greater mass by 506 kg than the striking vehicle. This variation is comparable to the nearside cases previously presented and is comparable to the findings in the literature by von Koch et al. (1995) as previously discussed.

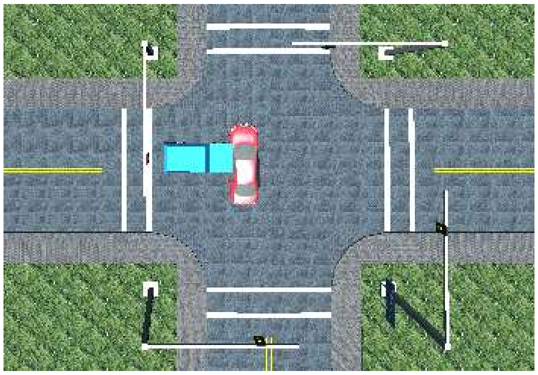
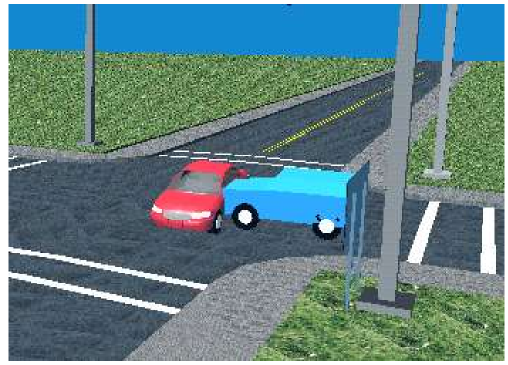
Table 6.6: Case data- 90° far-side impact

<table>
<tr><th colspan="5">90° far-side impact</th></tr>
<tr><th></th><th colspan="2">Low Severity Neck Injury Case</th><th colspan="2">Control Case</th></tr>
<tr><td></td><td colspan="2"></td><td colspan="2"></td></tr>
<tr><td>Personal Details</td><td colspan="2">Male
23 year old
185cm, 88 kg
Seat belt worn
Driver
Airbag Fitted, Not Deployed</td><td colspan="2">Male
58 year old
183cm, 102 kg
Seat belt worn
Driver
No Airbag Fitted</td></tr>
<tr><td>Vehicle Make and Model

Direction of Impact</td><td colspan="2">1999 Holden Commodore
Acclaim Series II VT Sedan

9 o'clock impact</td><td colspan="2">2001 Holden Commodore
VX Executive Sedan

9 o'clock impact</td></tr>
<tr><td>Calculated CDC</td><td colspan="2">09lyew03</td><td colspan="2">09lpmw02</td></tr>
<tr><td>Delta- V</td><td colspan="2">Unknown</td><td colspan="2">Unknown</td></tr>
<tr><td>EBS</td><td colspan="2">25.4 kmh</td><td colspan="2">17.2 kmh</td></tr>
<tr><td>Injuries</td><td colspan="2">AIS 1- Neck Strain
AIS 1 - Lacerations Left Scalp
AIS 1- Contusion Left Hip
AIS 1- Contusion Right Shoulder
AIS 1- Contusion Left Shoulder
AIS 1- Contusion Left Ribs</td><td colspan="2">No Injuries</td></tr>
<tr><td>Description of Crash</td><td colspan="2">Crossing Intersection, hit on side by 4WD, pushed into gutter- 180 degree turn</td><td colspan="2">Travelling through intersection
and hit by other vehicle</td></tr>
<tr><td>Impacting vehicle</td><td colspan="2">4WD with Bull Bar</td><td colspan="2">Suzuki Vitara</td></tr>
<tr><td>Vehicle Mass</td><td>Vehicle A
1627kg</td><td>Vehicle B
1815+pass+BB (1903)</td><td>Vehicle A
1545kg</td><td>Vehicle B
1039kg</td></tr>
</table>

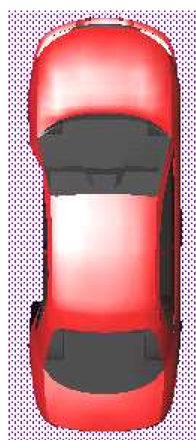
The results from the reconstructions undertaken in HVE are presented in the following section. The images in Figure 6.11 illustrate the point of impact between the two vehicles for each case. The image (a) shows a 4-Wheel Drive (SUV) -brand unknown (blue), impacting the case vehicle Holden Commodore (red). Image (b) shows the Suzuki Vitara (blue) impacting the Case vehicle Holden Commodore (red).

Figure 6.11: Image of point of impact from HVE, 90° far-side impact

90° far-side LSNI	90° far-side control
 a.	 b.

The amount of vehicle crush predicted by the HVE software are presented in Figure 6.12. The CDC for the LSNI case was calculated to be the same as that predicted by the HVE software as 09lyew03. There is a slight difference in the calculated and predicted CDC for the control case. The calculated value was 09lpmw02 and the value predicted by HVE was 09lpew03.

Figure 6.12: Vehicle crush- 90° far-side impact

a. LSNI	b. Control
	
09lyew03	09lpew03

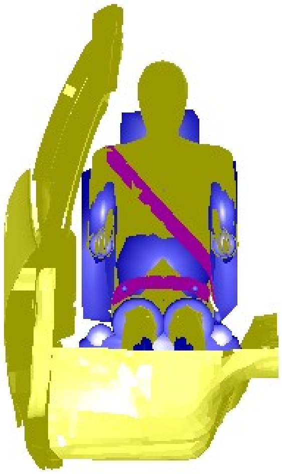
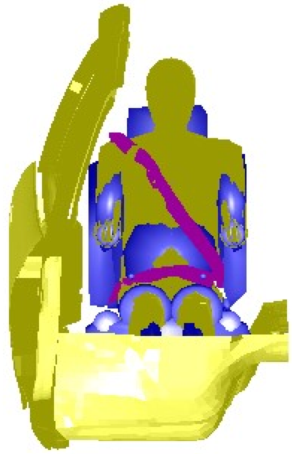
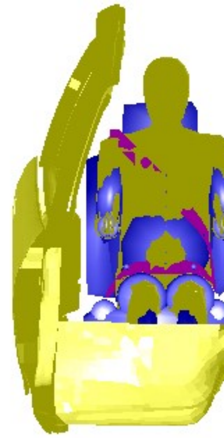
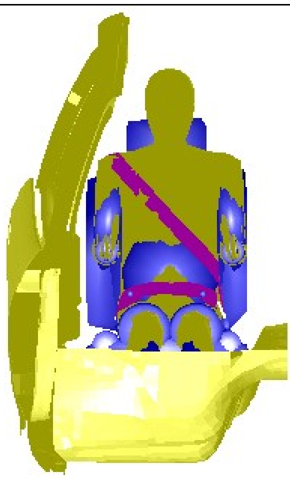
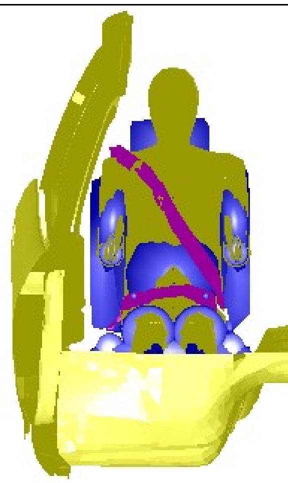
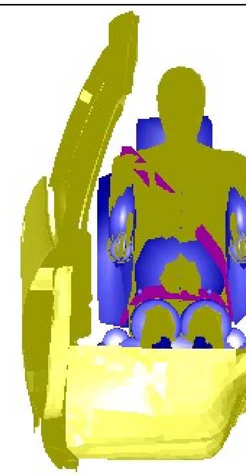
The vehicle lateral and frontal accelerations for the 90° far-side cases calculated using HVE were presented in Figure 6.3. These accelerations were used in the MADYMO reconstructions of the near side crashes. The results of the MADYMO crash reconstructions are presented in the following section.

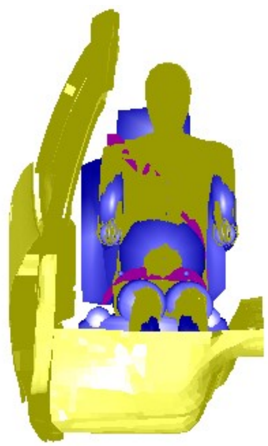
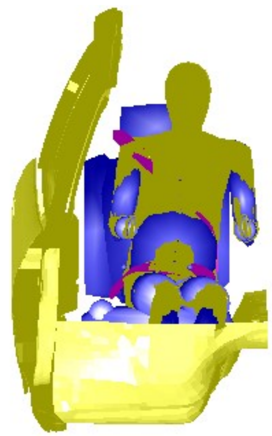
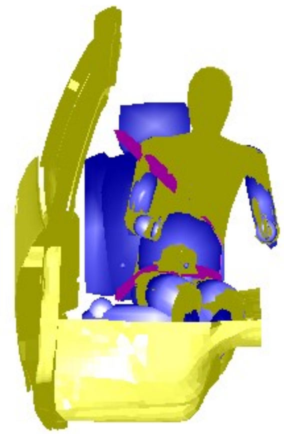
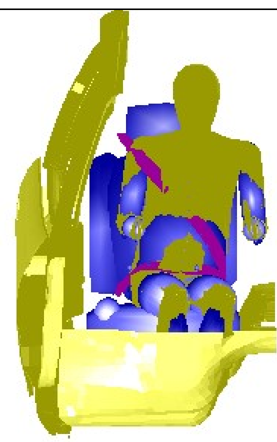
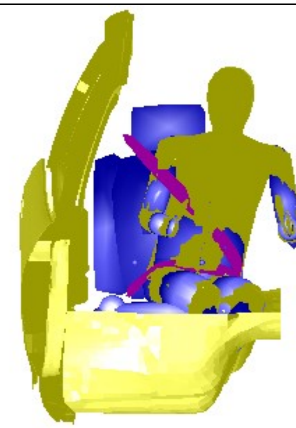
6.4.1 90° Far-side Kinematic Data

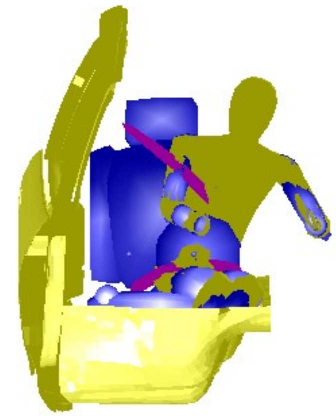
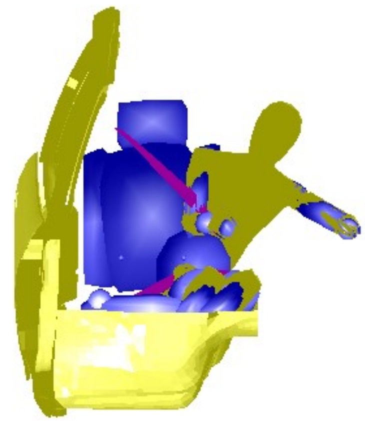
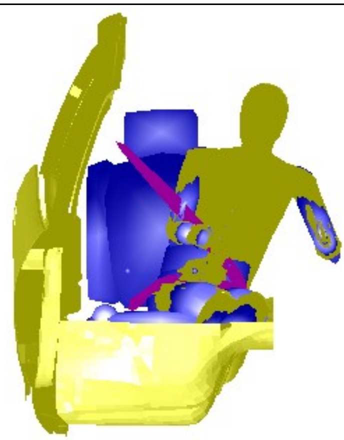
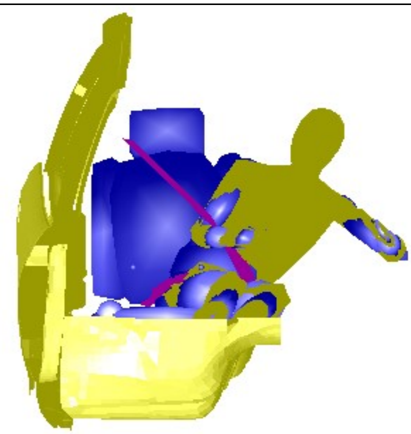
The kinematic images for the 90° far side impact are presented in Table 6.7. The data are presented in the same format as in Table 5.5. The images show that the LSNI case and the control case are very similar. For the LSNI case, lateral translation away from the door occurs at 40 ms. The motion of the driver in the control case is similar as occupant lateral translation away from the door begins at 40 ms. In the LSNI case at 80 ms the lap belt begins to restrain the pelvis, and the torso continues to laterally translate, as the sash belt is not supporting the upper body at this impact angle. The head also translates with the torso. Therefore, there is no lateral bending or head rotation. This occurs until the end of the trial with the torso almost vacating the sash belt completely. The occupant motion in the control is similar. It is important to note that in both trials, due to the crash configuration, that there is notable forward momentum of both the case and control vehicles at the time of the lateral 90° impact. The outcome of this is that any lateral bending would be coupled with head and neck flexion. This coupled motion could contribute to the mechanism of injury.

In both trials the console anchors the thighs and legs. During both the far-side LSNI case and the control case there is no contact between the sash belt and the neck. The sash belt has very little contact with the upper torso, as the human model shows the occupant laterally slipping out of the belt. This is supported by the findings of Faerber (1982) and Horsch et al. (1979) who report that the three-point seat belt provides little protection to the occupant in far side lateral impact. Additional views of the kinematic data have been provided in Appendix 5. These views show the motion of the human model with the facet skin layer removed so that the skull and vertebrae are visible, images have been provided for both trials.

Table 6.7: Kinematic images from MADYMO 5.4.1, 90° near-side impact

	Time:00 ms	Time:20 ms	Time:40
Low Severity Neck Injury Case			
Control			

	Time:60 ms	Time:80 ms	Time:100 ms
Low Severity Neck Injury Case			
Control			

	Time:120 ms	Time:140 ms
Low Severity Neck Injury Case		
Control		

6.4.2 90° Far-side Acceleration Data

The lateral first thoracic vertebra acceleration results for the far side cases are presented in Figure 6.13. The LSNI case results are presented in red and the control case results are presented in blue. In the LSNI case the first thoracic vertebra slowly accelerates reaching a peak of 68 ms^2 at 100 ms. The acceleration then slows briefly to 13 m/s^2 at 115 ms, followed by a slight increase in acceleration to 32 m/s^2 at 132 ms. At this point the acceleration slows reaching 0 at 153ms. The T1 acceleration for the control follows a similar pattern with the acceleration slowly increasing to a slightly higher peak of 94 m/s^2 at 100ms. The acceleration then also slows briefly to 50 m/s^2 at 119 ms, followed by a slight increase in acceleration to 63 m/s^2 at 149 ms. At this point the acceleration slows reaching 0 at 149 ms. Due to the configuration of the three-point belt, there is no contact between the sash belt, neck and upper body as in the near-side cases presented in the previous section.

The lateral head acceleration results for the far side cases are presented in Figure 6.14. The results show that the lateral head acceleration is similar between the LSNI (red) and the control case (blue). In both cases there is rapid acceleration that peaks at 99 m/s^2 at 103 ms for the LSNI case and 117 m/s^2 at 100 ms for the control case. This peak acceleration occurs when the pelvis contacts the consol. For both cases, the initial peak is followed by the acceleration slowing passing through 0 at 147 ms for the LSNI case and later at 164 ms for the control case.

In spite of the variation in the mass between the striking vehicle and the case vehicle and the variability in model between the VT and VX Commodore (presented in Table 6.6), there is little difference in the T1 and head acceleration data in both the LSNI case and the control case. From the greater mass of the striking vehicle and the greater predicted crush (Figure 6.12) for LSNI case it would be expected that this case had greater magnitude of impact. This is not reflected in the acceleration or the kinematic data.

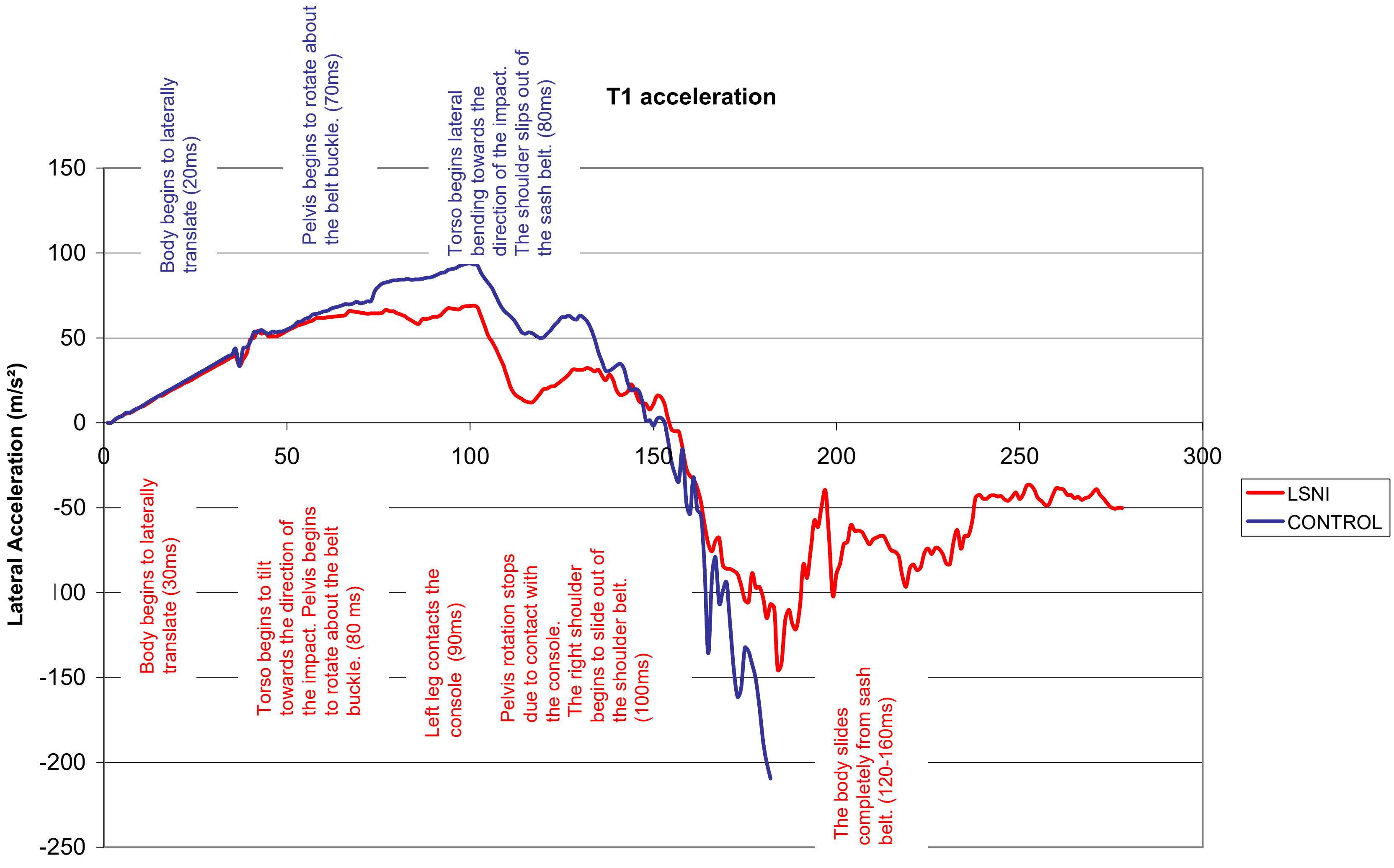


Figure 6.13: Lateral T1 acceleration for the 90° far-side impact

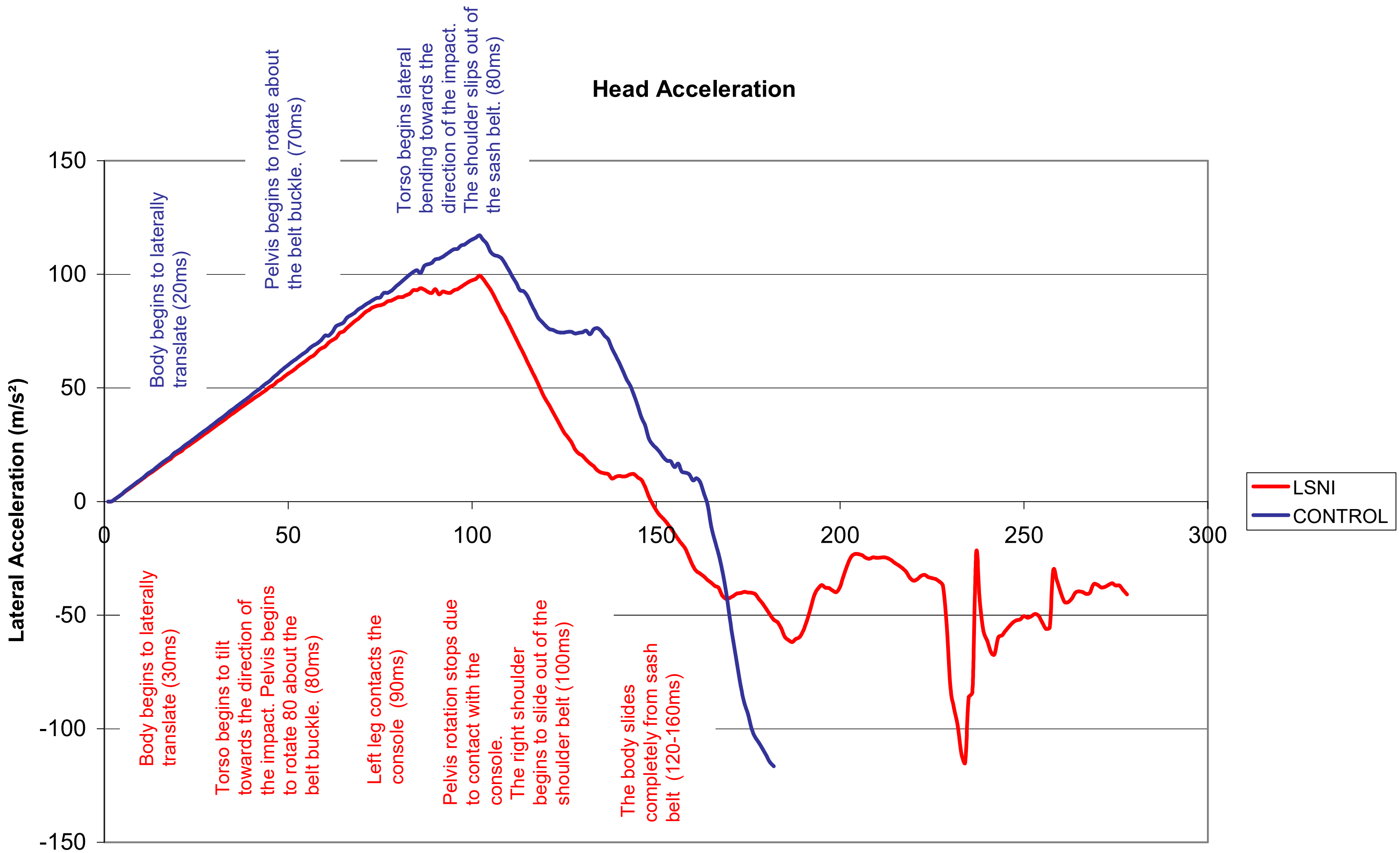


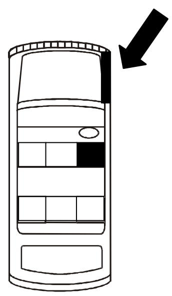
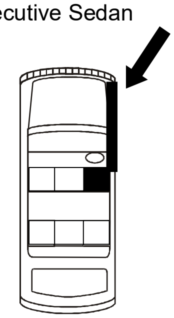
Figure 6.14: Lateral head acceleration for the 90° far-side impact

6.5 Oblique Near-side Results

The case information for the oblique near-side impact cases are presented in Table 6.8. The driver of the LSNI case was 18 years old, with a height of 178 cm and a weight of 70 kg. The driver of the control case was 36 years old with a height of 183 cm and a weight of 82 kg. As with the far-side cases in the previous section, the human model did not allow for the variation in occupant's age to be modelled.

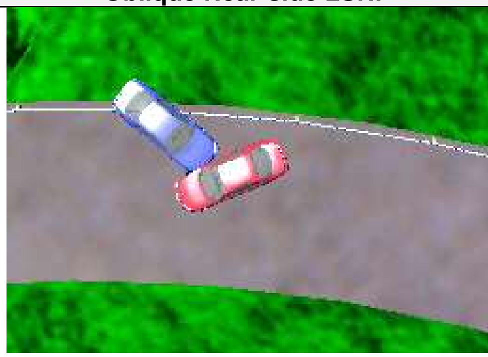
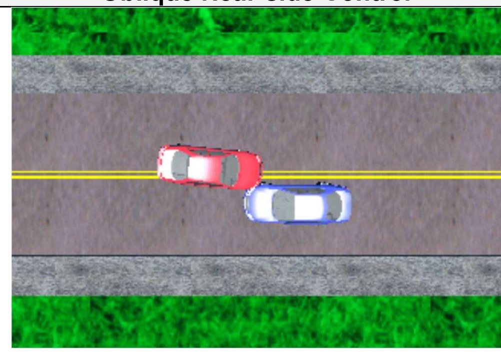
Both vehicles had a driver airbag fitted in the steering wheel that deploy in the impact. The driver in the LSNI case received injuries to a maximum of AIS 1 with an abrasion to the right arm. The driver of the control vehicle received injuries to a maximum of AIS 1 also with a contusion to the sternum, a contusion to the right forearm and a contusion to the left forearm. The impacting vehicle also varied between the two cases. The impacting vehicle in the LSNI case was of a lesser mass (1788 kg) than the case vehicle as it was a 1994 Holden Berlina with the reported mass of 1426 kg. The impacting vehicle in the control case was a Mazda 929 with a reported mass 1619 kg, comparable to the case vehicle at 1586 kg.

Table 6.8: Case data 45° oblique near-side impact

Oblique near-side impact		
	Low Severity Neck Injury Case	**Control Case**
		
Personal Details	Male 18 year old 178cm, 70 kg Seat belt worn Driver Airbag Fitted and Deployed	Male 36 year old 183cm, 82 kg Seat belt worn Driver Airbag Fitted and Deployed
Vehicle Make and Model Direction of Impact	99 Holden Clubsport 4 Door Sedan  **1 o'clock impact**	97 Holden Commodore VT Executive Sedan  **1 o'clock impact**
Calculated CDC	01rfew03	01ryes01
Delta-V	23.6 kmh	Unknown
EBS	25.9 kmh	approx 15.7 km/h
Injuries	AIS 1- Neck Strain AIS 1- Abrasion to right forearm	AIS 1- Contusion Sternum AIS 1- Contusion right forearm AIS 1- Laceration right forearm
Description of Crash	Driving minor highway- hairpin bend, road went in a different direction to what driver thought- driver went over central line, hit on coming car.	Recollection not clear – driving along then airbag deployed. Damage to right side of the car, side swiped another vehicle.
Impacting vehicle	1994 Holden Berlina	Mazda 929
Vehicle Mass	Vehicle A 1788kg | Vehicle B 1426kg	Vehicle A 1586kg | Vehicle B 1619kg

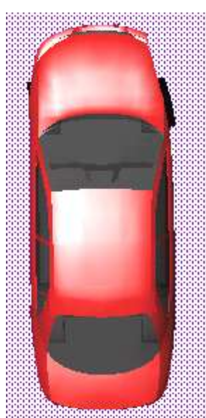
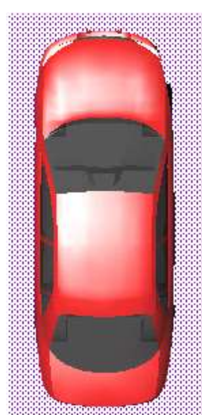
The results from the reconstructions undertaken in HVE are presented in the following section. The images in Figure 6.15 illustrate the point of impact between the two vehicles for each case. The image (a) shows a Holden Berlina (blue), impacting the case vehicle Holden Commodore (red). Image (b) shows the Suzuki Vitara (blue) impacting the Case vehicle Holden Commodore (red).

Figure 6.15: Image of point of impact from HVE, 45° oblique near-side impact

Oblique Near-side LSNI	Oblique Near-side Control
 a.	 b.

The amount of vehicle crush predicted by HVE software for the case and control vehicle are presented in Figure 6.16. The CDC for the oblique nearside control case was calculated to be slightly different at 01ryes01 than what was predicted using the HVE software at 01ryes03. There is a slight difference in the calculated and predicted CDC for the LSNI case with 01rfew03 calculated using Crash 3 and 01ryew03 calculated using HVE. These trials were deemed acceptable as the difference between the CDC for the real-world trial and the HVE simulation did not represent a large variation in the shape and location of the crush. Not being able to directly replicate the CDC of the real-world crashes using HVE, illustrated the difficulties in researching oblique impacts due to the variability of those angled impacts compared to a linear lateral 90° impact.

Figure 6.16: Vehicle crush 45° oblique near-side impact

a. LSNI	b. Control
	
01ryew03	01ryes03

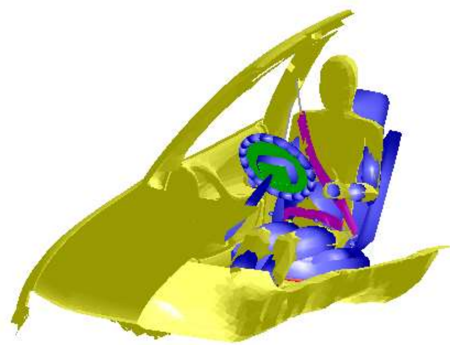
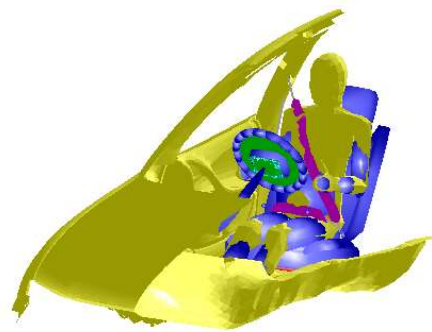
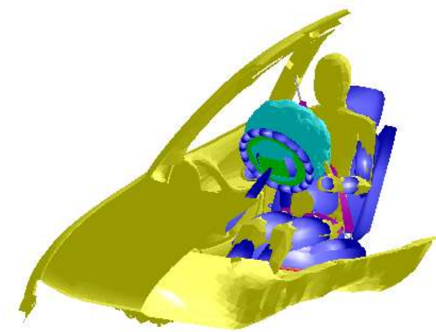
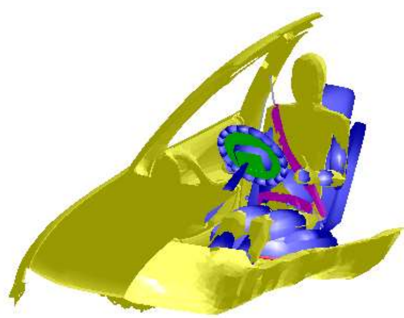
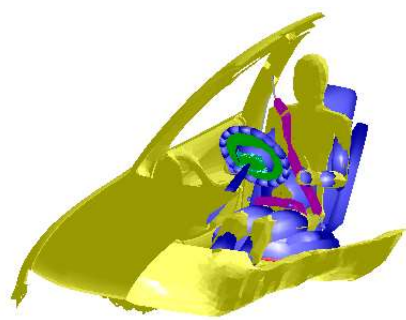
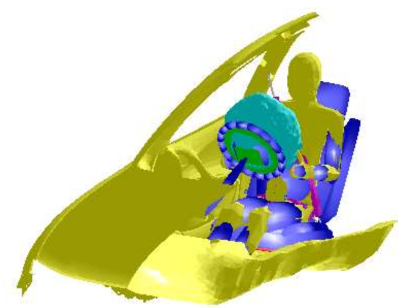
The vehicle lateral and frontal accelerations for the 90° lateral far side cases calculated using HVE were presented in Figure 6.3. These accelerations were used in the MADYMO reconstructions of the near-side crashes. The results of the MADYMO crash reconstructions are presented in the following section.

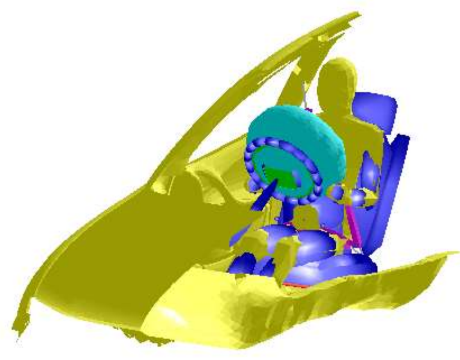
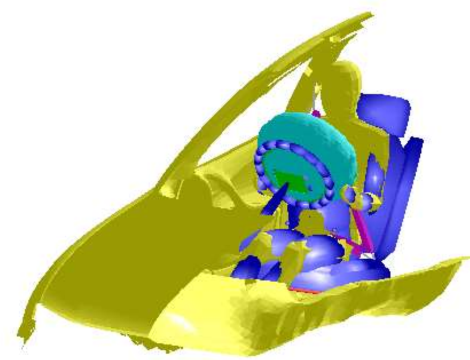
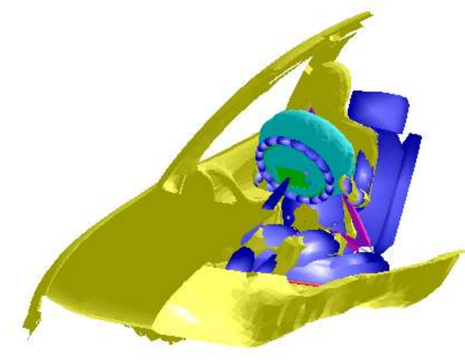
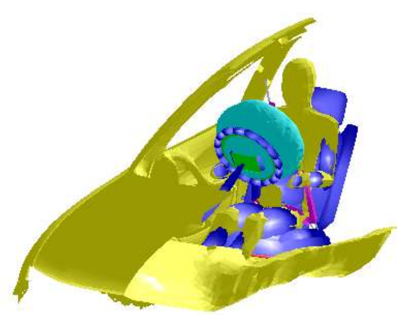
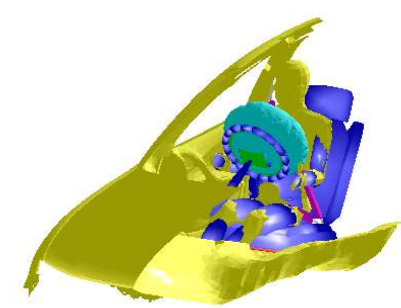
6.5.1 Oblique Near Side Kinematic Data

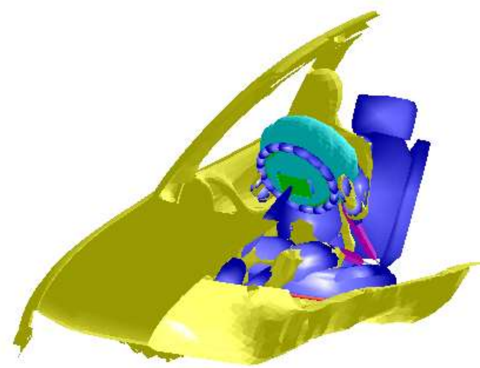
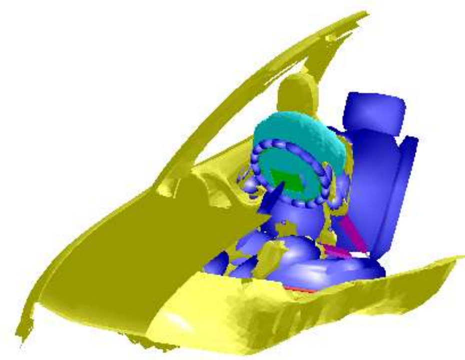
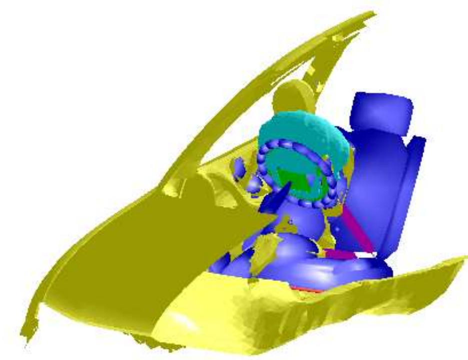
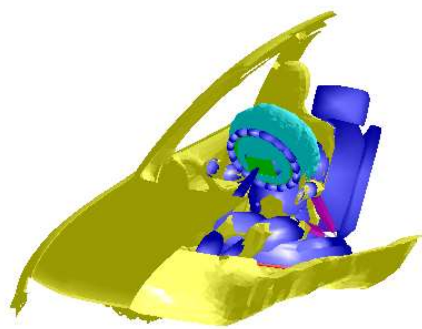
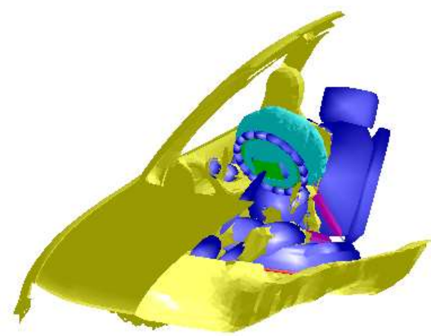
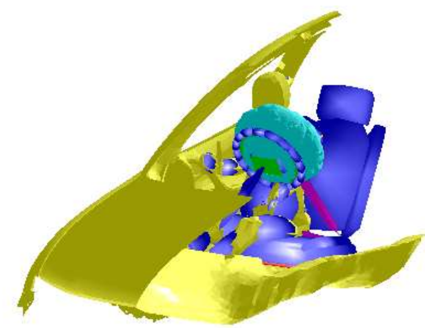
The kinematic images for the 45° oblique near-side impact are presented in Table 6.9. The data is presented in the same format as in Table 6.5. In summary the images show that the LSNI case and the control case are very similar. In both cases, there is no driver body motion until 40ms. The airbag begins to deploy at 40 ms and is fully deployed at 60 ms. The occupant motion follows a similar pattern in both cases. The body begins to translate both laterally and forward. In the LSNI case the arm contacts the door at 130 ms and at 150 ms the body contacts the arm. During this trial there is no head or torso contact with the airbag. In the control, the arm also contacts the door and the body contacts the arm (180 ms).

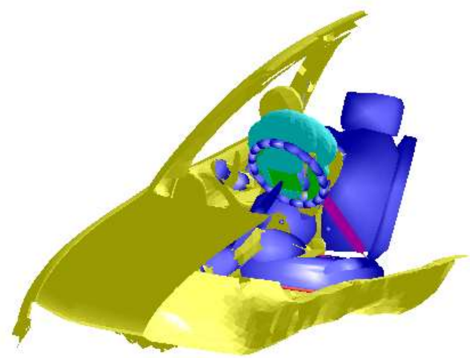
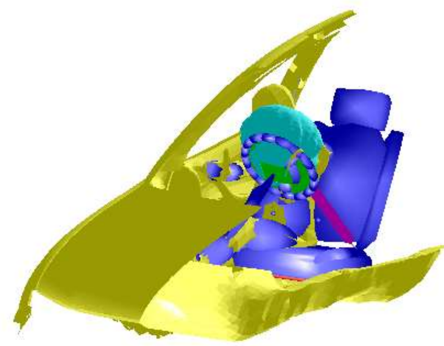
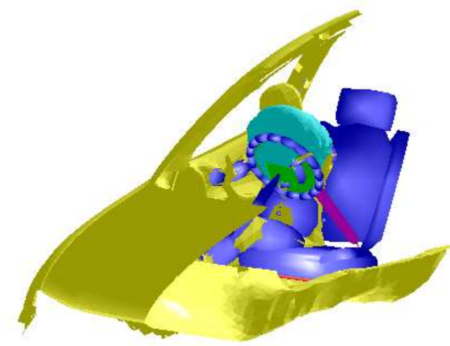
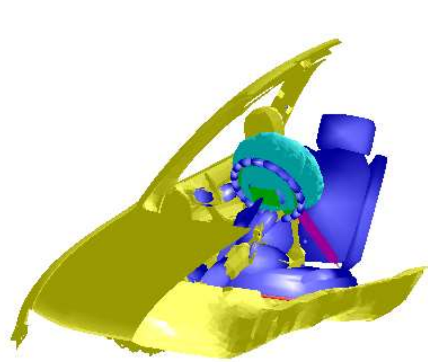
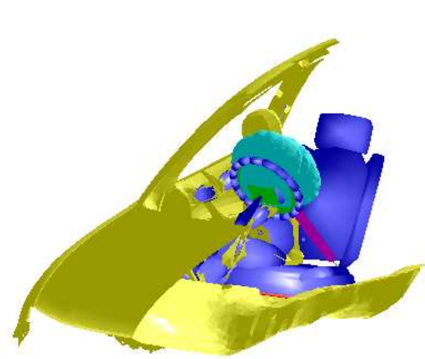
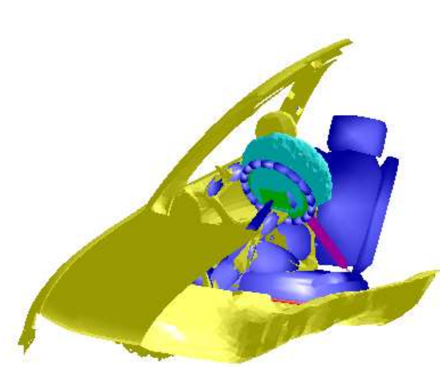
The most interesting point in both of these trials is that the sash belt contacts the neck. For the LSNI case, the contact occurs at 130 ms. For the control case the contact occurs at 140 ms. At the time of neck contact with the sash belt in each case, the head rotates about the contact point. The contact between the belt and the neck is similar to that shown in the near-side reconstructions in Section 6.4. It is also shown that the sash belt contact with the neck, has an influence of the accelerations in both the LSNI and control cases. This is consistent with the findings of Kallieris et al. (1990) that shows that in 90° and 60° near-side lateral impact there is seatbelt-induced loading on the neck. Specifically, Kallieris et al. (1990) found that there was higher peak load in the 60° test than the 90° tests. Horsch et al. (1979) found that neck loading from belt was dependant on belt anchor, anchor height, anchor fore-aft position, subjects seating position, seating posture and height. These variables were not modified in this research, as the same, belt anchor points, seat position, occupant posture and occupant height were used in both the test vehicle and control case across all impact directions. Future testing of human model occupants of different height percentiles across different makes of vehicle is required to identify what factors influence neck loading by the sash belt, and how this can be limited. Additional images of the kinematic data have been provided in Appendix 5. These images show the motion of the human model with the facet skin layer removed so that the skull and vertebrae are visible, images have been provided for both trials.

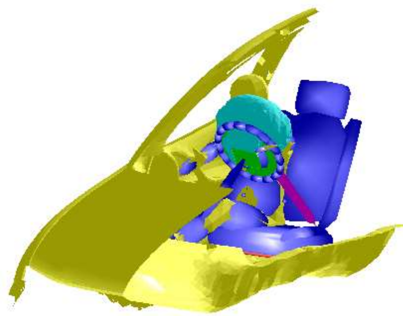
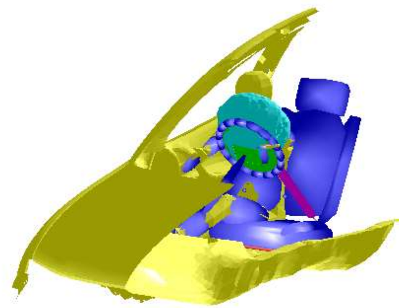
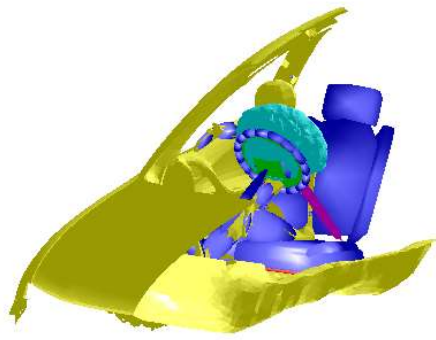
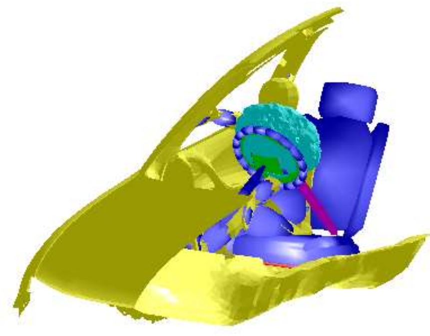
Table 6.9: Kinematic images from MADYMO 5.4.1, oblique near-side impact

	Time:00 ms	Time:20 ms	Time:40 ms
Low Severity Neck Injury Case			
Control			

	Time:60 ms	Time:80 ms	Time:100 ms
Low Severity Neck Injury Case			
Control			

	Time:120 ms	Time:140 ms	Time:160 ms
Low Severity Neck Injury Case			
Control			

	Time:180 ms	Time:200 ms	Time:220 ms
Low Severity Neck Injury Case			
Control			

	Time:240 ms	Time:260 ms
Low Severity Neck Injury Case		
Control		

6.5.2 Oblique near-side acceleration data

The lateral acceleration for first thoracic vertebra, from the oblique control (blue) and LSNI (red) cases is presented in Figure 6.17. The lateral acceleration between the two trials is similar with the T1 initially decelerating in both cases with the control case peaking at -32 m/s² at 92 ms, and the LSNI case reaching a peak at -33 m/s² at 100 ms. At this point, the deceleration slows for both cases passing through zero. Following this, there is rapid acceleration with the LSNI case reaching a peak at 67 m/s² at 128 ms and the control case reaching a peak at 71 m/s² at 140 ms. For each of the cases the data becomes extremely noisy and difficult to interpret. This coincides with the sash belt contacting the neck. This sash belt contact with the neck suggests that the lateral kinematics of the occupants in oblique impacts are comparable to the occupants in 90° near-side impacts. During the previously presented 90° near-side cases in section 6.5, the sash belt also contacted the neck in both the control and injury case.

The lateral head acceleration (y-direction) results for the oblique cases are presented in Figure 6.18. The predicted results suggest that the lateral head magnitude of the acceleration is similar between the low severity neck injury case (red) and the control case (blue) throughout. In both trials the head decelerates with the control reaching the peak of -35 m/s² at 91 ms and the LSNI case reaching the peak of -34 m/s² later at 101 ms. At this point the head deceleration slows in both cases reaching zero. This is followed by rapid acceleration reaching a peak of 38 m/s² at 162 ms for the control case and 41 m/s² at 169 ms for the LSNI case.

The frontal acceleration (x-direction) for first thoracic vertebra in the oblique impact for the control and LSNI cases are presented in Figure 6.19. There are also similarities between frontal acceleration for the first thoracic vertebra between both cases. The control case has slight initial deceleration reaching a peak of -16 ms² at 14 ms. There is rapid initial deceleration for the LSNI case reaching a peak of -39 m/s² at 41 ms. The LSNI deceleration slows to 4 m/s² at 69 ms, at this point the deceleration reaches a peak of 64 m/s² at 129 ms this peak coincides with the sash belt contact with the neck and following this point the predictions become noisy and difficult to interpret. In the control case, the deceleration slows to 4 m/s² at 68 ms, at this point there is an increase in the deceleration that reaches a peak of -57 m/s² at 127 ms. Following this point, the results become noisy and difficult to interpret. This is because at this point the sash belt contacts the neck. Kallieris et al. (1990) has reported that the neck experiences a higher maximum loading in 60° impact than 90° impact. From this finding, it would be expected that the neck loading in the 45° oblique impact would be greater than the 90° impact.

There is also little difference between head frontal acceleration results for the oblique cases presented in Figure 5.20. In the LSNI case the head decelerates in the frontal direction peaking at 95 m/s^2 at 155 ms. In the control case, the head also decelerated in the frontal direction peaking at a lesser value of 82 m/s^2 at 155 ms.

As with the 90° far-side case presented previously there was little variation in head and T1 acceleration. This is inconsistent with the variation in mass between the impacting vehicle and case vehicle. In the LSNI trial the case vehicle has a greater mass (1788kg) than the impacting vehicle (1426kg). In the control case the case vehicle and the impacting vehicle have a similar mass. These findings are inconsistent with that reported in the literature. The literature (Von Koch et al. 1995) reports that typically LSNI occurs when the impacting vehicle has a greater mass than the case vehicle. This suggest that the variation in mass between impacting vehicle an case vehicle may not have as great an influence on injury in obliques impacts as in 90° side, front or rear impact.

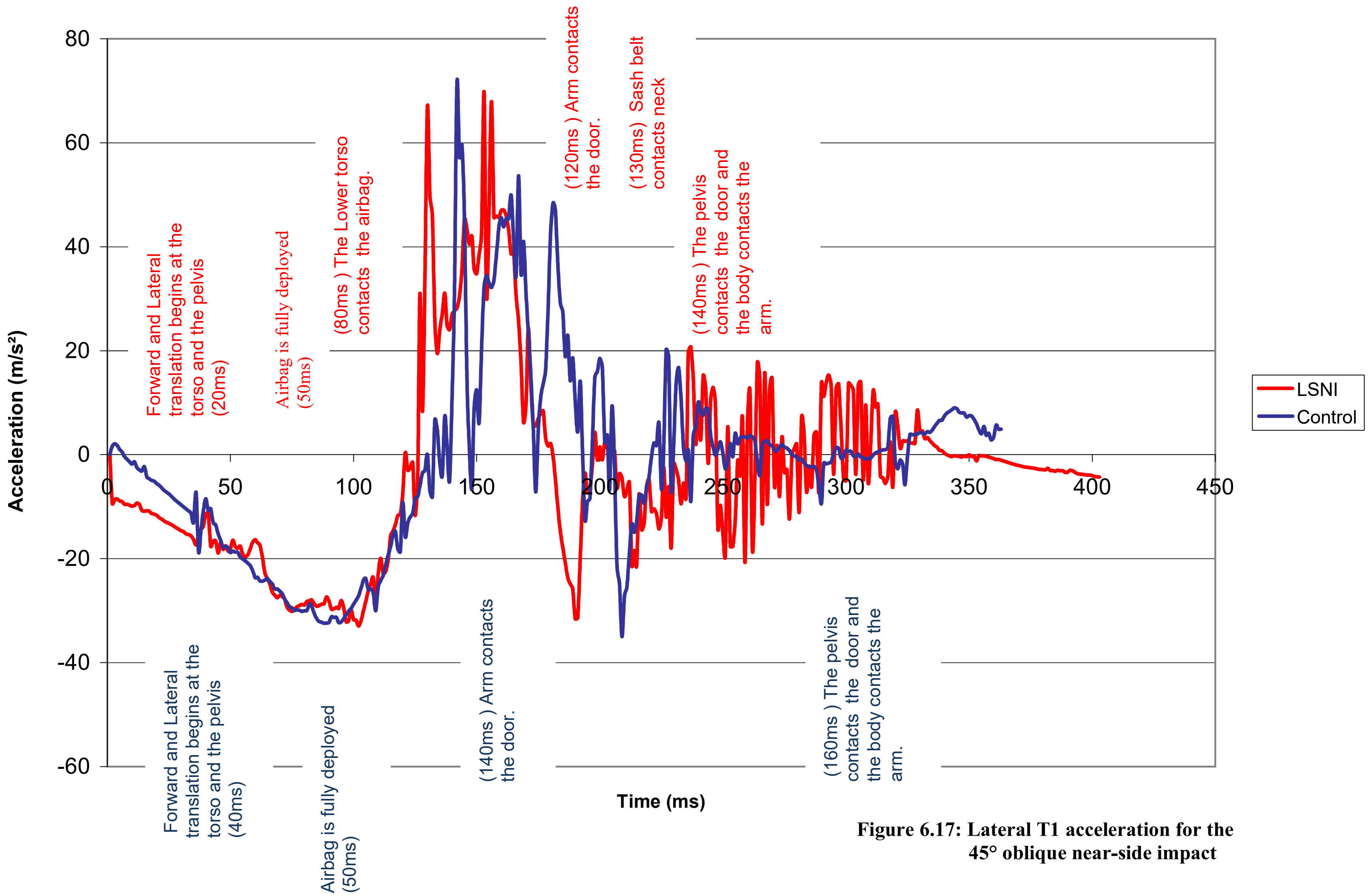


Figure 6.17: Lateral T1 acceleration for the 45° oblique near-side impact

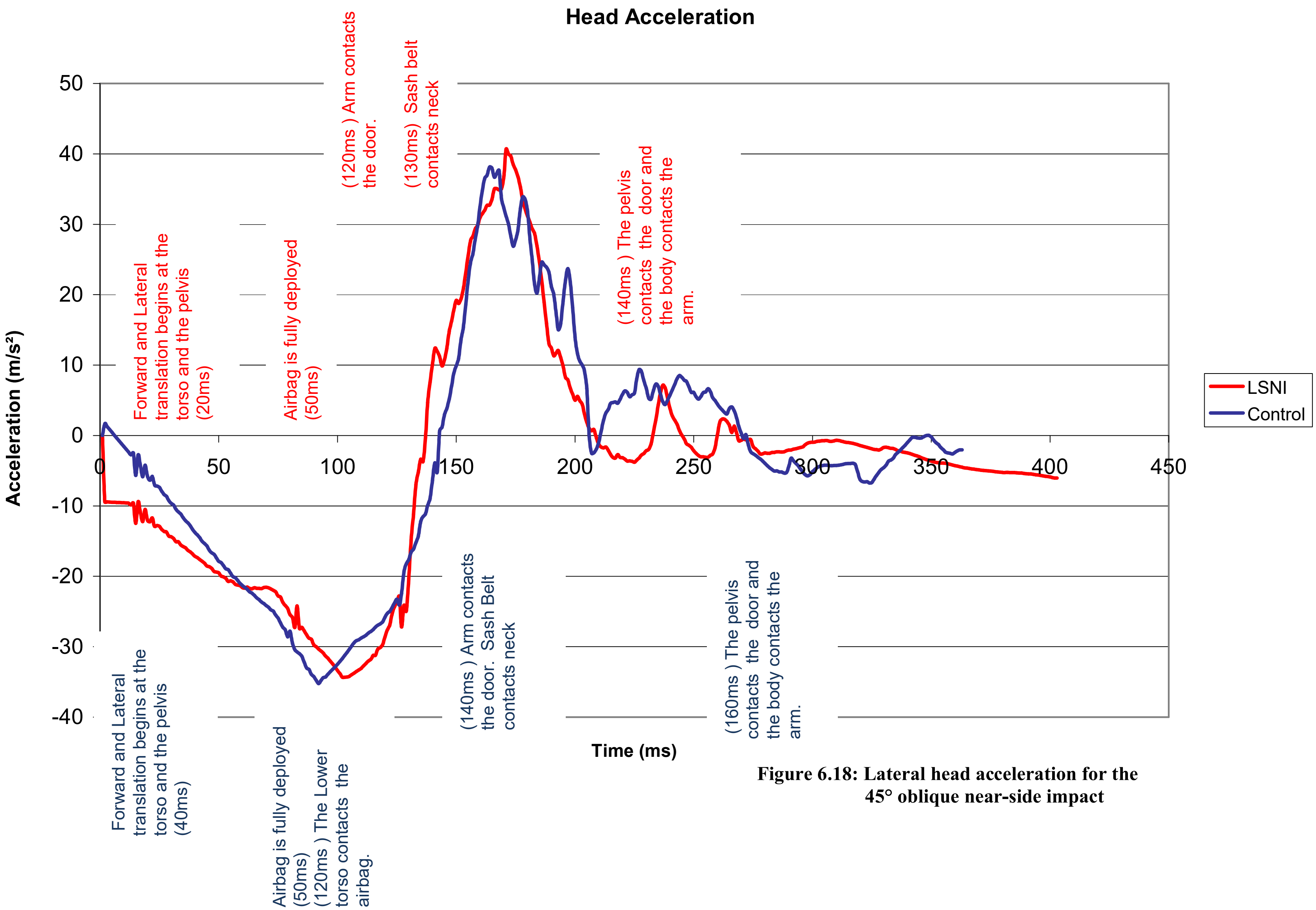


Figure 6.18: Lateral head acceleration for the 45° oblique near-side impact

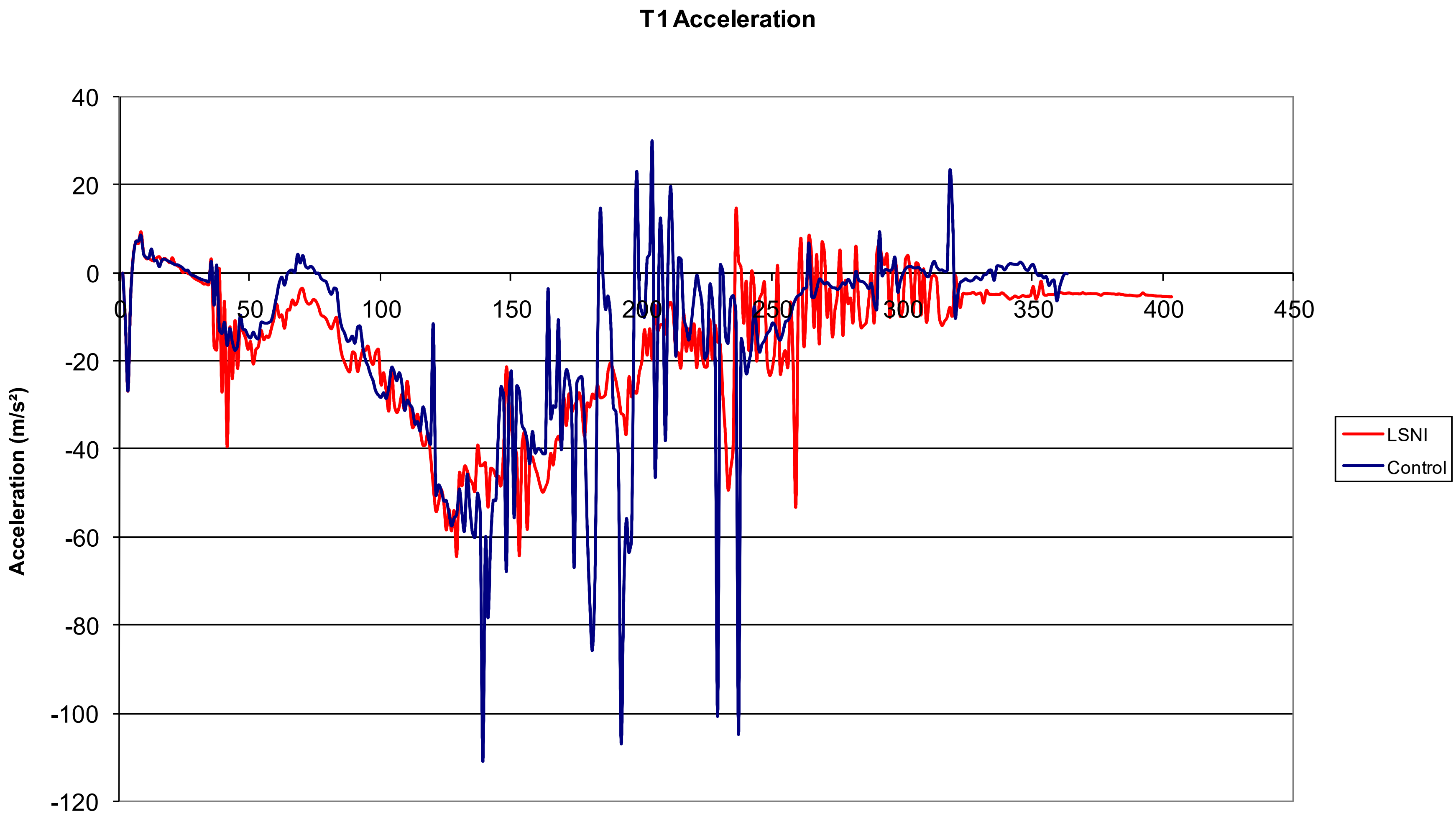


Figure 6.19: Frontal T1 acceleration for the oblique 45° near-side impact

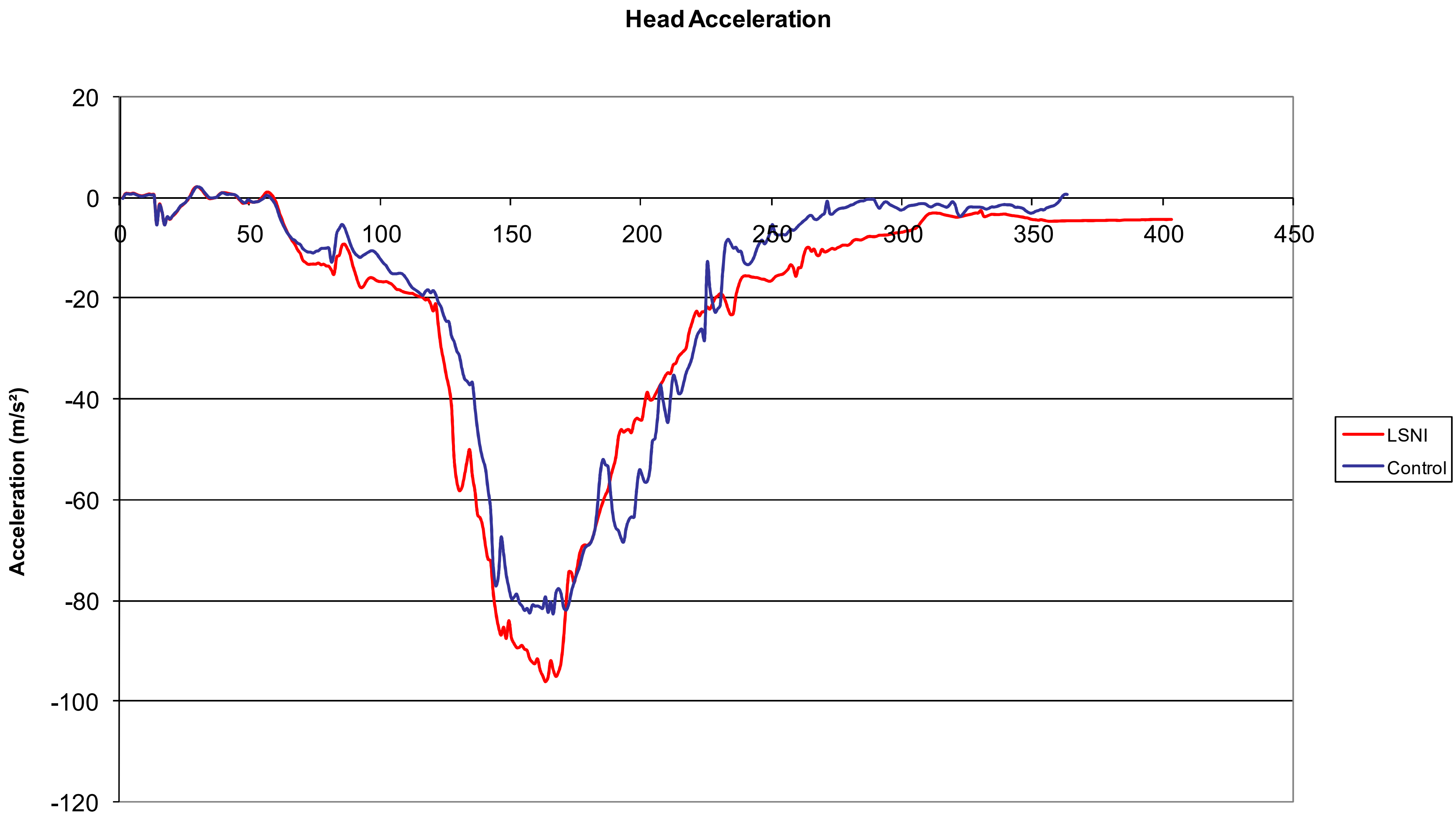


Figure 6.20: Frontal head acceleration for the 45° oblique nearside impact

6.6 Comparison of Impact Directions

Some additional factors have been suggested in the literature as contributing to LSNI. Appendix 5 illustrates the kinematic data, with the combined human model with the facet skin displayed as transparent, to show the detail of the cervical spine. The images are presented for all trials. Panjabi et al. (1998) suggest that LSNI occurs because cervical spine forms an s-shape during impact. Davidson et al. (1998) also suggests that torso ramping contributes to low severity neck injury. The kinematic data in this study in both this chapter and Appendix 5 shows no s-shape in the cervical spine, even in the cases where there was neck contact with the sash belt. There is also no evidence of torso ramping in any of the simulations.

Some of the additional factors that have been identified in literature as contributing to LSNI, were beyond the scope of this study to investigate. Head contact has been identified as an important factor in low severity neck injury. It is important to note that the occupant in the 90° near-side LSNI case reported a laceration to the right side of the head. The occupant in the 90° far-side LSNI case reported lacerations to the left side of the head. Head contact was not reconstructed in the computer simulations as it is inconclusive that these head lacerations were the result of a head contact or flying debris.

Horsch et al. (1979) identified that the subject's seating posture and height had an influence on the loading of the belt against the neck. Exploring the anatomical differences between subjects and the occupant's initial postural position is also beyond the scope of this study. The head restraint does not appear to influence LSNI as there is no contact between the head and the head restraint in any of the trials. The inclusion of an airbag in the steering wheel in the 45° oblique near-side case also had no influence on the injury. This is because there was little interaction between the occupant and the airbag at this impact angle. Investigating the influence of side airbags was beyond the scope of this study (due to there being no cases in the crash database at the commencement of this research).

6.7 Summary of findings

The purpose of this chapter was to investigate the head and neck mechanism of LSNI from side impacts. This was undertaken by reconstructing six real world cases, three where an occupant received a LSNI and three control cases where the occupant received no neck injury. Three different impact directions were investigated, 90° near -side, 90° far-side and a 45° near-side oblique impact. The kinematic and acceleration results in Sections 5.3 to 5.5 show that the head and neck mechanics differ for each impact angle. This is consistent with the findings of Kallieris et al. (1991) who found that there was a different mechanism of neck injury for near and far side impacts. The near-side impacts (90° and 45°) differ from the far-side impact (90°) primarily in the way that the sash belt interacts with the neck. In the near-side impacts, the sash belt contacts the neck providing a rotation point for the head. In the far-side impact there is little contact between the torso and the sash belt and no contact with the neck.

In this study, the lateral head accelerations and the lateral first thoracic accelerations were examined to determine the mechanism of injury. The peak lateral head accelerations for each of the impact angles for each of the cases were found to be: nearside low severity neck injury 128 m/s^2 and control 288 m/s^2, far side low severity neck injury 99 m/s^2, and control 117 m/s^2; oblique low severity neck injury 41 m/s^2 and control 38 m/s^2. The peak lateral acceleration for the first thoracic vertebra for each of the impact angles were found to be: nearside low severity neck injury 290 m/s^2, and control 39 m/s^2; far side low severity neck injury 68 m/s^2 and control 194 m/s^2, oblique low severity neck injury 67 m/s^2, and control 71 m/s^2. It is difficult to determine the influence that peak lateral acceleration has on the mechanism of LSNI from these cases alone. In the near-side impact, the control case has a greater peak head acceleration than the low severity case. The far-side and oblique impact cases show that the LSNI cases have only a slightly greater lateral acceleration. The peak accelerations for the first thoracic vertebra are equally inconclusive. The near-side impact data show the LSNI case has a far greater peak T1 acceleration than the control case, although the data shows the LSNI resulting in a smaller peak T1 acceleration than the control case. The variability in the results may be a result of what is shown throughout the literature regarding rear impact, that LSNI from side impact is not just influenced by impact direction, and a number of other factors couple with the impact direction to result in injury.

The acceleration histories paired with the kinematic predictions may provide more insight to the potential mechanism of neck injury of low severity neck injury. The kinematic results show, that during the 90° near-side impact and the 45° oblique near-side impact the sash belt contacts the neck. This contact occurs in both the LSNI and control cases. The sash belt contact with the neck was also shown to have an influence on the acceleration data with the peak accelerations in the lateral 90° near-side cases occurring at the time of contact. The kinematic data also shows that, following the contact between the sash belt and neck, there is head rotation about the vertical (z) axis. Further investigation is required to determine the mechanism of low severity neck injury. This investigation should include the influence of head rotation about the z axis and the loading of neck as a result of sash belt contact. Further research is required to determine if there is a relationship between head contact and LSNI.

6.8 Summary Table of Acceleration Data

Table 6.10 presents a summary of the model output data from the simulations undertaken in this chapter. The results have been discussed in the respective results sections in this chapter.

Table 6.8: Summary Table of Acceleration Data

Variable	LSNI Case	Control Case
90 degree near-side Acceleration Case		
First Thoracic Lateral Acceleration		
First Peak Acceleration	56m/s²	59m/s²
Time of First Peak Acceleration	110ms	218ms
Second Peak of Acceleration	290m/s²	39m/s²
Time of Second Peak Acceleration	153ms	264ms
Head Lateral Acceleration		
First Peak Acceleration	-48m/s²	-50m/s²
Time of First Peak Acceleration	117ms	264ms
Second Peak of Acceleration	48ms²	288m/s²
Time of Second Peak Acceleration	137ms	268ms
Third Peak Acceleration	128ms²	-
Time of Third Peak Acceleration	190ms	-
90 degree Far-side Acceleration Case		
First Thoracic Lateral Acceleration	LSNI Case	Control Case
First Peak Acceleration	68m/s²	94m/s²
Time of First Peak Acceleration	100ms	100ms
Second Peak of Acceleration	13m/s²	50m/s²
Time of Second Peak Acceleration	115ms	119ms
Third Peak Acceleration	32m/s²	63m/s²
Time of Third Peak Acceleration	132ms	149ms
Head Lateral Acceleration		
First Peak Acceleration	99m/s²	117m/s²
Time of First Peak Acceleration	103ms	100ms
Time Acceleration returns to zero	147ms	164ms
45 degree Oblique Acceleration Case		
First Thoracic Lateral Acceleration	LSNI Case	Control Case
First Peak Acceleration	-33m/s²	-32m/s²
Time of First Peak Acceleration	100ms	92ms
Second Peak of Acceleration	67m/s²	71m/s²
Time of Second Peak Acceleration	128ms	140ms
Head Lateral Acceleration		
First Peak Acceleration	-35m/s²	-34m/s²
Time of First Peak Acceleration	91ms	101ms
Second Peak of Acceleration	41m/s²	38m/s²
Time of Second Peak Acceleration	169ms	162ms

Chapter VII
Discussion and Conclusions

7.1 Introduction

A much needed investigation into low severity neck injury (LSNI) from side impact crashes, has been presented in this thesis. This was undertaken to the expand base of knowledge regarding this injury from other impact directions, in particularly rear impacts and some front impacts. This research is unique, as it is the first time that in depth analysis of real world side impact crashes has been undertaken to determine the factors that contribute to the injury. In this chapter key items are discussed. These are;

- What is understood about LSNI from side impact,
- What contributions to the knowledge of LSNI, does this thesis make,
- What are the limitations of this research,
- What are the future applications of the findings in this research.

7.2 Summary and Discussion

Low Severity Neck Injury has been one of the most prevalent disabling (non-fatal) injuries resulting from motor vehicle crashes. The high cost of this injury to the community is not disputed. LSNI from rear impact has been researched extensively across a number of decades. This has produced a wide body of work regarding the common factors associated with injury as summarised in Chapter 2. The incidence of LSNI has been reported widely. When considering the reported incidence of LSNI from side impact, existing research reports it to occur at a rate of 8.8% to 26% (Teamming et al., 1998, Jakobsson 1998, Morris et. al, 1996, von Koch et. al. 1995).

No work has been previously undertaken that takes only cases where LSNI caused from a side impact, examine the factors that cause injury. In spite of this, work has been previously undertaken investigating head and neck motion during lateral impacts, using either post mortem human subjects or volunteers. Previous work undertaken by Horsch et al. (1979) and Kallieris (1990), investigated the occupant's interaction with the seat belt during lateral impact. The finding of that early piece of work was that seat belts provide limited protection in side impacts. Typically, past PMHS and volunteer research investigates lateral impacts at an impact angle of 90° (equivalent to 3 o'clock and 9 o'clock impacts). Limited work exists considering occupant mechanics at other side impact directions. Some earlier work undertaken by Kallieris et al. (1990) investigated an additional side impact angle of 60° and Faeber (1982) investigated additional side impact angles of 60° and 120°. The findings in these earlier studies are of interest to this piece of work as Kallieris et al. (1990) found that neck loading was greater in the 60° impacts (an impact direction less than 90°. Faeber (1982) made a contrasting finding, that occupants received the best protection from seat belts in the 60° and 120° cases compared to those in 90° impacts. How these findings relate to this piece of work is discussed in the following section the contributions that this thesis makes to knowledge.

7.3 Contributions to the Knowledge of LSNI

To address the gap in research that investigates LSNI from side impact, two data bases containing real world crashes were investigated. These databases were the Monash University Accident Research Centre database (Australia) and the Loughborough University Co-operative Crash Injury Study (UK). This work makes a contribution to research as it identifies the incidence of LSNI (rate of injury) as well as factors that are common to occupants receiving a neck injury and the common characteristics of side impact where an LSNI has occurred.

7.3.1 Common Factors of LSNI in Side Impact

In this research has been identified that in the Australian database 39.5% of occupants receiving a LSNI were in a side impact. In the UK database 18% of occupants receiving a LSNI were in a side impact. The Australian data is higher than that in the UK database and that reported in the research presented in Chapter 2, this may be due to the Australian database excluding crash angles such as rear impacts and rollovers. In spite of Australian data being skewed, this research and other side impact low severity data shows that the injury is of concern and warrants analysis and measures, with the focus of reducing future incidence. This is supported by the recent longitudinal research by Styrke (2012) that shows that successful vehicle design changes to seats and head restraints are reducing the rate of LSNI from rear impacts, but as a result of this the relative incidence of LSNI from side impact is increasing.

This thesis expands on those factors that are understood to be important in rear impact and explores their importance in the context of LSNI in side impact. The common factors of LSNI from rear impact are gender (female), seatbelt use, age (adult) and poor head restraint positioning and geometry. Due to the large lateral component of side impact these factors can not automatically be assumed to be a relevant in LSNI from side impact. This thesis makes the contribution to research by identifying the common factors associated with low severity neck injury in side impacts. The Australian data shows that a similar number of males and females receive a low severity neck injury from side impact. The UK data shows that more females (39%) compared to males (27%) receive a low severity neck injury in a side impact. Both databases showed that drivers experience the highest incidence of low severity neck injuries from side impacts. The Australian data in this research shows that male occupants were equally affected as female occupants, although the common age for injured occupants differs between the genders. Males aged between 37-57 were more commonly injured, compared to females aged between 17-26. More drivers were injured compared to any other seating position in the vehicle. Near side impacts were found to be more common than far side impacts and car to car crashes were the most common crash configuration. An important finding in this research is that LSNI is shown to occur in side impacts over a variety of impact angles. The most common impact directions that were found to result in LSNI were, lateral 90° (3 o'clock and 9 o'clock) and oblique angled impacts of (1 o'clock, 2o'clock, 10 o'clock and 12 o'clock). An interesting finding in this piece of work identified that 18% of Australian occupants and 42 % of UK occupants who received a LSNI also received a head injury. This finding shows that future preventative factors to reduce the incidence of LSNI, would also need to limit head contact during a side impact.

7.3.2 Occupant Kinematics

The findings in the previous section that shows that LSNI can occur at various crash angles supports the need for human surrogates/ human models to accurately replicate human motion at a multiple side impact angles. This thesis contributes to the body of research as the combined human model was validated at additional impact angles that had not previously been undertaken in previous research. The combined human model was compared to Wayne State University Cadaver data and Naval Biodynamics volunteer data for the 90° computer simulation. For the 45° simulation the combined model was compared to the Naval Biodynamics volunteer data. The 45° oblique setting as the combined model had not been previously validated at this impact angle. This work was undertaken in this study to test the models suitability to investigate real-world impacts at crash directions of 1'oclock, 2 o'clock, 10 o'clock and 11 o'clock impacts. The comparisons of the head and first thoracic vertebra lateral acceleration showed the model to be a suitable tool to investigate head and neck mechanics in side impacts. The kinematic data for the model showed a good likeness to the experimental data, with respect to the key events and the timing of these events.

In the final stage of the investigation (presented in Chapter 6), six real world side impact cases where reconstructed in detail. The direction of impact of the chosen cases, were comparable to the impact angles identified in the real world data (Chapter 4) as typical. To establish the acceleration of the case vehicle during the impact, the computer software, HVE (Human, Vehicle, Environment) was used. Three different impact angles were investigated with two cases in each group, one where an occupant received a low severity neck injury the other a control case where an occupant did not receive a low severity. The control case was investigated for comparison. The crash angles reconstructed were 90° near side, 90° far side and 45° oblique (1 o'clock) near side impacts. The occupant position investigated was the driver for all cases. A key finding of this research is that the occupant kinematics and the occupant's interactions with the interior of the vehicle differ at each side impact direction. In the 90° near side impact case and the oblique near side cases it was identified contact between the sash belt and the neck. As a result of this contact the head and upper neck rotate about the sash belt in the z-axis. As this contact was found to occur in both the injury case and the control case it cannot be identified as a mechanism of injury from this research. The contribution that this thesis makes to research is that this methodology has not been used before to investigate neck injury from side impacts before. This work shows that in spite of the advances in the preventative factors designed to reduce LSNI in rear impacts are not effective in reducing LSNI in side impacts. This research also shows that head restraints are ineffective in preventing neck injury in side impacts as there was no interaction between the occupant's head and the head restraint in any of the trials. These finding is relevant to the new work (Linder et al. 2013, Dehner et al. 2013, Ziraknejad et al. 2013) being undertaken in head restraints to prevent or mitigate LSNI injury. The new head restraint technology develop to prevent LSNI in rear impacts, will not automatically be transferrable to side impact crashes.

7.3.3 Limitations of Research

Further work is still required to develop a full understanding of LSNI resulting from side impacts. Research shows that males and females are equally likely to receive a LSNI form a side impact. A limitation of this research is that the crashes reconstructed were only undertaken for male occupants. The occupant kinematics and results in Chapter 6 are not directly transferrable to female occupants. This is to due variations in anatomy between male and females including height.

All of validations of the detailed neck model have been undertaken by using measurement taken at head and first thoracic vertebra. Validating this model at each of the cervical spine levels could lead to a more accurate and human like model. Due to the limited cadaver spine data available for later impacts, these validations were beyond the scope of this research.

A limitation of the data analysis and modelled crash reconstructions undertaken in this thesis, is that only cases with front airbags were investigated. Since this research commenced new Australian vehicles are installed with side airbags. In a side impacts, side airbags are likely to influence the occupants head and neck mechanics. How this would influence LSNI was not address in this piece of research.

7.4 Future Direction of the Research

In spite of the new developments in automotive design such as active head restraints and seats, rates of low severity neck injury are not decreasing. Over a 10 year period rates of LSNI were shown to increase, although work does show rate from rear impact are in decline in Sweden Stryke (2012), other research shows that rates for rear impact have decreased for males but more work needs to be done for females Linder (2013). Further work is required to determine effective safety measures to reduce LSNI from side impacts at all side impact angles. With the wider introduction of side impact airbags into Australian vehicles, it is important to investigate the deployment of these airbags and occupant kinematics with respect to contact between the sash belt and the neck. The role that head rotation plays in injury mechanism and how head rotation coupled with lateral bending in 90° impacts and how it is coupled with head flexion in the 45° oblique cases. Future research is also required investigate the relationship between head contact and low severity neck injury. Given the finding in this research that a high number of occupants receiving a LSNI also received a head injury, this is a possible additional mechanism of LSNI. Any future research into low severity neck injury also need to consider any anatomical factors that may predispose an occupant to LSNI as well as occupant seating posture prior to impact and seat geometry.

The findings in thesis that shows the common factors of an occupant receiving a LSNI was a driver in a near side car to car impact. It is important to note that this impact configuration in Australia and the UK is to the driver's right side which is the same direction that a driver is required to give way to at intersections and roundabouts. This provides the opportunity for future driver education and behaviour change programs to include components on promoting driver awareness and the importance of giving way to other vehicles to prevent LSNI.

Appendix I
Anatomy of the Cervical Spine

This section provides a literature review of the anatomy and biomechanics of the cervical spine (neck). This has been provided for two reasons. The first reason is to make a comparison with the combined human model that has been combined with the human body model used for analysis in this thesis (Chapter 3, Chapter 4 and Appendix 3). The second reason is to provide a base of understanding regarding the mechanism of a cervical spine injury and the structures within the cervical spine that are likely to be injured. In this review a detailed description of all of the bones, muscles, ligaments, discs and the mechanical limitations of the cervical spine has been included. Research investigating healthy motion of the cervical spine, as well as the spines tolerance to abnormal motion has presented at the end of this review.

A1.1 The Vertebral Column

The cervical spine is a part of the vertebral column. The unique function of the vertebral column is its ability to provide support and yet be considerably flexible. Each structure of the vertebral column has a specific function. Bones, muscles, ligaments and discs combine create "natural motion" as well as limiting “undesirable motion” preventing injury.

The vertebral column has three functions, providing;

- support and stability for the body
- attachments for the skull, thorax and pelvic girdle
- protection for the spinal chord

The vertebral column (Figure A1.1) is divided into 5 parts. These are the cervical, thoracic, lumbar, sacral and coccyx. The cervical spine consists of 7 bones (vertebrae), the thoracic spine has 12 vertebrae and the lumbar spine has 5 vertebrae. The sacrum and coccyx are greatly different in structure. The sacrum consists of 5 fused vertebrae and the coccyx consists of 4 fused vertebrae. The normal adult spine has three natural curves (Gray, 2001).

Figure A1.1: The vertebral column- lateral view (Gray, 2001)

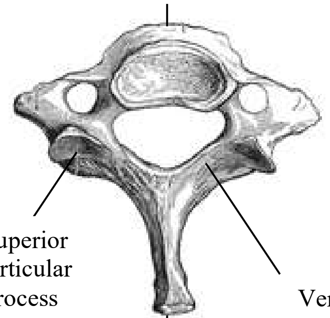
A1.2 The Cervical Spine

The cervical spine is the region of the spine that is of interest in this piece of research and therefore, has been described in detail. The cervical spine has many unique structures that differ from those of other regions of the vertebral column. These structures allow the cervical spine to have a greater range of motion than any other part of the vertebral column. It is also important to understand the sensitivity of these structures to undesirable motion and loading and the injuries that occur when they fail. Death or disability is a likely outcome of a serious cervical spine injury. Literature also shows that disability can result even from minor injuries to the cervical spine (Kraft 1998). The following section describes the anatomical structures that make up the cervical spine, describing normal function and likely failures of these structures.

A1.3 Bone

The first seven bones of the spine make up the cervical spine. No two cervical vertebrae are identical (Mertz, 1971). The cervical vertebrae increase in mass and the length of the spinous process, from the first cervical bone to the seventh cervical bone (Patrick, 1987, Mertz, 1971). The first (C1) and second cervical (C2) vertebrae are the most unique, and have the specific function of generating head motion. Vertebrae C3 through to C7 generally have a similar shape and fulfil a similar function. Each of these vertebrae have four facet joints, two that face upwards and two that face downwards. This allows the vertebrae to lock together providing stability to the spine, these joints are discussed further, later in this section. A typical cervical vertebrae is shown in Figure A1.2 with each main part labelled and the function listed below (Gray, 2001).

Figure A1.2: Typical cervical vertebrae



Spinous Process	Allows for attachment of muscles and ligaments
Vertebral Forearm	These align to form the vertebral canal, through which the spinal cord passes
Body/Centrum	Each body in the cervical spine is separated by an inter-vertebral discs
Superior Articular Process	These are articulating surfaces that form part of the joints

A1.3.1 First and Second Vertebrae (Atlas and Axis)

C1 "atlas" and C2 "axis" vertebrae differ in structure to the other cervical vertebrae as there is no intervetebral disc between C1 and C2. This allows for forward and backward movement of the head (flexion/extension). The atlas (C1) has no vertebral body and no spinous process like the other cervical vertebrae. This bone is structurally different to bear the weight of the skull, as the superior articular surface of the atlas forms a synovial joint with the occipital condyles of the scull (Mertz, 1971). The primary movement of this bone (the alanto- occipital joints) is flexion and extension (Patrik 1987). A pivot is formed by the odontoid process of the axis (C2) that allows the atlas and the attached head to rotate as the primary movement of the axis is rotation. Compression of the spinal chord from unwanted posterior movement is prevented by the function of the transverse ligament of the atlas (see ligaments section) (Mertz, 1971).

A1.4 Bone Fracture

Bone injury is not usually associated with low severity neck injury as this is considered to be a more severe injury. A bone fracture of the upper cervical spine (atlas/axis) can result in death. The main mechanism of bone injury, being axial compression, usually occurs with head contact through severe impacts. The greatest associated risk with bone injury in the cervical spine, is the potential of spinal cord injury. As the individual vertebrae are connected, a cylindrical pathway (neural canal) is formed, to house the spinal cord. This pathway is usually well protected by the neural arch, although intrusion into this pathway can occur, with the extreme force resulting in damage to the spinal cord. Spinal cord injury can result in death, if the injury is to the upper cervical spine (atlas or axis). A spinal cord lesion to the neck that does not cause death, usually results in paralysis of all four limbs and is called quadriplegia (Spence, 1990). This is the most serious type of injury is rates an AIS5 or AIS6 (AAAM, 1985).

The association between low severity neck injury and bone injury is not well documented. The Abbreviated Injury Scale (AIS) does not classify bone injury as a type of low severity neck injury. Only soft tissue sprains and strains are considered. In addition the Quebec taskforce for whiplash does classify bone fracture as a low severity neck (whiplash) injury (Spitzer et al. 1995).

A1.5 Ligaments

Ligaments are the connective tissue that joins a bone to other bones. The ligaments have three major functions as reported by (Gross et al. 2009):

- to enhance the mechanical stability of joints
- to guide joint motion
- to prevent excessive/unwanted motion.

The ligaments that are associated with the joints of the cervical spine are anterior longitudinal ligament, posterior longitudinal ligament, ligamentum flavum, intertransverse, atlantal cruciform ligament and the alar ligament. These ligaments are illustrated in figures A1.3 and A1.4.

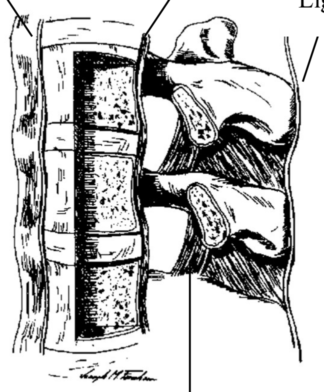


Figure A1.3: Major ligaments of the spine (Norkin et al. 1992)

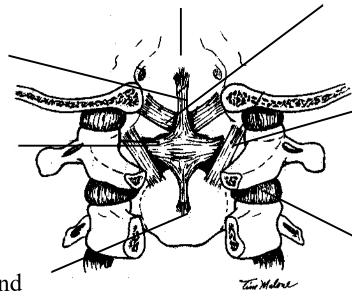


Figure A1.4: Deep ligaments of the spine (Norkin et al. 1992)

The anterior longitudinal ligament runs from the sacrum to the second cervical vertebrae where it becomes the anterior atlanto-axial ligament. It reinforces the discs as it blends with the fibres of the anulus fibrosis of the disc. This ligament is compressed in flexion and in extension. It may be slack when the cervical spine is in neutral (in resting position) or when there is disc damage.

The posterior longitudinal ligament runs through the vertebral canal from the sacrum to the second cervical spine where it becomes the tectorial membrane. It also provides reinforcement to the discs. The posterior longitudinal ligament experiences highest strain when it is in flexion when the ligament is slack in extension. With the presence of disc damage this ligament may be stretched in extension.

The ligamentum flavum runs from the sacrum to the second cervical vertebrae along the posterior surface of the vertebral canal. Part of this ligament covers the articular capsules of the zygopophyseal joints. This ligament is under the highest strain when it is stretched during flexion. The tension in this ligament is constant when the cervical spine is in neutral position. This tension increases the stability of the spine. The ligamentum flavum is weakest in the mid-cervical region.

The previously mentioned ligaments also act on the other regions of the spine. The following ligaments discussed, atlantal cruciform ligament and alar ligaments act only on the cervical spine. The atlantal cruciform ligament maintains stability of the atlantoaxial joint and prevents anterior displacement of the first and second cervical vertebrae. The alar ligaments are a pair of ligaments that are attached to the medial aspect of the occipital bone. These ligaments are activated in flexion and relaxed in extension. These ligaments limit rotation and lateral flexion of the head and neck.

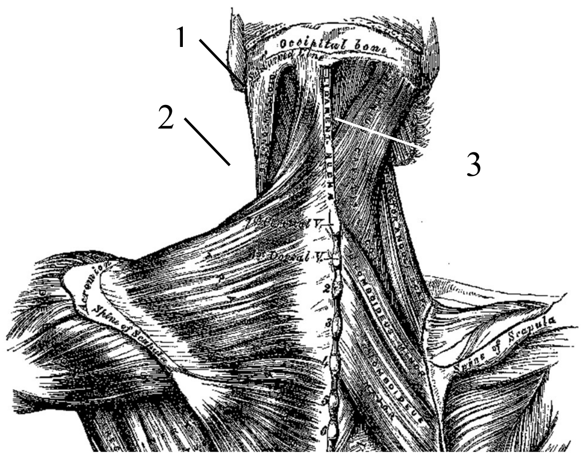
Ligaments have a high modulus of elasticity. Hence, it is possible for large forces to build up in ligaments with little movement. Once ligaments have reached the maximum force that they can tolerate, it only requires a small amount of additional force is required to produce a tear (Patrick, 1987).

A1.6 Muscles

A detailed diagram of the muscles acting on the cervical spine is presented in figures A1.5 and A1.6. Many of these muscles attach to the spinous process and to the transverse process of the cervical vertebrae. These muscles are arranged in symetrical pairs. The trapezius (2) is the largest of the muscles acting on the neck. It primary function is to produce scapular movements i.e. shrugging and bracing the shoulders, although the trapezius can produce some head movement. When one side of the trapezius is contracted the head will tilt to that side, when both sides of the trapezius is contracted the head is tilted backwards.

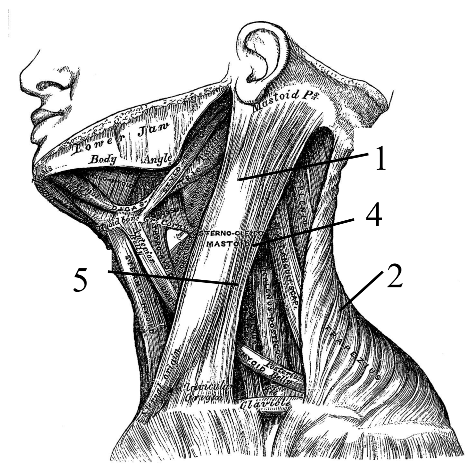
Some muscles in the neck such as splenius attach to the skull and move both the head and vertebral column. The sternoclidomastoid (1) located at each side of the front of the neck is one of the muscles responsible for head rotation. It does not attach to the cervical vertebrae although when both acting together the sternoclidomastoid flexes the neck. When acting alone, the head is rotated to the opposite side. The sternocleidomastoid can also cause or limit lateral bending (Patrick, 1987). This pair of muscles also resists flexion and extension under inertial loading (Patrick, 1987).

The muscles longus capitus, longus colli, rectus capitus anterior, and the scalenus anterior (5) flex the head and neck as well as resisting extension under inertia loading (Patrick, 1987)



1. Sternocleidomastoid
2. Trapezius
3. Splenius Capitis

Figure A1.5: Posterior muscles of the neck (Gray, 2001)



1. Sternocleidomastoid
2. Trapezius
4. Posterior Scalene
5. Anterior Scalene

Figure A1.6: Lateral muscles of the neck (Gray, 2001)

A1.7 Muscle Injury

Muscles initiate movement through the tendons to the bones in the body. Muscle injury has been defined by Tidus (2008) as the loss of muscle function caused by the physical disruption of muscle structures involved in producing or transmitting force. The mechanisms of muscle injuries are varied. While skeletal muscles can generate high forces without failing, occasions can occur where the muscle can be injured from too much force being transferred through the tendon to the muscle (Whiting et al. 2008). The types of muscle injury can be divided into acute muscle strain, contusions and exercise induced muscle injury. Typically acute muscle strain occurs when a passive muscle is overstretched or an active muscle is either eccentrically or concentrically overloaded. The severity of acute muscle injury is influenced by three things: the magnitude of force, the rate of force application and the strength of musculo-tendinous structures responding to the force.

A contusion (bruise) to a muscle occurs as a result of a compressive impact to the muscle. The result of a muscle contusion is a intramuscular haemorrhage. Exercise induced muscle injury has not been discussed in this review.

A1.8 Joints

As previously mentioned, joints or articulations are formed when ligaments join bone to bone. The other connective tissue that assists to form a joint, is articular cartlage. A dense layer of articular cartilage covers the articular surfaces of joints. The articular cartilage has the two primary functions to spread the loads acting on the joints and reduce friction between opposing surfaces (Norkin et al. 1992).

There are two types of joints in the vertebral column, cartilaginous and synovial joints. Cartilagenous joints are created by bands of ligaments joining the adjacent vertebrae to each other enclosing the part intervertebral discs.

Synovial joints that are made up of the articular process of the vertebrae called zygapophyses. Zygopophyseal joints play a large part in the type and magnitude of vertebral motion (Lord et al. 1993). The joints are created between the inferior articular process of one vertebrae and the inferior articular process of the adjacent vertebrae (Lord et al. 1993). The zygopophyseal joints are angled at approximately 45° although the lower joints become steeper. It is the angled articular surfaces of these joints that limit the motion to only gliding. It is the obliquely angled articulating surface of the zygopophyseal joints that allow for gliding motion to occur as the articulating surface of the superior articular process of each joint faces backward and upward. As the joints glide with respect to one another the articular surfaces loose contact with one another called "sublaxating" It is here where the where the fibroadipose tissue covers the exposed articulating surface (Lord et al. 1993).

The zygopophyseal joints also play a role in resisting unwanted motion. Their oblique orientation resists excessive forward and downward displacement of the vertebrae Lord et al. (1993).

A1.9 Limitation of Joints

A variety of sources report zygapophyseal joint (facet joint) damage in association with low severity neck injury (Lord et al. 1993, White et al. 1990, Gibson et al. 2000). The types of damage produced are tears in the joint capsule and small fractures (Lord et al. 1993). The fracture produced is rarely visible by radiography and by advanced equipment is required to detect them.

Through experimental studies the anesthetising of suspected injured joints, it has been possible to associate the damage of the zygapophyseal joints with the symptom of pain. Gibson et al. (2000) undertook a study where ninety-two subjects exhibiting chronic neck pain (with both clinical and crash data available) were given a local anaesthetic. This anaesthetic was either injected into the space of the zygapophyseal joint or into the nerve supply feeding the joint. This work is of particular significance to this research because side impacts were accountable for 23% of the sample (with 11% left and 12% right). Rear impacts accounted for 40%, frontal 34%, and rollovers 3%. The anesthetic pain blocks were successful in 88 of the patients. The location of these blocks are presented in Table A1.1.

Table A1.1: Sites of Successful Blocks of Cervical Joints (Gibson 2000)

Segmental Level	Position of the Positive Blocks			
	Bilateral	Left	Right	Total
C2/3	6	7	17	30
C3/4	1	1	8	10
C4/5	1	1	3	5
C5/6	1	11	16	28
C6/7	1	8	6	15
Total	10	28	50	8

The joints at C2/3 level and C5/6 level responded most to the anaesthetic. The results showed that the right-side joints were more affected. When considering rear impacts, Gibson et al. has suggested that right side may be more frequently injured as Australian drivers (sitting on the right side of the car) may turn their head left and upwards to see the rear view mirror. The interaction with the seatbelt, positioned from the right shoulder may also have contributed (Gibson 2000).

A1.10 Inter-vertebral Disc

Inter-vertebral discs have been identified as the most important connections between vertebral bodies. The inter-vertebral discs attach to the vertebral body securely, so much that their removal is difficult without breaking pieces of bone Patrick (1987). The major parts of the disc are the anulus fibrosis, the retaining capsule and the softer central nucleus pulposus.

The functions of the discs are to bind the vertebrae, to absorb shock and to assist motion. They are able to fulfil their shock absorbing function due to their soft structure and their high water content. Therefore the function of discs can be affected by dehydration. The role of inter-vertebral discs in motion is to increase the movement of the vertebrae. This occurs as they compress and extend between the two vertebrae. During flexion, the anterior part of the annulus fibrosis compress and bulges, as the posterior part is stretched. In extension the bulge occurs posteriorly and the anterior part of the disc stretched (Norkin et al., 1992).

A1.11 Disc Rupture

Either rupture of the nucleus pulposus or a break in the annulus fibrosis causes fluid to leak from the disc (Jenkins, 1981). A disc rupture can cause a narrowing of the inter-vertebral space, although even without narrowing of the inter-vertebral space disabling symptoms can still occur. The disc is weakest posteriorly as the annulus fibrosis is thinner there. The most common disc protrusions occur either posteriorly or posterio-laterally, at the edge of the posterior longitudinal ligament. When a disc ruptures posterio-laterally, it is likely to press upon nerves in the area. This can produce a reflex spasm in the muscles of the back (Jenkins, 1981). Barnsley (1993) reports that damage to the inter-vertebral disc has been detected in those reporting a low severity neck injury from rear impacts. The damage includes separation of the disc from the vertebral end plate and tears of the anterior annulus fibrosis (Barnsley, 1993). These injuries may be explained by the extension produced by this kind of impact.

A1.12 Types of Motion

The cervical spine is able to produce three translations and three rotations. The three translations are lateral translation, axial compression, anterior/ posterior translation and the three rotations are lateral bending, axial rotation and flexion/extension. These have been illustrated and defined in the Figures A1.7 to A1.11.

Figure A1.7: Flexion

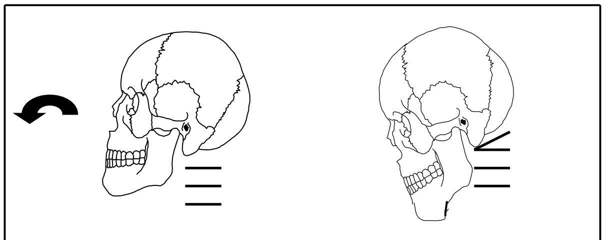


Figure A1.7

Flexion- Forward movement of the head, it is limited when the chin reaches the chest. Flexion is initiated by the shortening of the anterior muscles in the neck.

Principal Muscles: Sternoclidomastoid and Scalenes (Spence 1990)

Figure A1.8. Extension

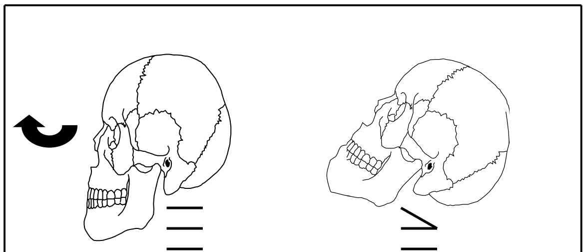


Figure A1.8

Extension- Backward movement of the head. Extension is initiated by the shortening of the posterior neck muscles, and the lengthening of the anterior neck muscles.

Principal Muscles: Semispinalis, Multifidus, Rotatores and Splenius (Spence 1990)

Figure A1.9. Lateral Flexion/Extension

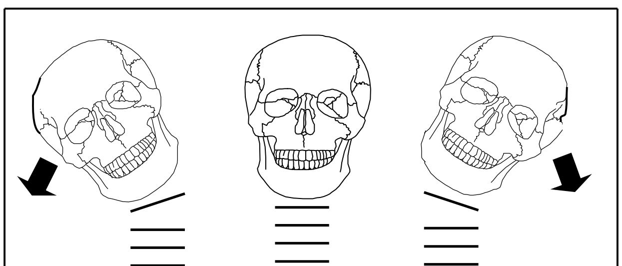


Figure A1.9

Lateral Bending- Sideways movement of the head bringing the ear towards the shoulder. The neck muscles of the side of movement are shortened while the opposing muscles are lengthened.

Principal Muscles: Intertransversarii and Levator scapulae(Spence 1990)

Figure A1.10: Rotation

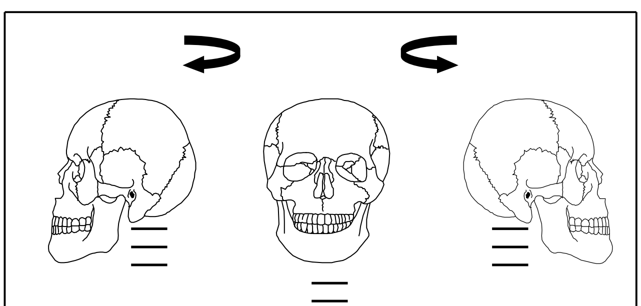
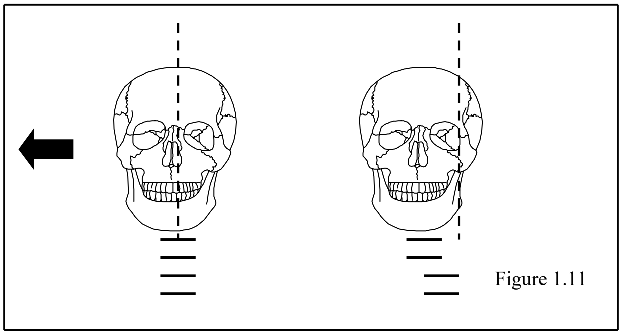


Figure A1.10

Rotation- Turning of the head bringing the chin to the shoulder.

Principal Muscles: Sternoclidomastoid, Semispinalis, Multifidus, Scalenes and Splenius (Spence 1990)

Figure A1.11. Lateral Translation



Lateral Translation- Lateral / Linear motion of the head left or right.

Other types of motion can be produced by combining (coupling) any of these six motions. Adjacent vertebrae only move a small amount in relation to each other, it is the addition of these small movements that result in greater movements of the whole vertebral column. Examples of two adjacent vertebrae interacting with each other during three translations and three rotations have been presented in Figure A1.12.

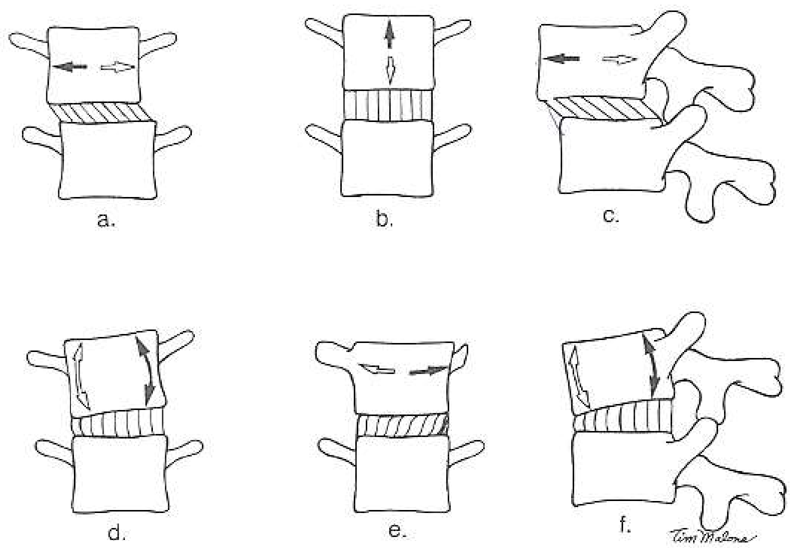


Figure A1.12: Motion between vertebrae (Norkin et al. 1992
a) lateral translation
b) superior/inferior translation (axial compression)
c) anterior/ posterior translation
d) side to side rotation (lateral bending)
e) axial rotation (transverse)
f) anterior/posterior rotation (flexion/extension)

A1.12.1 Axial Compression

Axial compression of the cervical spine can be produced from gravity and forces of ligaments and muscle (Norkin et al. 1992). The neck is also axially compressed when there is an axial impact to the head. The discs and vertebral bodies resist axial compression, and under normal circumstances, the vertebral column is able to maintain a state of equilibrium. This is done with the discs deforming under the axial compression distributing the force (stress) to the vertebral end plates (Norkin 1992). Although as mentioned previously, axial compression to the head can produce potentially fatal fractures to the cervical spine. (Nahum et al. 2001)

A1.12.2 Flexion

During head flexion, the anterior structures of the cervical spine are compressed as the posterior structures are stressed (Norkin et al. 1992). The anatomical structures that limit excessive flexion are the posterior outer annulus fibrosis, the joint capsule of the zygapophyseal joints and the posterior ligament (Norkin et al. 1992).

A1.12.3 Extension

During head extension the posterior structures are either unloaded or compressed as the anterior structures are stretched (Norkin et al. 1992). The anatomical structures that limit excessive extension are the anterior outer annulus fibrosis, the joint capsule of the zygapophyseal joints and the anterior ligament. In addition to these structures motion may also be limited by contact of the spinous processes (Norkin e. al. 1992).

A1.12.4 Lateral Bending

To achieve lateral bending the superior vertebrae tilts and rotates over the inferior vertebrae (Norkin et al. 1992). The behaviour of the cervical spine in lateral bending is dependent on the direction of the bending. Right lateral bending compresses the right side of the disc and left lateral bending compresses the left side of the disc. The opposite side of the disc is stretched (Norkin et al. 1992). During lateral bending, the outer annulus fibrosis and colateral intransverse ligament provide stability and limit unnatural motion.

Appendix 2
Collision Deformation Classification

The collision deformation classification is a standardised method of describing vehicle damage post-crash and is based on the recommendations of the Society of Automotive Engineers (1980). Using this method allows for vehicle damage to be compared between different crashes. There are 7 characters in this classification that describe; direction , location, size area and extent of damage. An example of a CDC is in seen in chapter 4 as 03rpew03.

The first two characters identify the principal direction of force. This is classified using a clock face to identify the direction that the impacting object struck the case vehicle. It is illustrated in Figure A2.1.

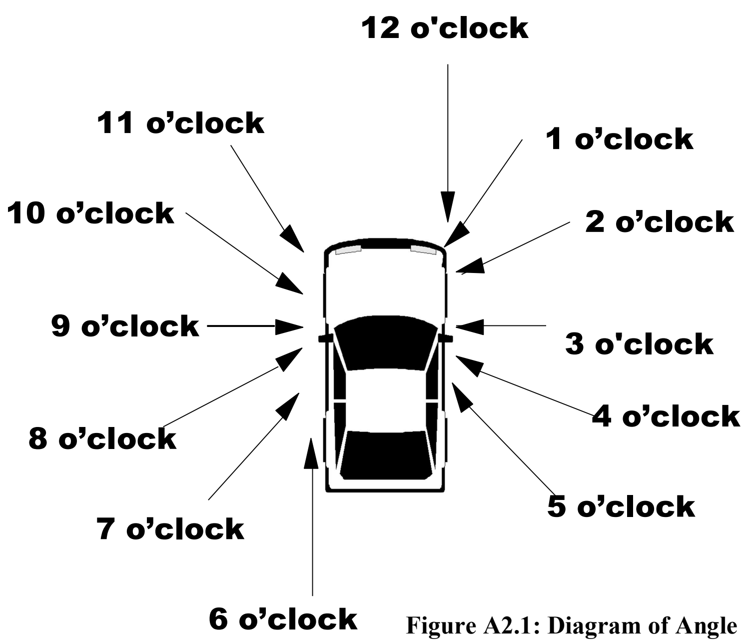


Figure A2.1: Diagram of Angle of Impact

The third character identifies the area of damage this is described generally as front, back, left, right, top or undercarriage (see Figure A.2.2). The distribution of damage is identified by the fourth character as illustrated in Figure 2.3. The height of damage is identified by the fifth character (Figure A2.4) and the depth of damage is identified by the sixth character (Figure A2.5).

Appendix 2
Collision Deformation Classification (cont')

A2.2

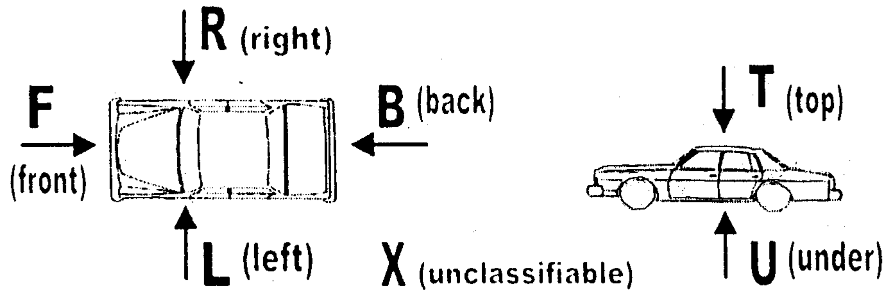
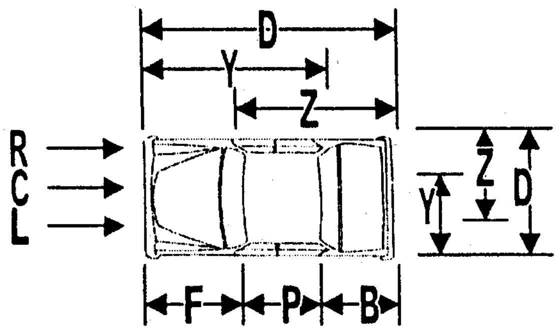


Figure A2.2: Area of Damage

A2.3



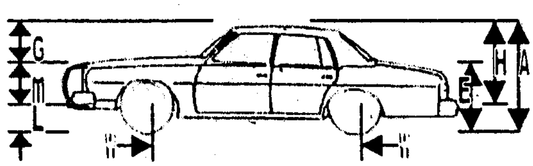
D Distribution
R Right
C Centre
L Left
P Passenger
B Back

Y = L+C
= F+P

Z = R+C
= P+B

Figure A2.3: Distribution of Damage

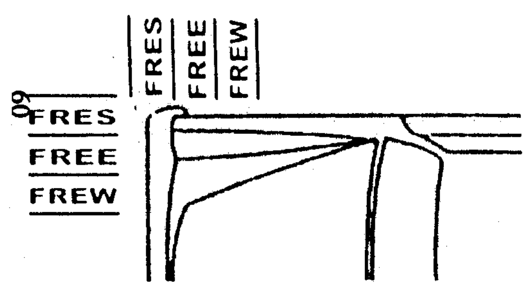
A2.4



A **All**
H Chassis Upwards
E Below Glass Level
G Glass Level Upwards
M Middle Section
L Chassis Downwards

Figure A2.4: Height of Damage

A2.5



W **Wide**
N Narrow
S Sideswipe
O Rollover
A Overhang/Under ride
E Corner

Figure A2.5: Depth of Damage

Appendix 3
Model Co-ordinate System

The co-ordinate system of the human model is presented in Figure A3.1. The figure shows the x- axis to be positive in the direction the model is facing. The y-axis is positive along the left arm. And the y-axis is positive upwards through the head. The co-ordinate system for the combined human model are the same.

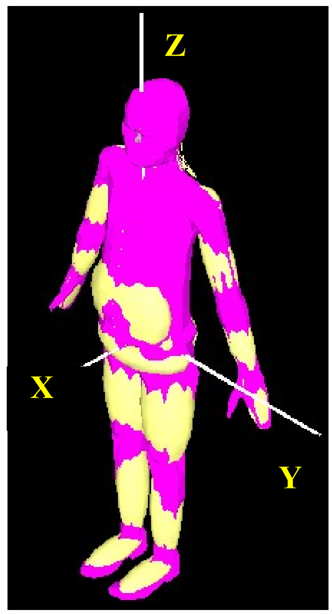


Figure A3.1: Co-ordinate System of Human Model

Appendix 4
Naval Bio-dynamic Laboratory
Sled Acceleration Pulses

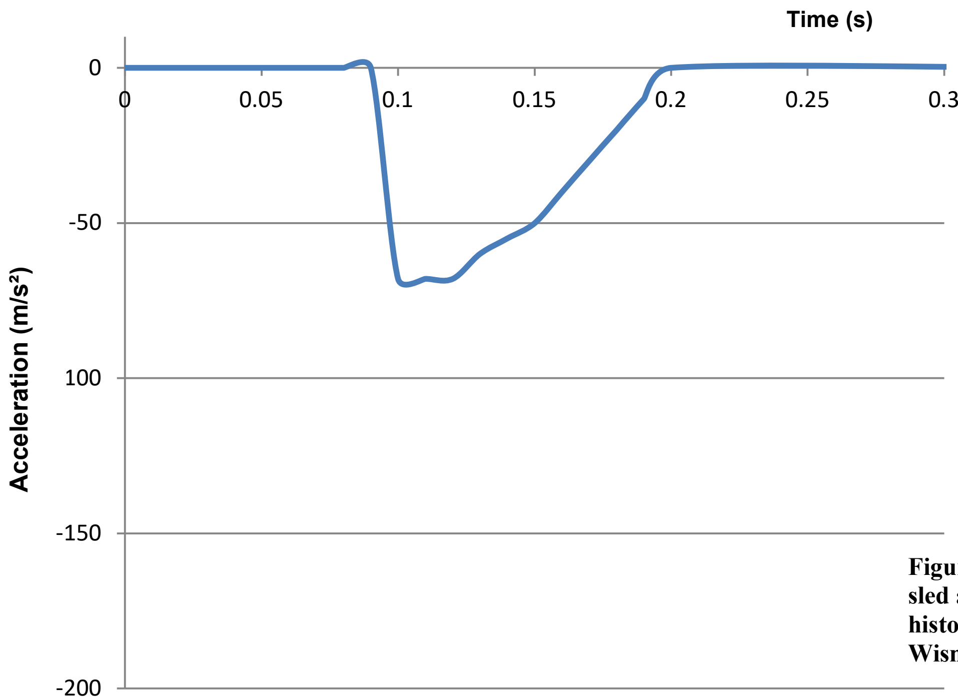


Figure A4.1: Mean lateral sled acceleration-time history, adapted from Wismans et al. (1986)

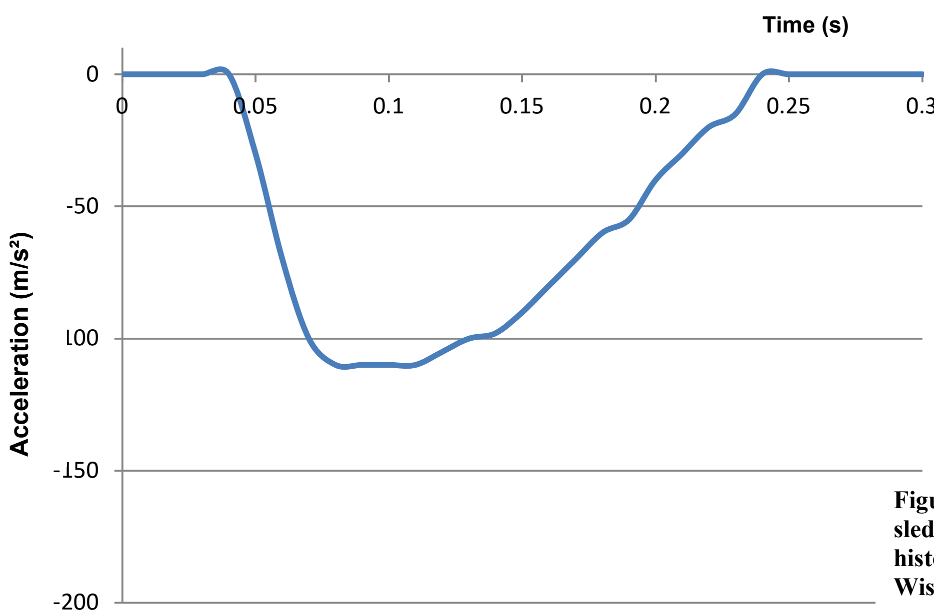


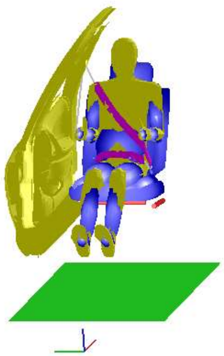
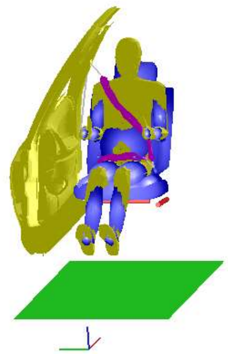
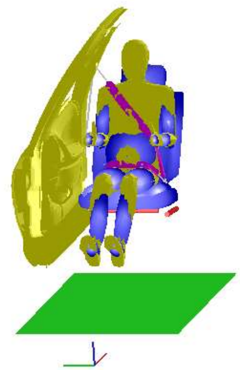
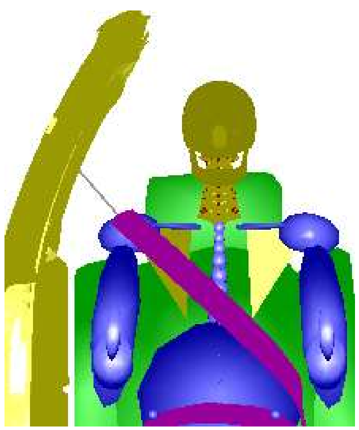
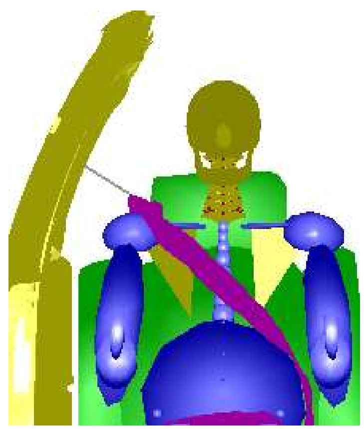
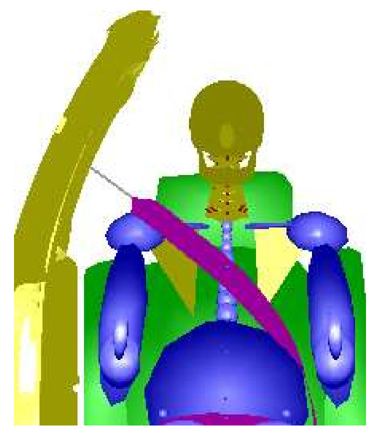
Figure A4.2: Mean oblique sled acceleration-time history, adapted from Wismans et al. (1986)

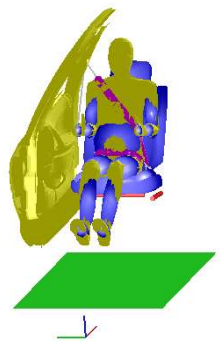
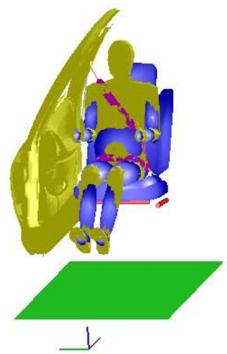
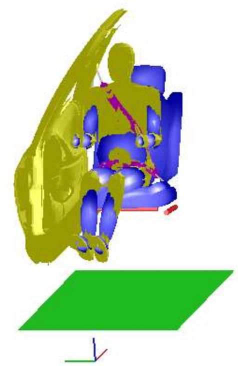
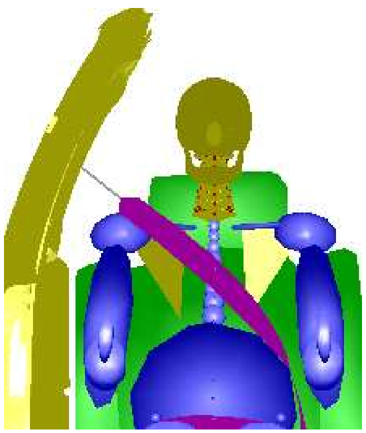
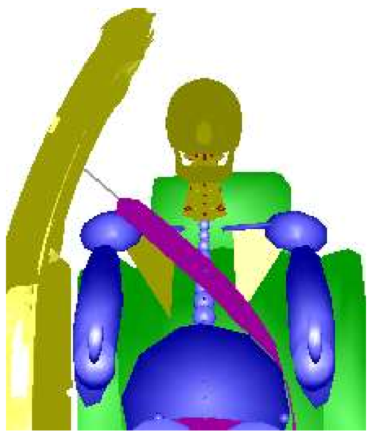
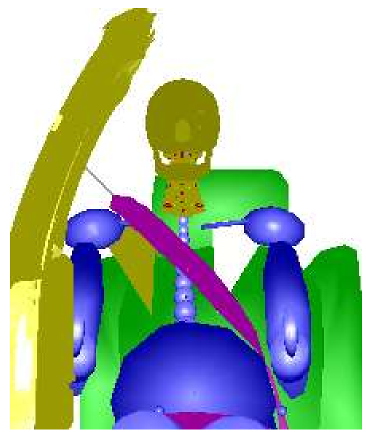
Appendix 5
Kinematic Data- Skeletal Spine View

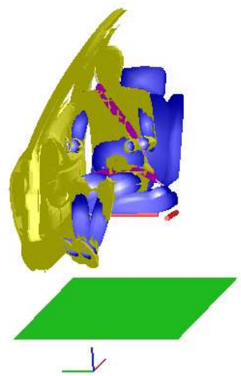
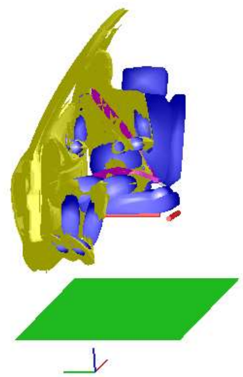
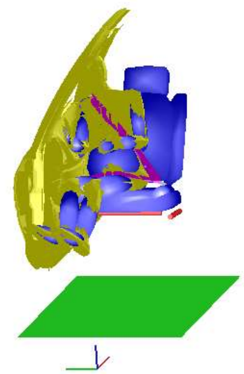
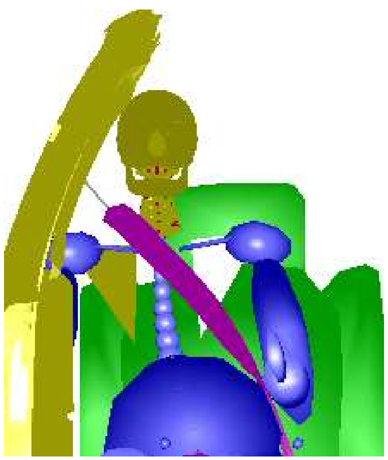
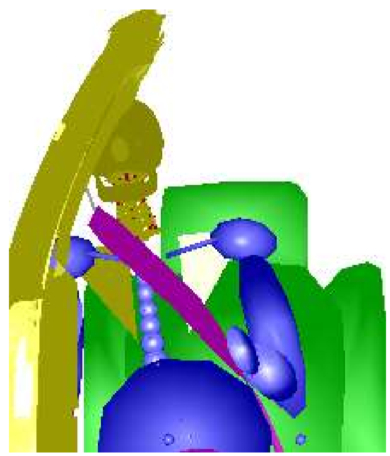
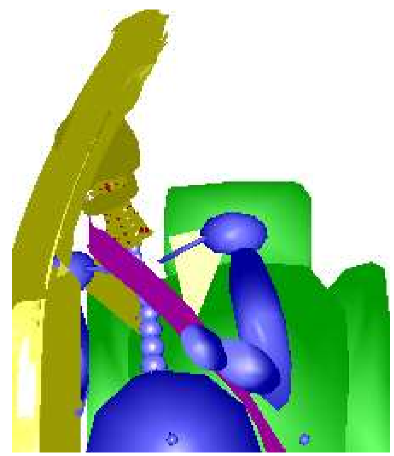
The following six figures have been presented to illustrate the kinematics of the neck with the facet skin layer displayed as transparent. This data has been presented to illustrate the lateral curvature of the cervical spine. The data is presented in three rows. The first row contains the time (ms), the second row contains the original view of the kinematic data as presented in Chapter 4. The third row is the kinematic data with the facet skin layer of the combined human model displayed as transparent. The data for each trial has been presented in a different figure, listed as:

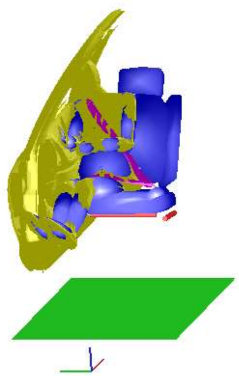
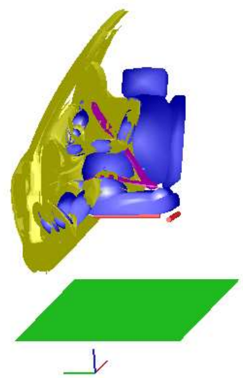
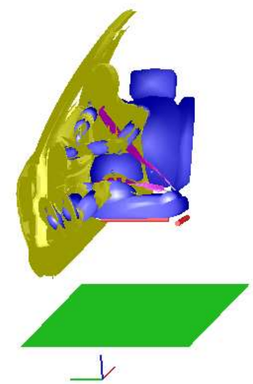
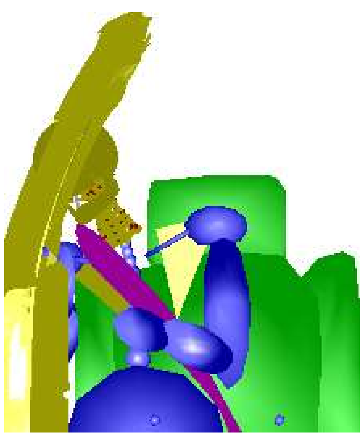
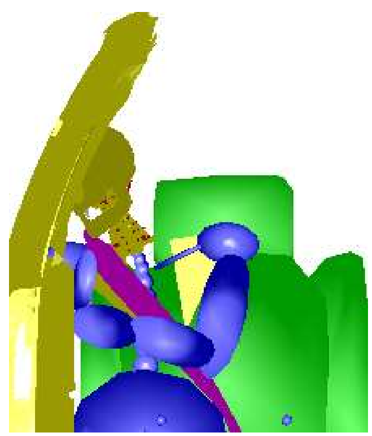
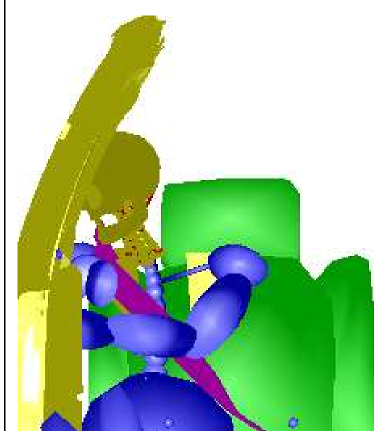
- Table A5.1: 90° lateral near side LSNI case
- Table A5.2: 90° lateral near side control case
- Table A5.3: 90° lateral far side LSNI case
- Table A5.4: 90° lateral far side control case
- Table A5.5: 45° oblique near side case
- Table A5.6: 45° oblique near side control case

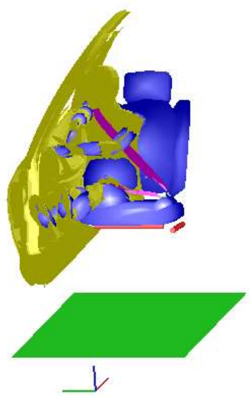
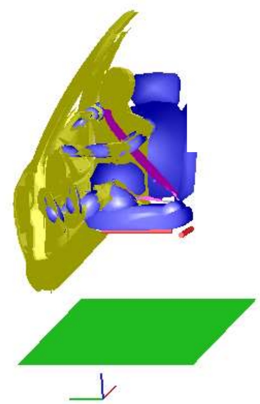
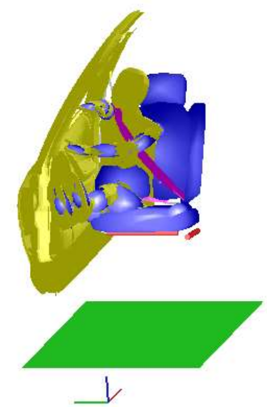
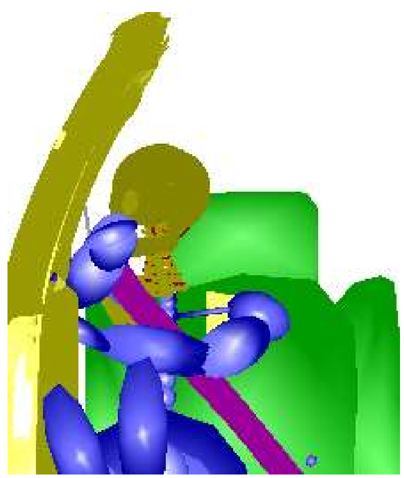
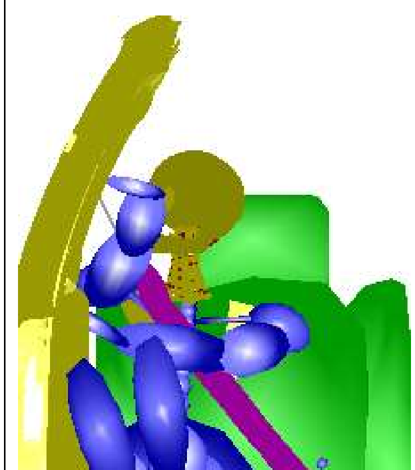
Table A5.1: 90° lateral near side LSNI case

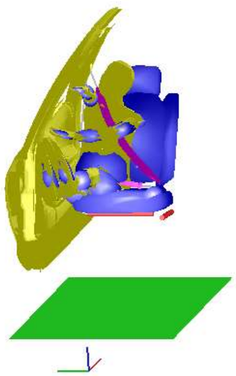
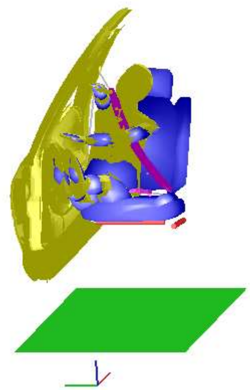
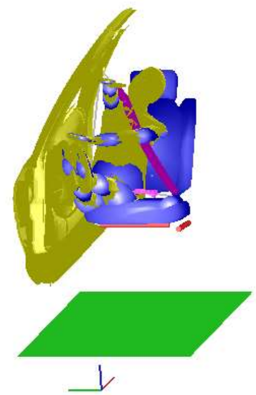
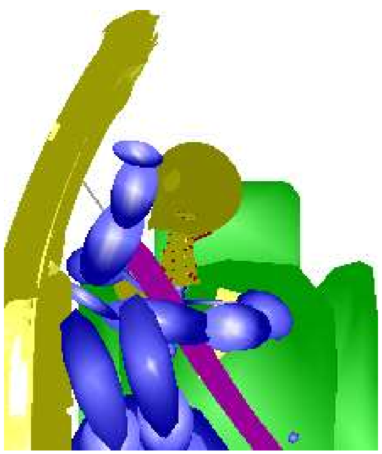
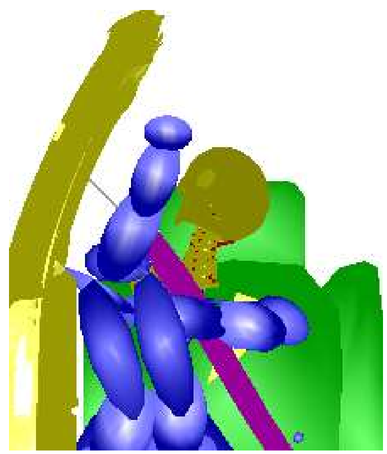
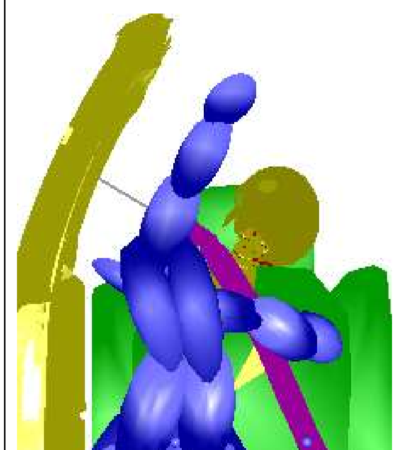
	Time:00ms	Time:20ms	Time:40ms
Original View			
View of Skeleton			

Time:60ms	Time:80ms	Time:100ms
		
		

Time:120ms	Time:140ms	Time:160ms
		
		

Time:180ms	Time:200ms	Time:220ms
		
		

Time:240ms	Time:260ms	Time:280ms
		
		

Time:300ms	Time:320ms	Time:340ms
		
		

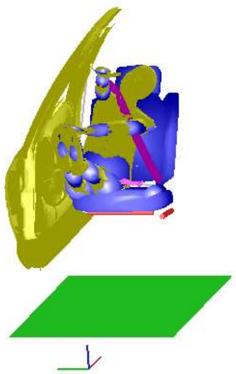
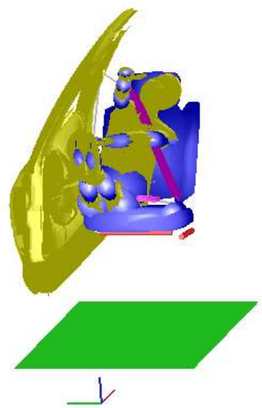
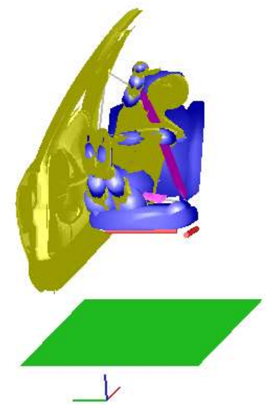
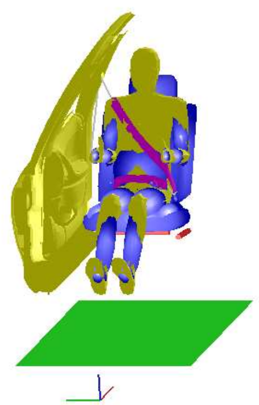
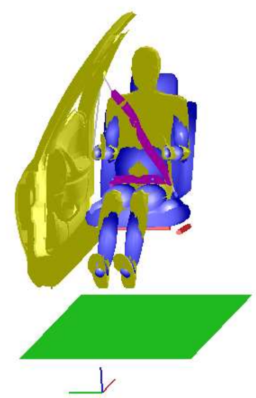
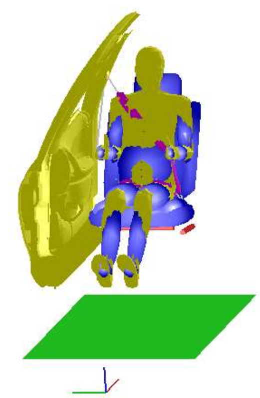
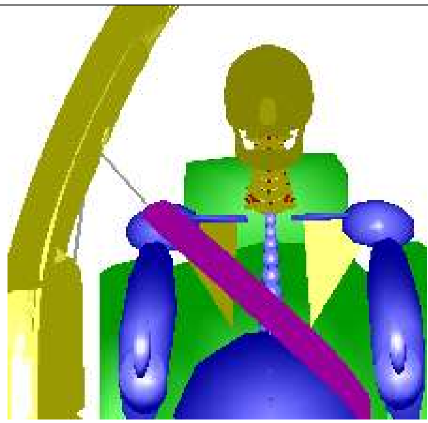
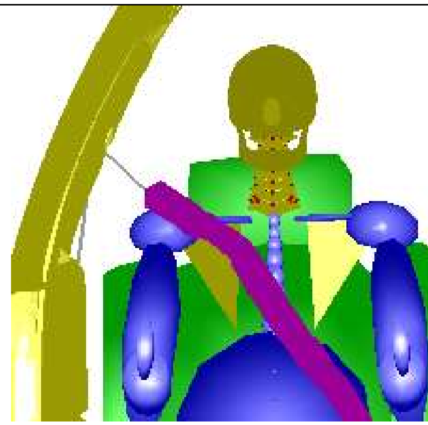
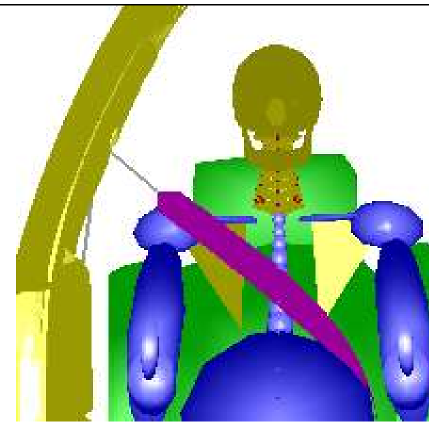
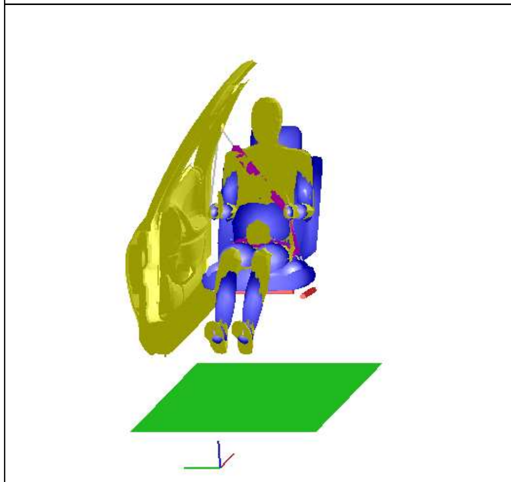
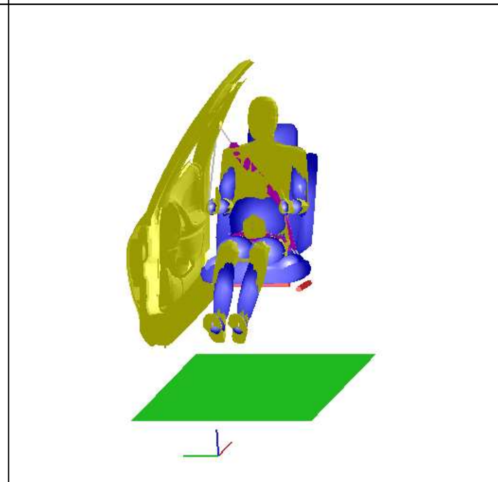
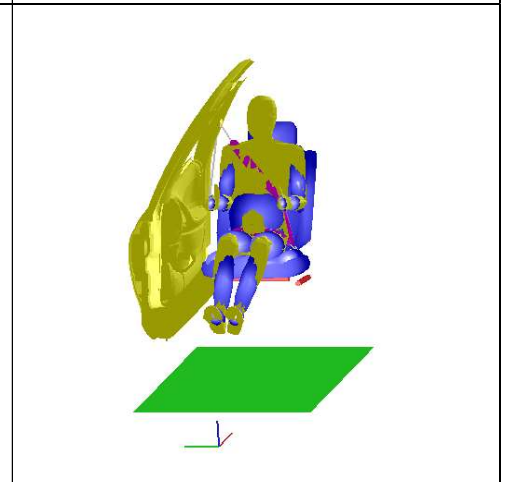
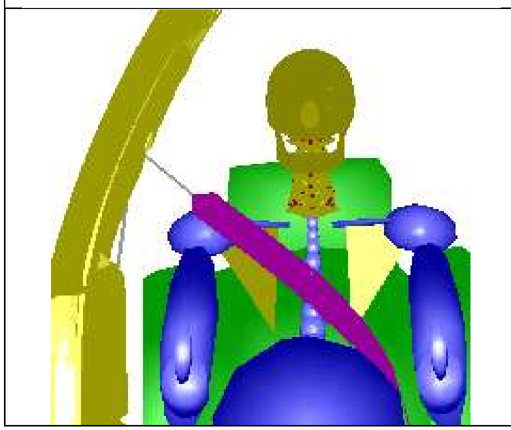
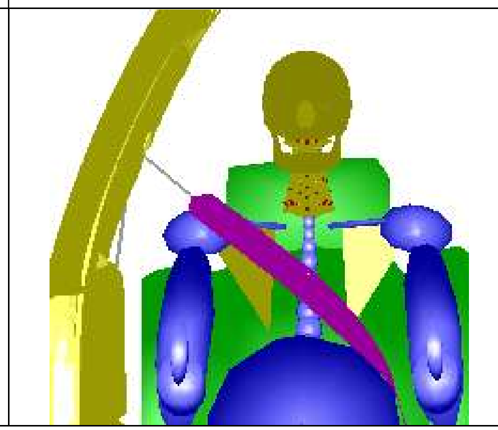
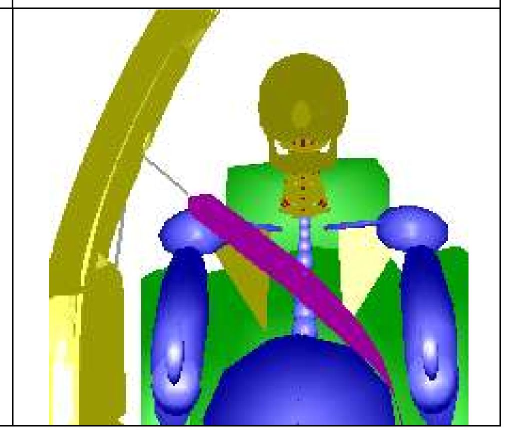
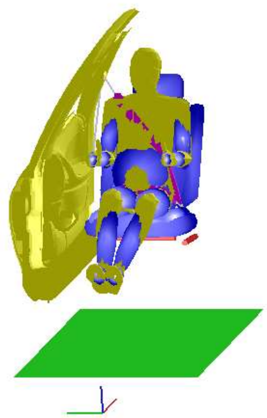
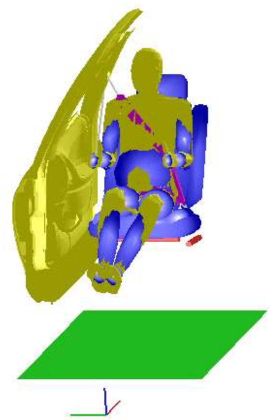
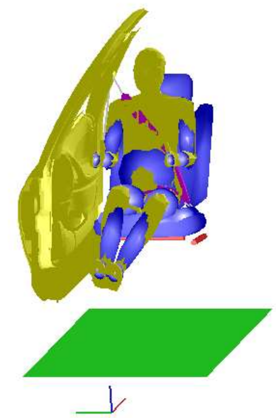
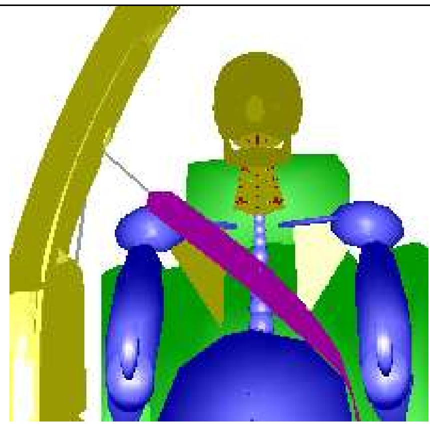
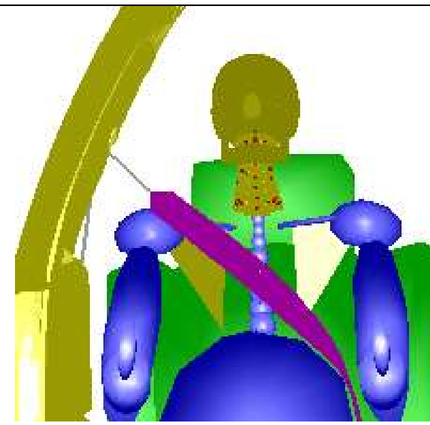
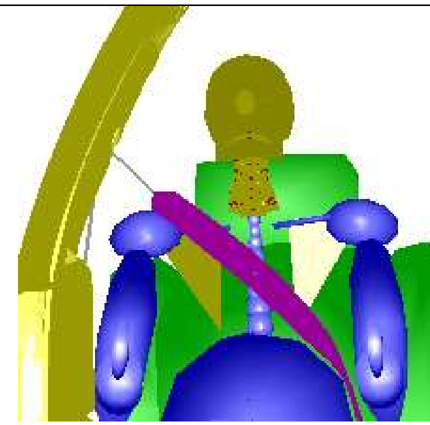
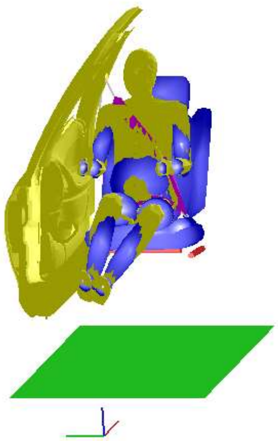
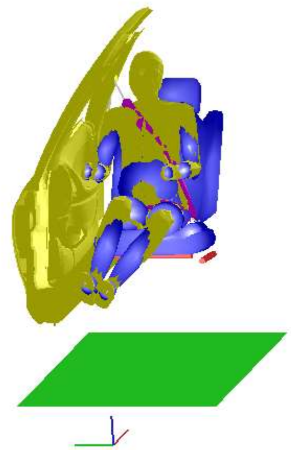
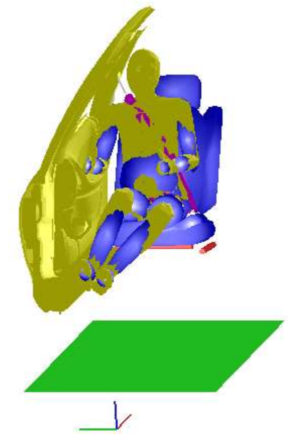
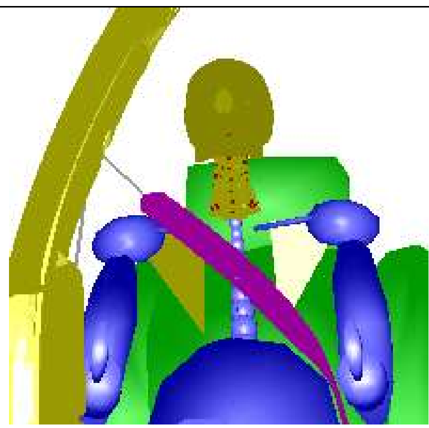
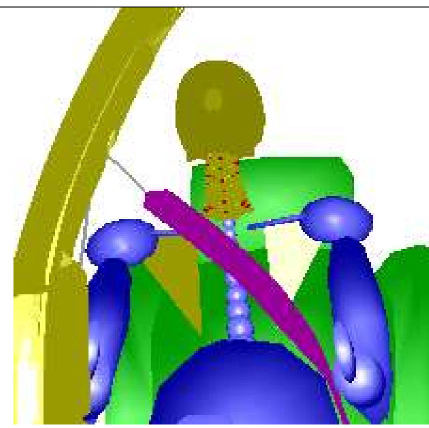
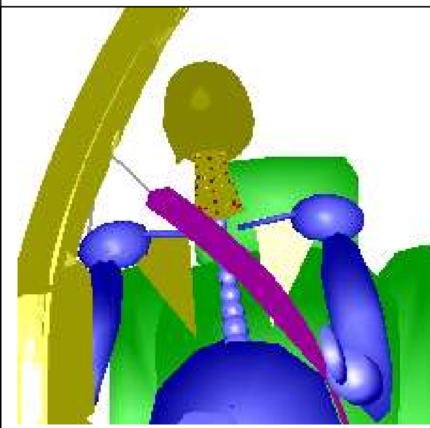
Time:360ms	Time:380ms	Time:400ms
		
		

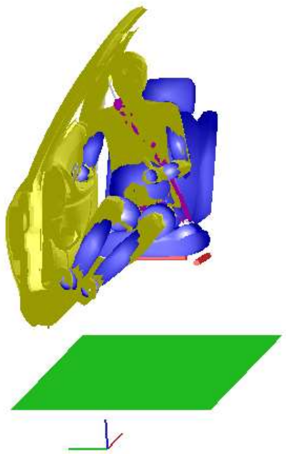
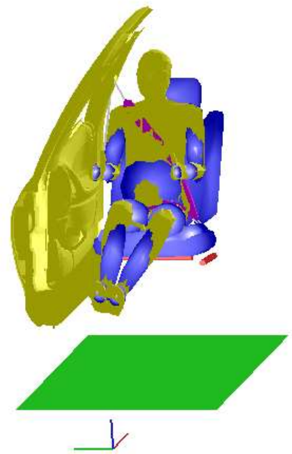
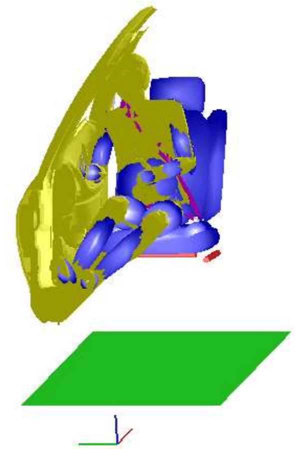
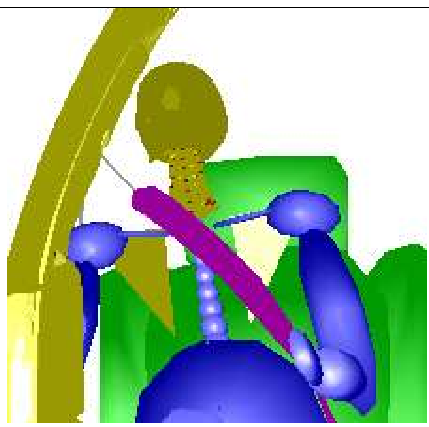
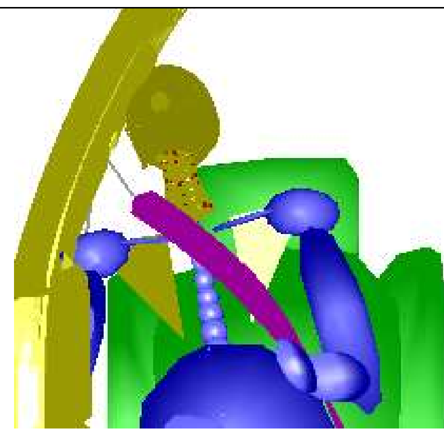
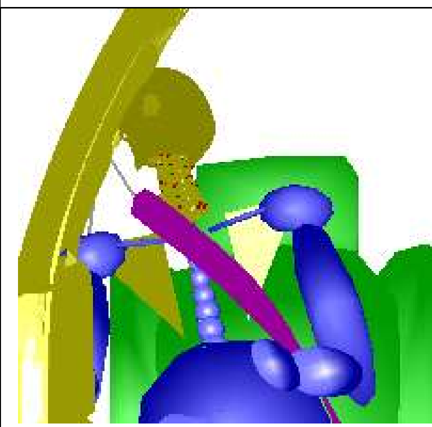
Table A5.2: 90° lateral near side control case

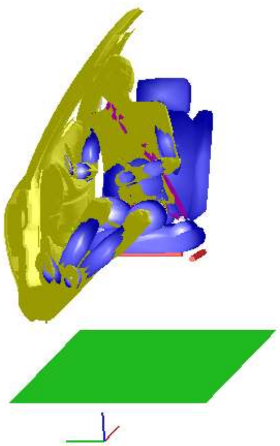
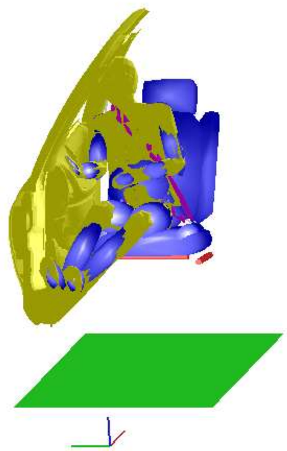
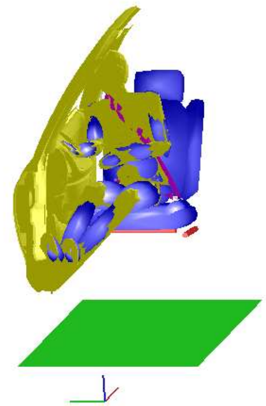
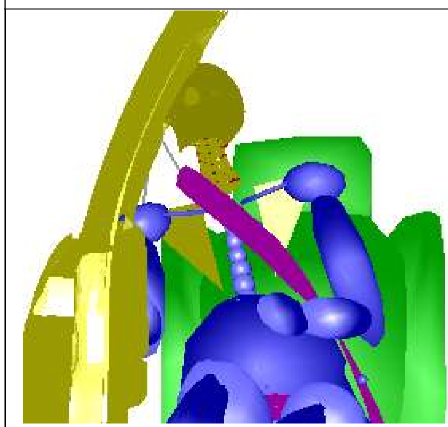
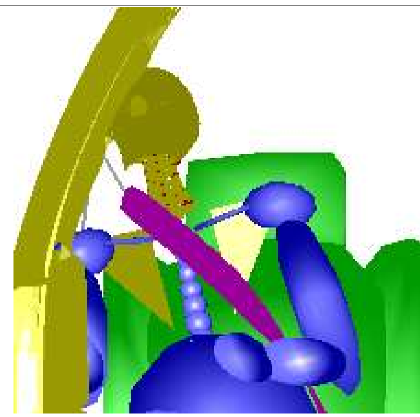
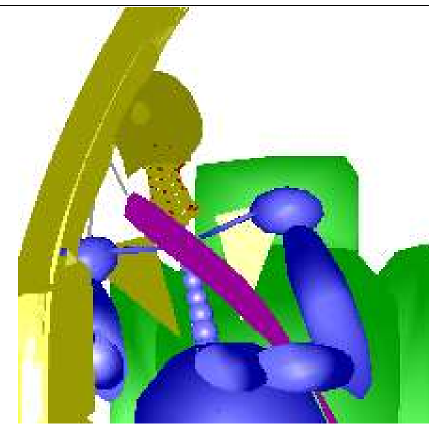
	Time:00ms	Time:20ms	Time:40ms
Original View			
View of Skeleton			

Time:60ms	Time:80ms	Time:100ms
		
		

Time:120ms	Time:140ms	Time:160ms
		
		

Time:180ms	Time:200ms	Time:220ms
		
		

Time:240ms	Time:260ms	Time:280ms
		
		

Time:300ms	Time:320ms	Time:340ms
		
		

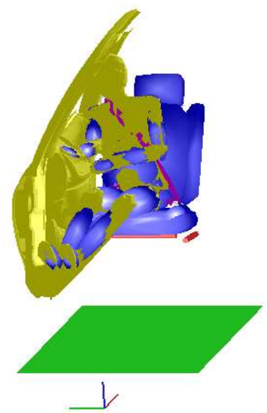
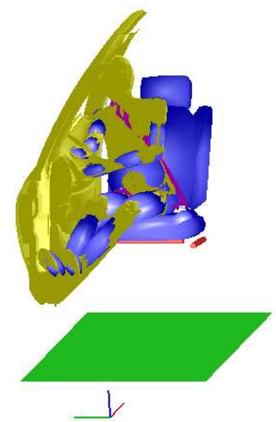
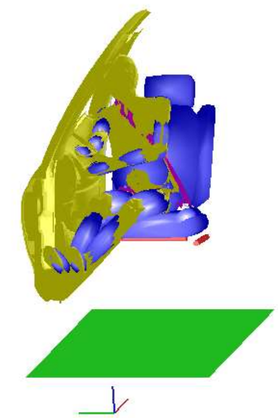
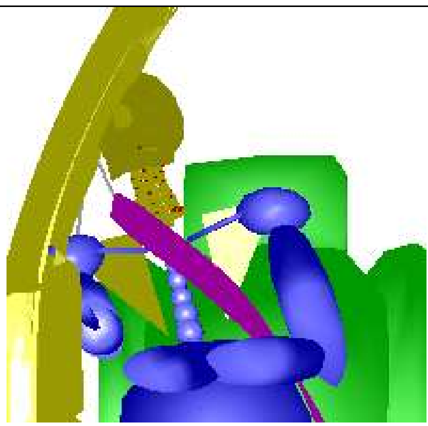
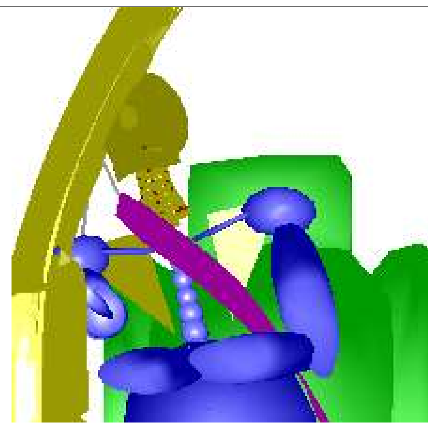
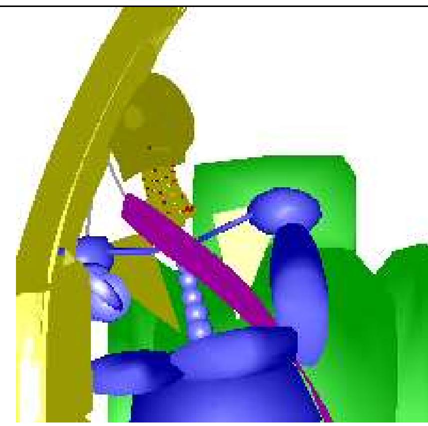
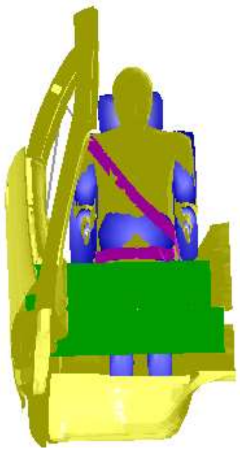
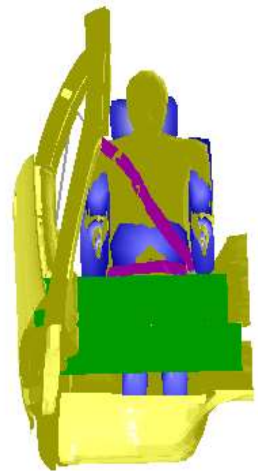
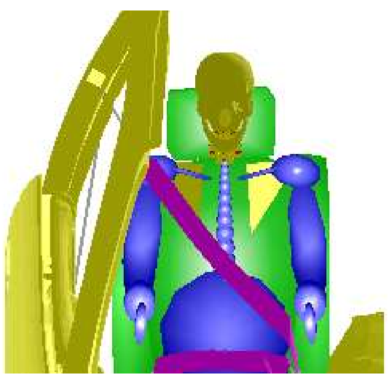
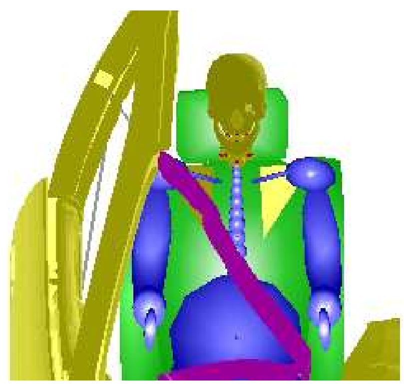
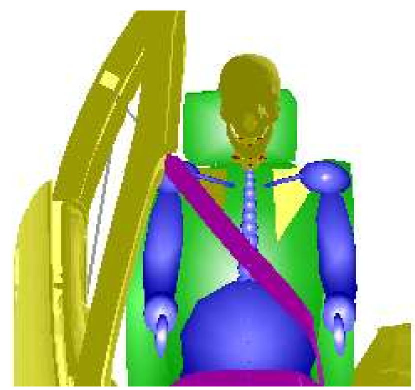
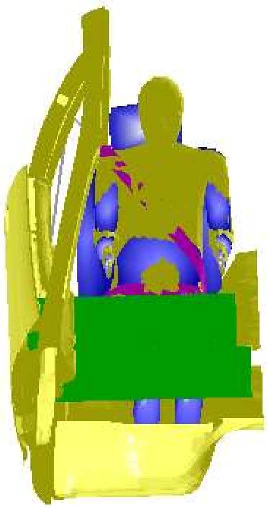
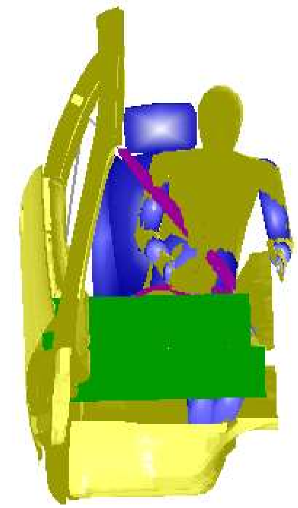
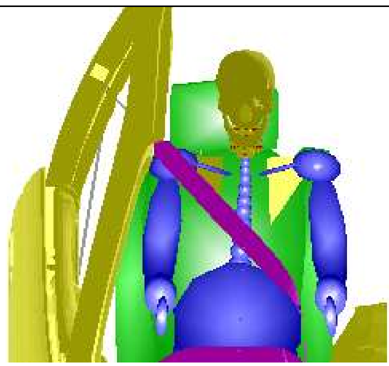
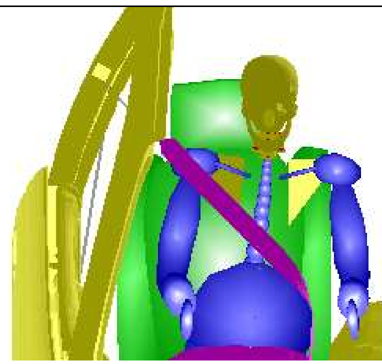
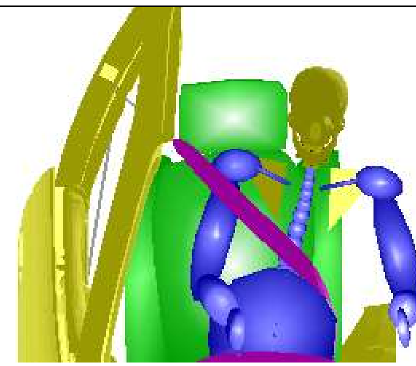
Time:360ms	Time:380ms	Time:400ms
		
		

Table A5.3: 90° lateral far side LSNI case

	Time:00ms	Time:20ms	Time:40ms
Original View			
View of Skeleton			

Time:60ms	Time:80ms	Time:100ms
		
		

Time:120ms	Time:140ms	Time:160ms

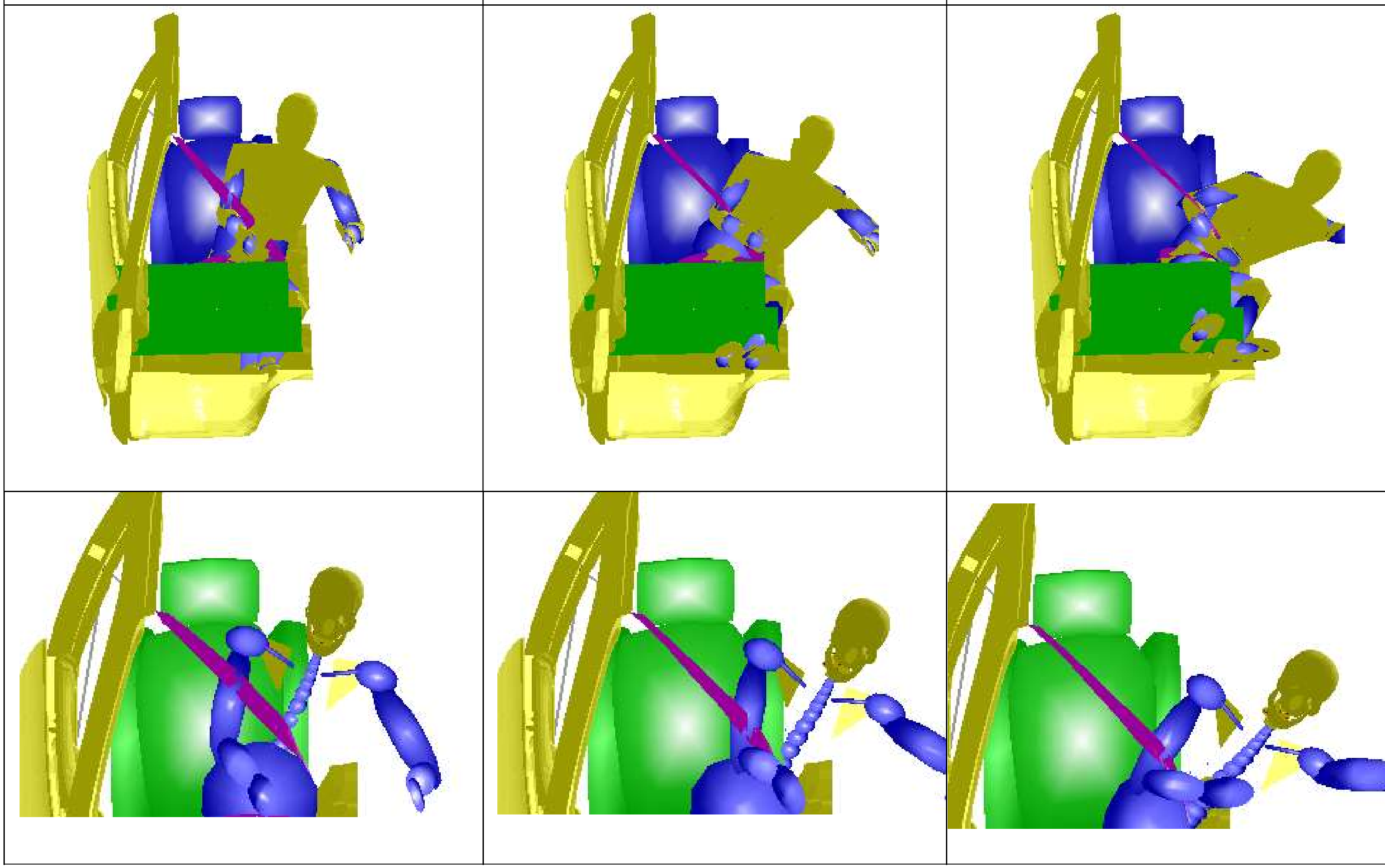
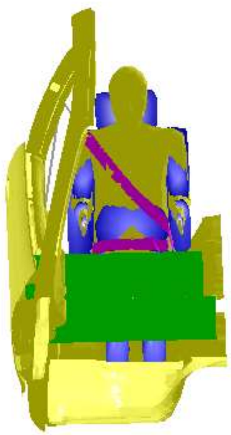
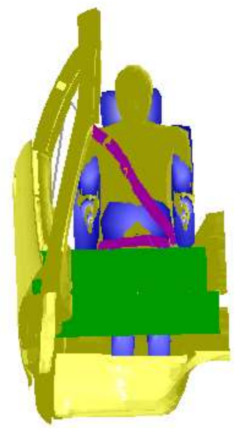
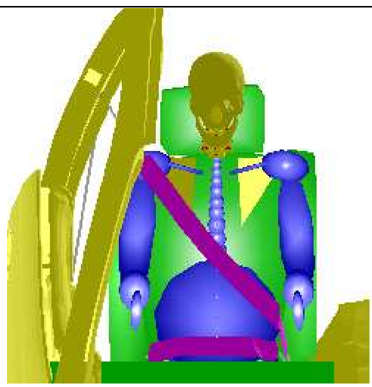
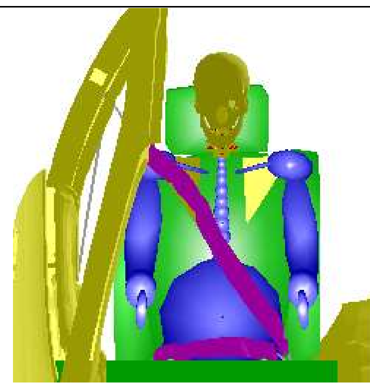
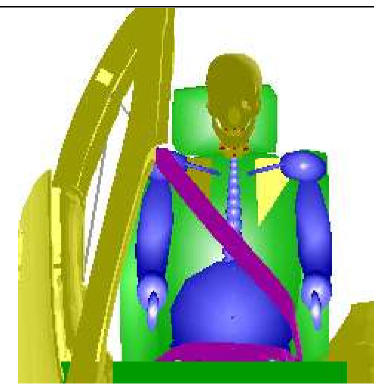
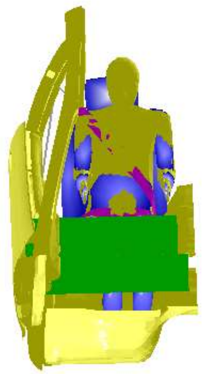
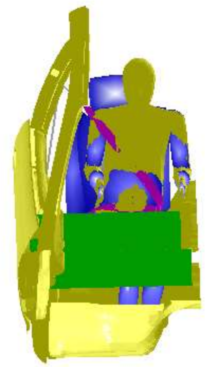
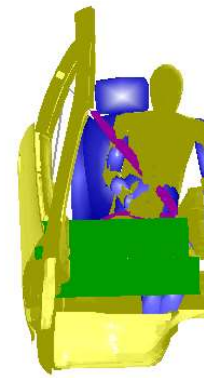
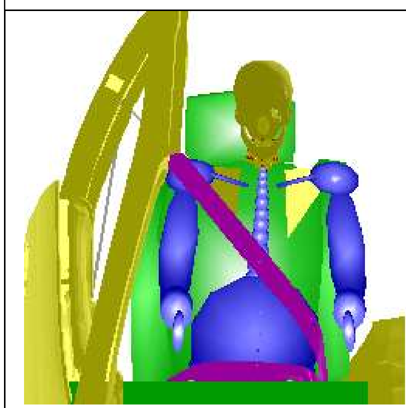
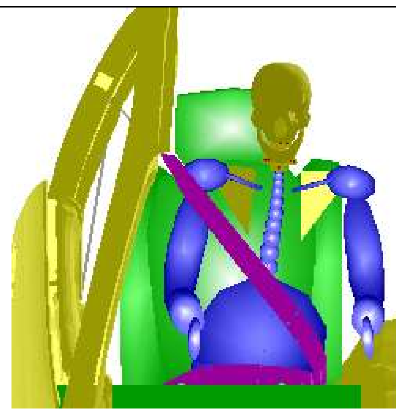
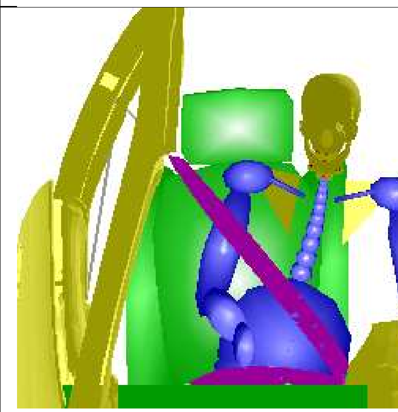


Table A5.4: 90° lateral far side control case

Time:00ms	Time:20ms	Time:40ms
		
		

Time:60ms	Time:80ms	Time:100ms
		
		

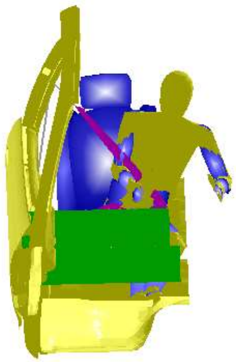
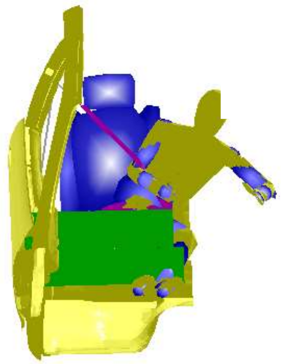
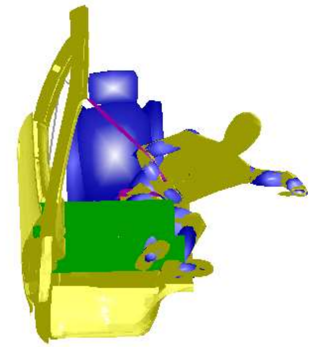
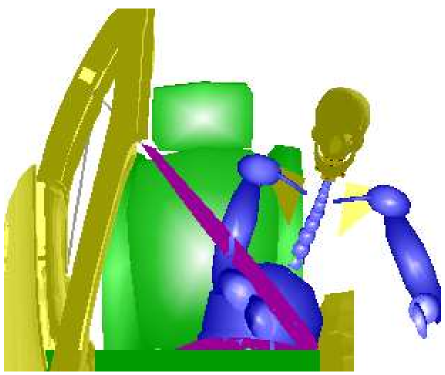
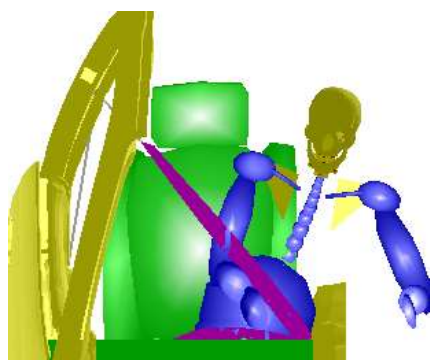
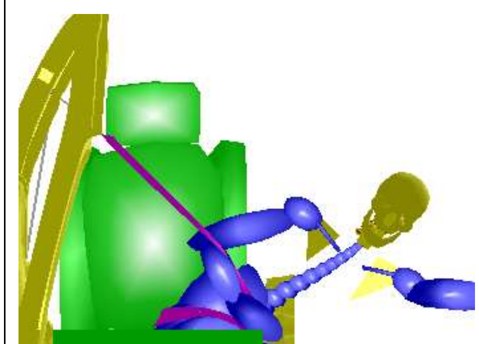
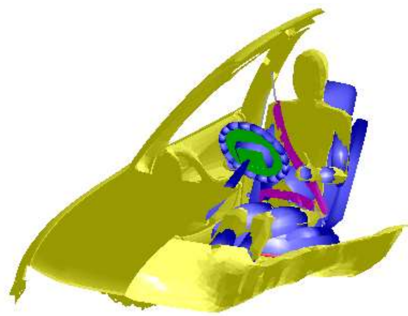
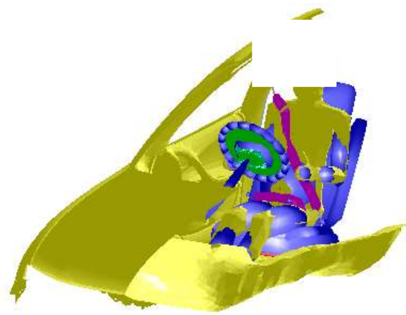
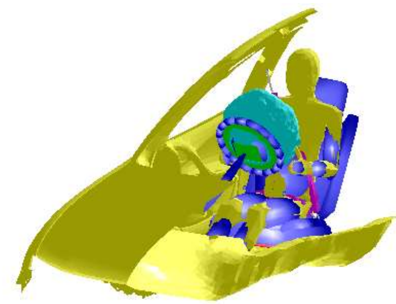
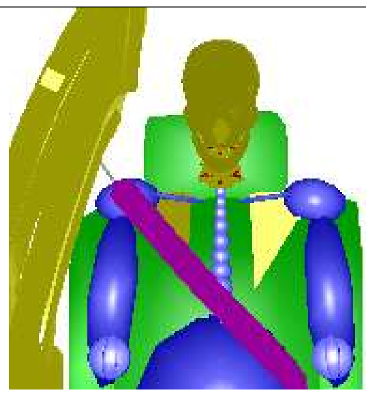
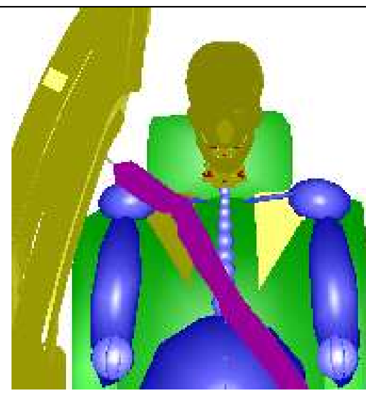
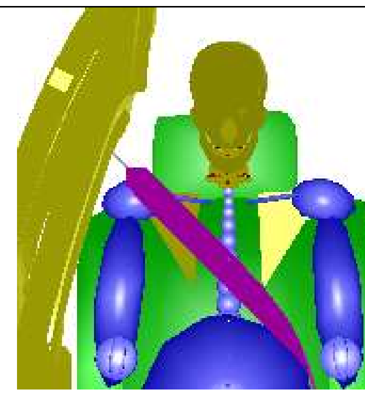
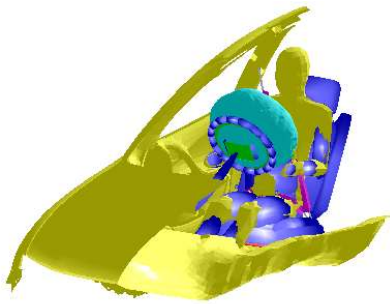
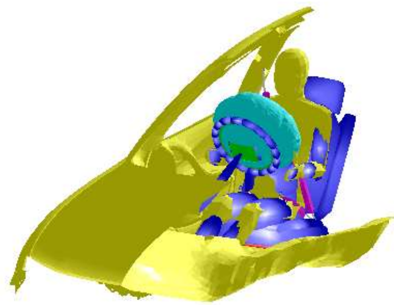
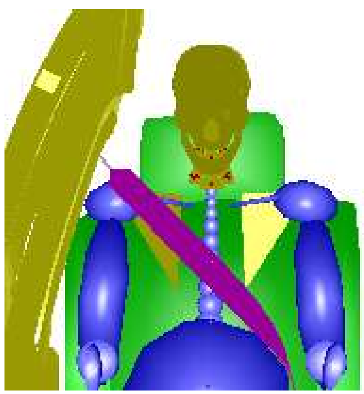
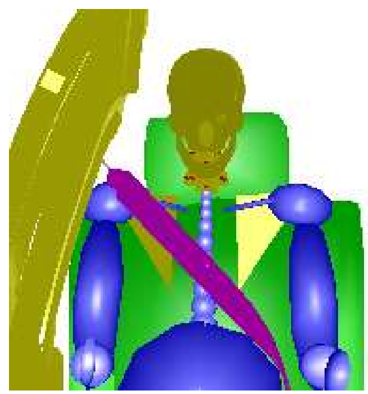
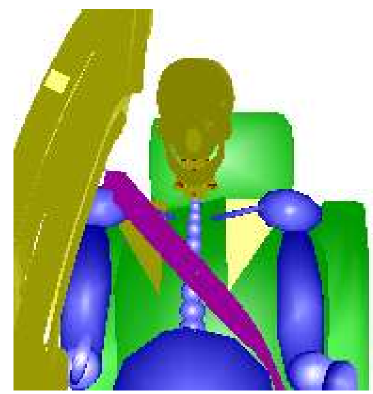
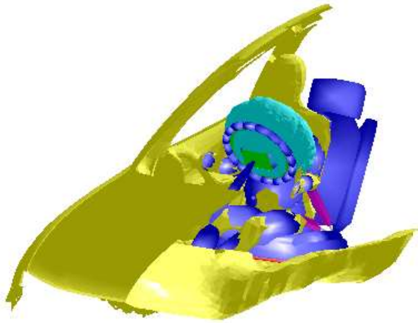
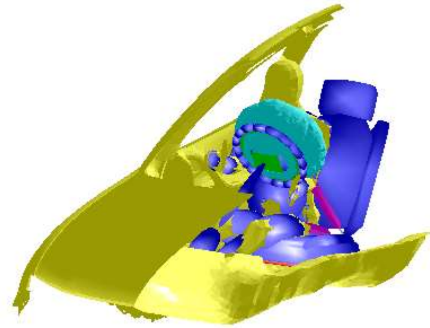
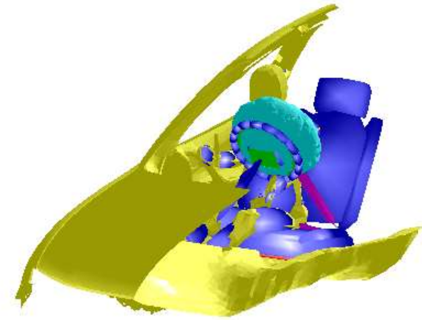
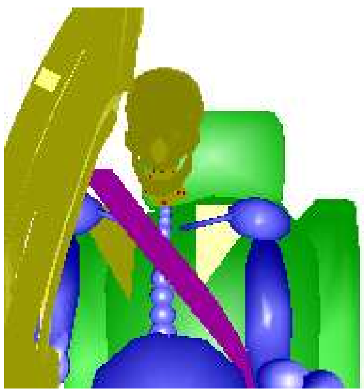
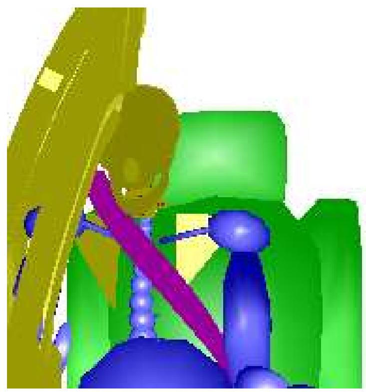
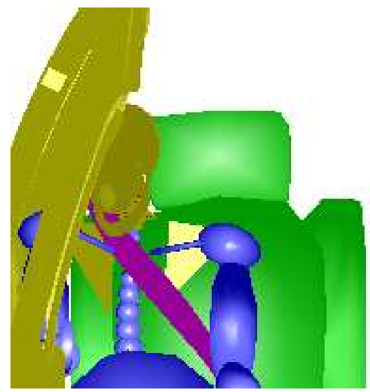
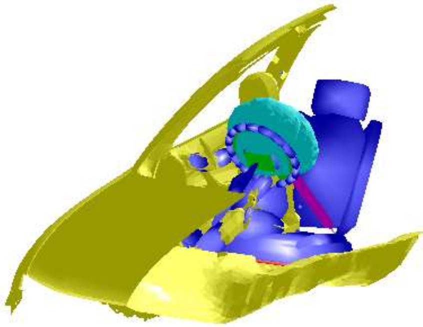
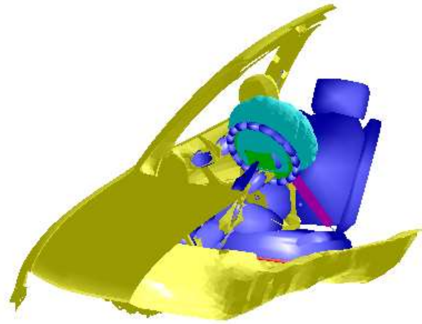
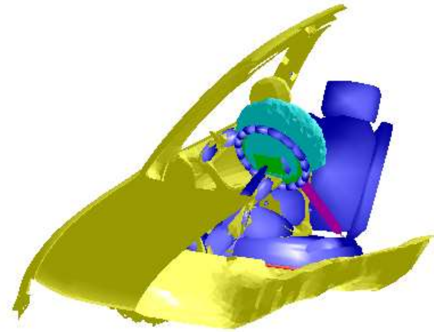
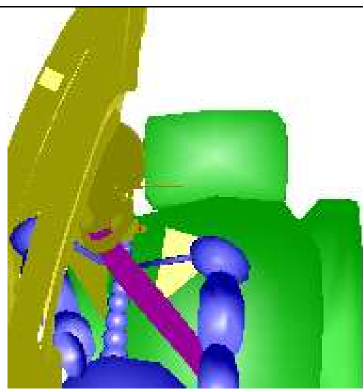
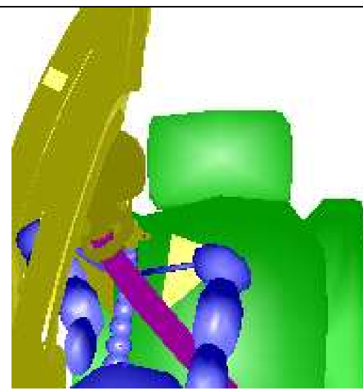
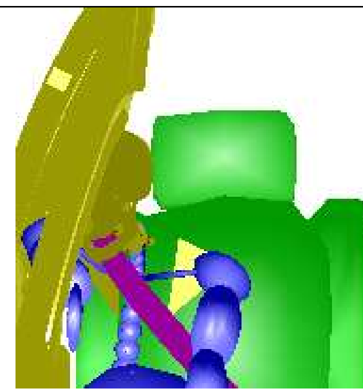
Time:120ms	Time:140ms	Time:160ms
		
		

Table A5.5: 45° oblique near side LSNI case

Time:00ms	Time:20ms	Time:40ms
		
		

Time:60ms	Time:80ms	Time:100ms
		
		

Time:120ms	Time:140ms	Time:160ms
		
		

Time:180ms	Time:200ms	Time:220ms
		
		

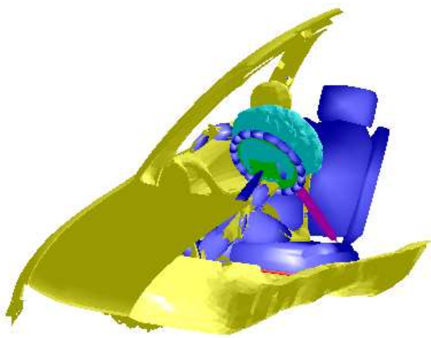
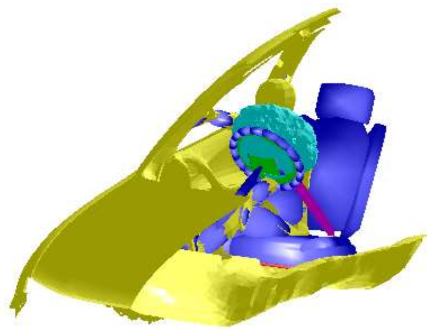
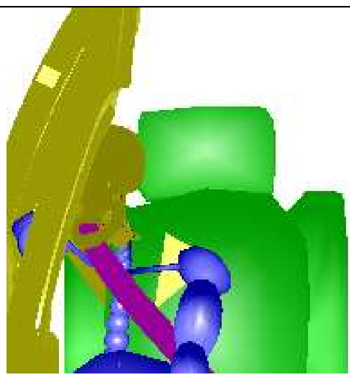
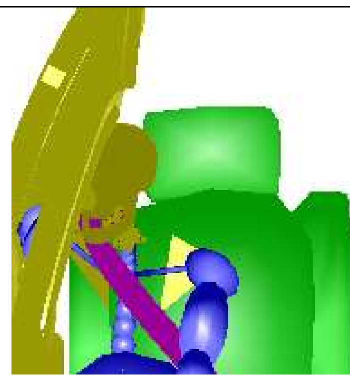
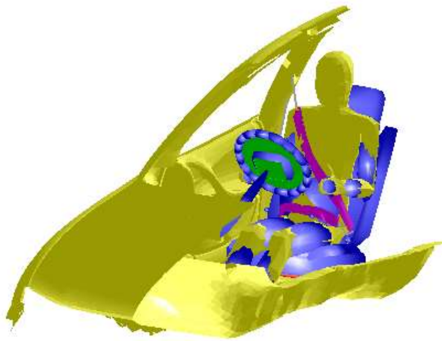
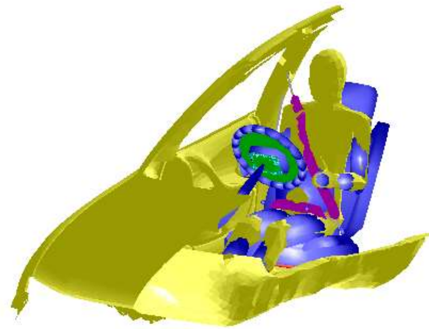
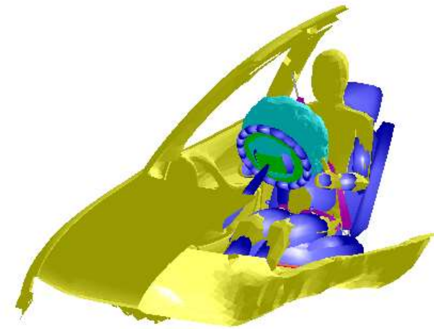
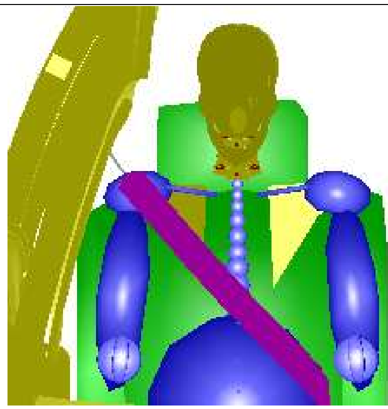
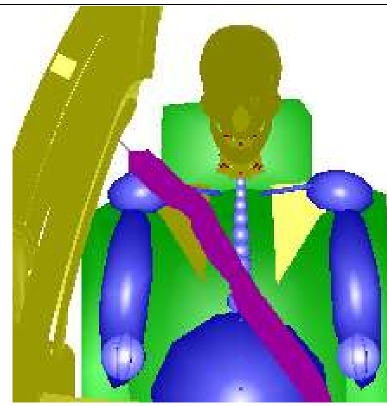
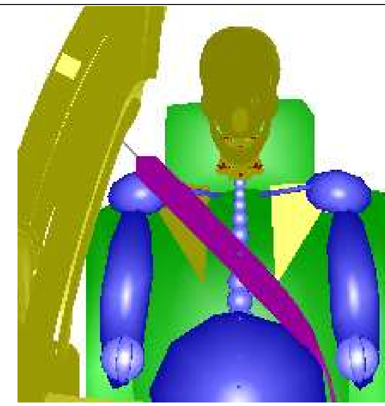
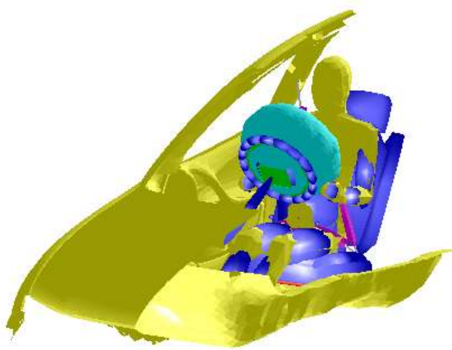
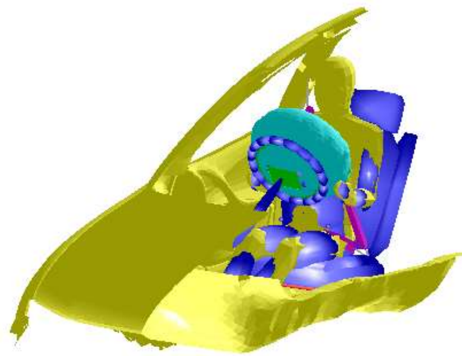
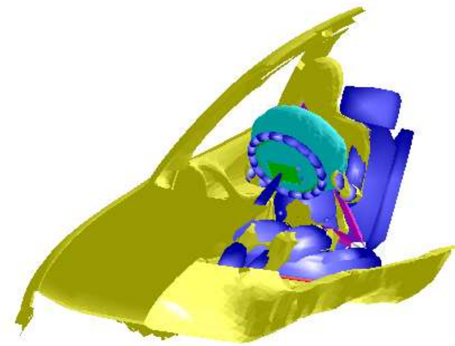
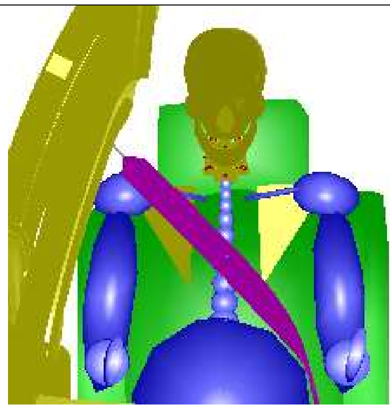
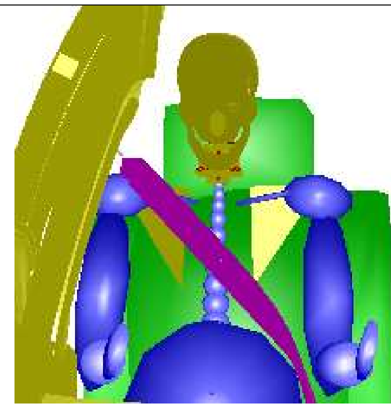
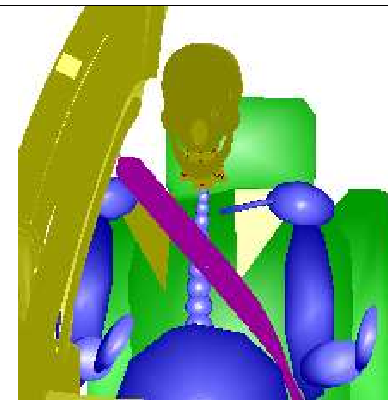
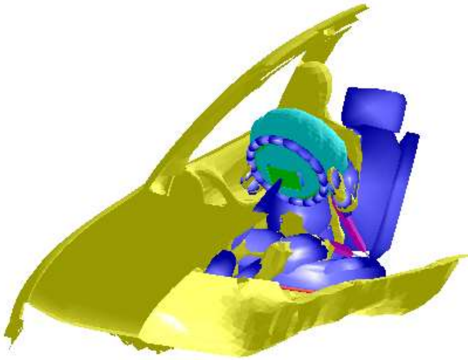
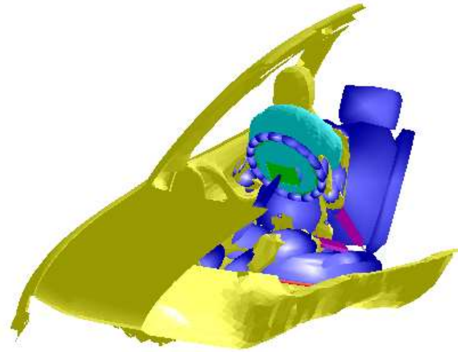
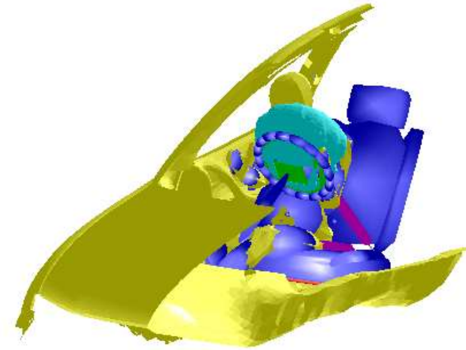
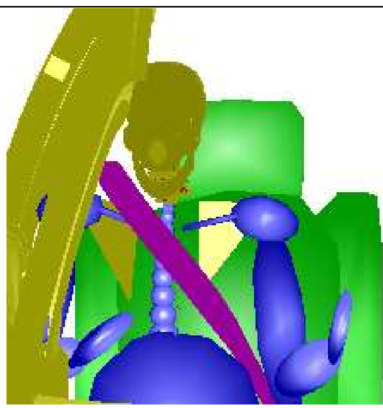
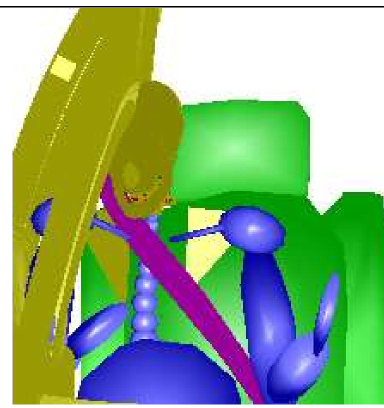
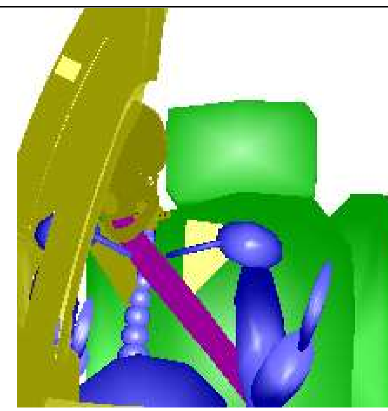
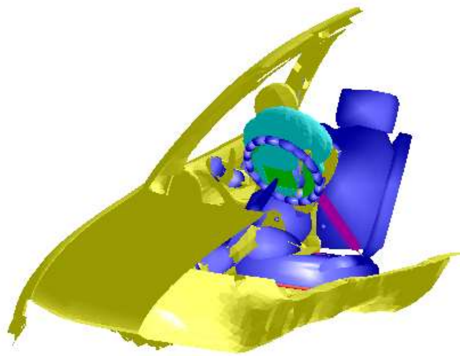
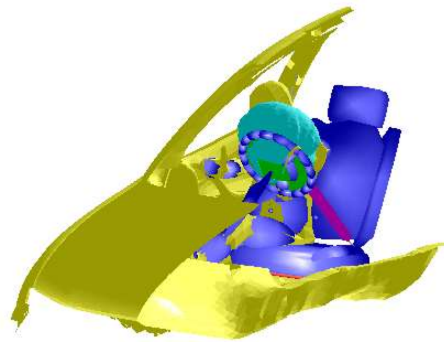
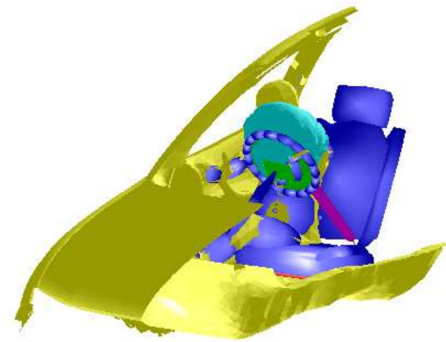
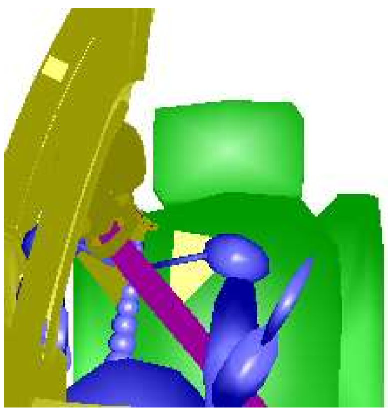
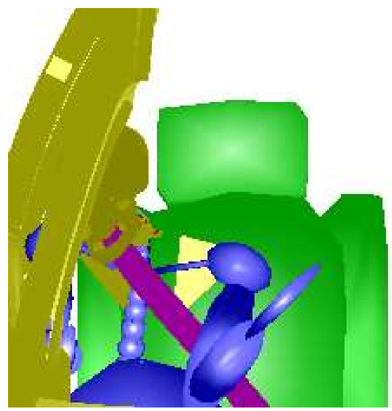
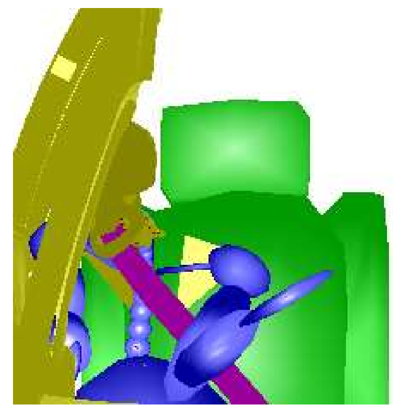
Time:240ms	Time:260ms
	
	

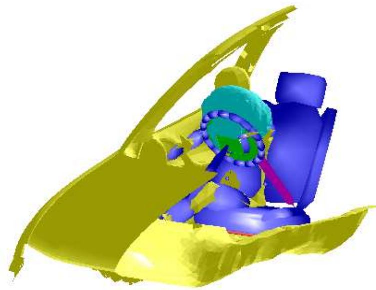
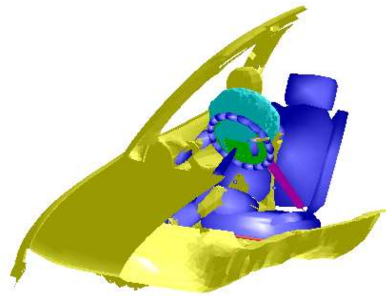
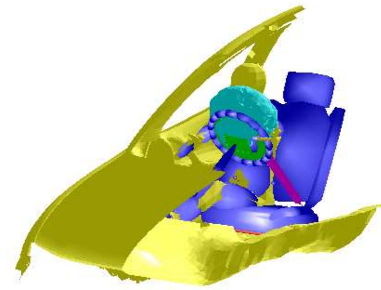
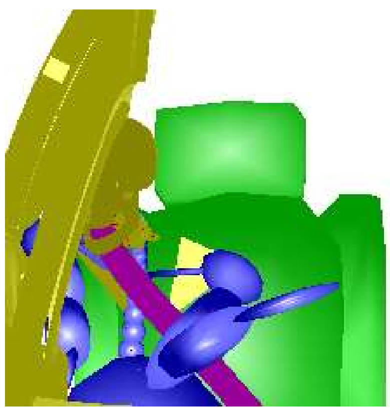
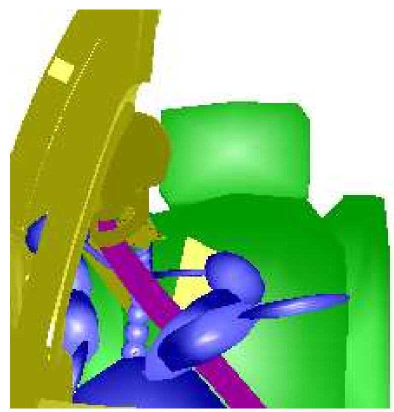
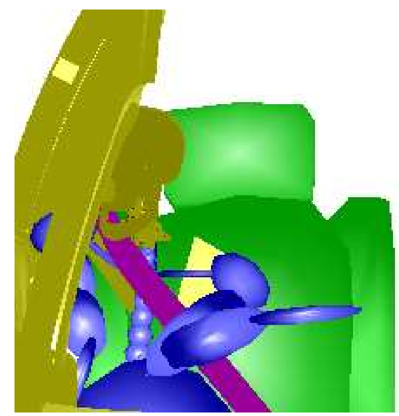
Table 5.6: 45° oblique near side control case

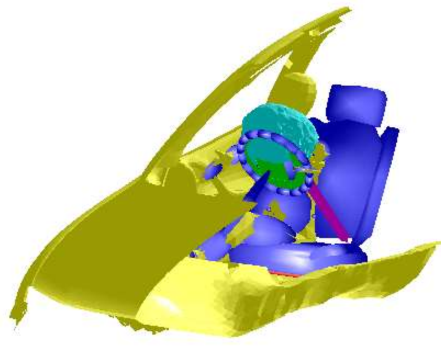
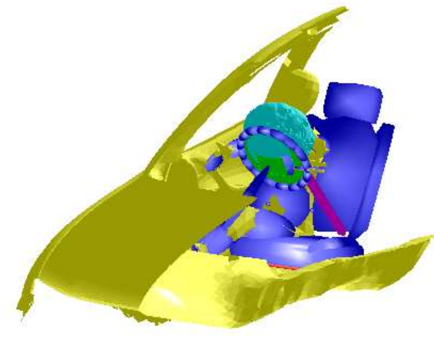
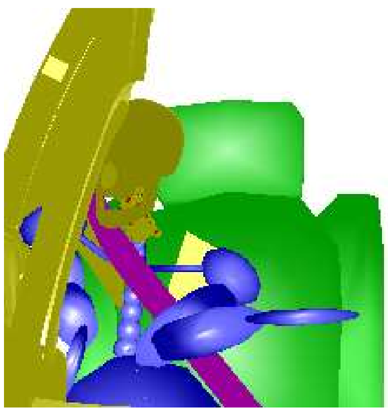
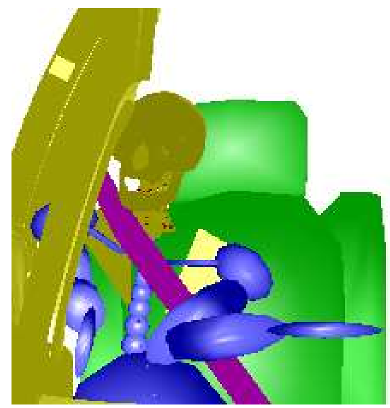
	Time:00ms	Time:20ms	Time:40ms
Original View			
View of Skeleton			

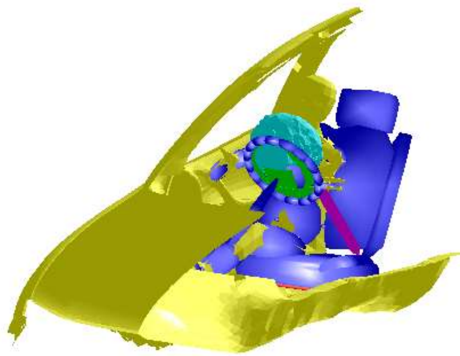
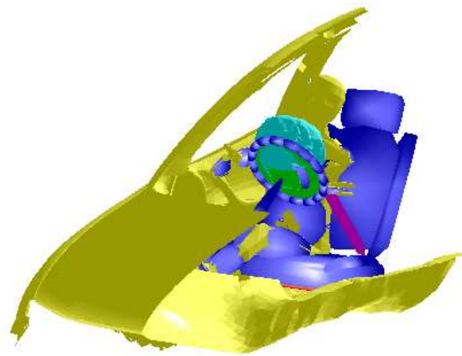
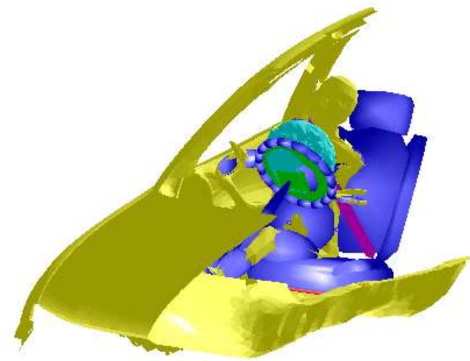
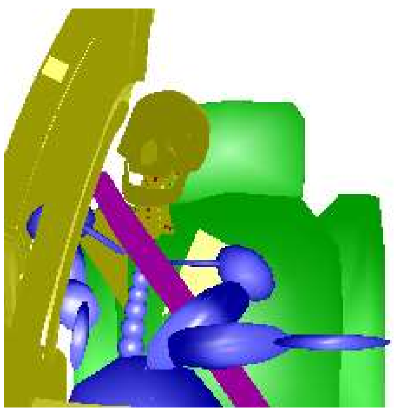
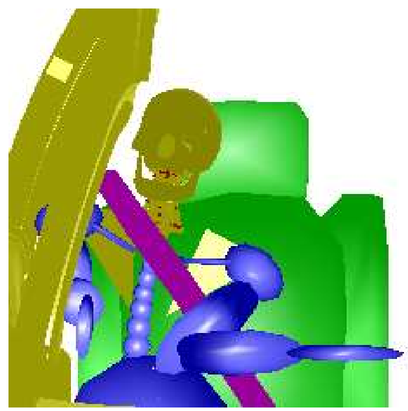
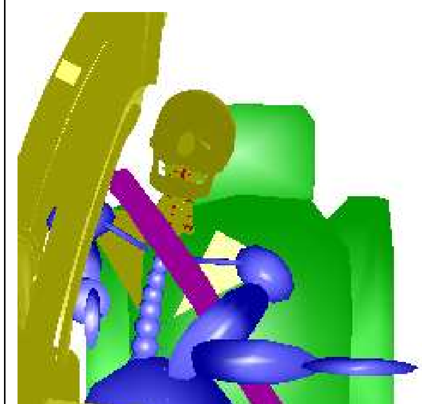
Time:60ms	Time:80ms	Time:100ms
		
		

Time:120ms	Time:140ms	Time:160ms
		
		

Time:180ms	Time:200ms	Time:220ms
		
		

Time:240ms	Time:260ms	Time:280ms
		
		

Time:300ms	Time:320ms	Time:340ms
		
		

Time:360ms	Time:380ms	Time:400ms
		
		

References

Association for the Advancement of Automotive Medicine. (1985) *The Abbreviated Injury Scale, 1985 Revision*, AAAM, Arlington Heights, Illinois.

Barnsley L., Lord S., Bogduk N., (1993) Chapter 1: The Pathophysiology of Whiplash, *Spine: State of the Art Reviews.* Edited by Robert Teasell and Allen Shapiro. Vol 7/Number 3, September. Hanley and Belfus Inc, Philadelphia.

Bendjellal F., Tarriere C., Gillet D., Mack P., Guillon F., Head and Neck Responses Under High G-Level Lateral Deceleration. 21st Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 872196.

Bertholon N., Robin S., Le-Coz J-P., Potier P., Lassau J-P., Skalli W., (2000) Human Head and Cervical Spine Behavior During Low- speed Rear End Impacts: PMHS Sled Tests with a Rigid Seat. IRCOBI Conference, Montpellier France.

Bring G., Björnstig U., Westman G., (1996) Gender Patterns in Minor Head and Neck Injuries: An Analysis of Casualty Register Data. Accident Analysis and Prevention. Vol 28 (3), p359-369.

Carrigan and Maitland (1998) Whiplash Injuries. *Vertebral Musculosckeletal Disorders.* Butterworth and Heinemann UK.

Cavanaugh J.M, Waliko T.J, Maihorta A, Zhu Y, King A.I, (1990) Biomechanical Response and Injury Tolerance of the Thorax in Twelve Sled Side Impacts, 34th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 902307.

Cavanaugh JM, Zhu Y, Huang Y, et al. 1993. Injury and Response of the Thorax in Side Impact Cadaveric Tests. In Proceedings of the 37th Stapp Car Crash Conference, Warrendale, Society of Automotive Engineers, Paper no. 933127.

Chen H., Yang K.H., Wang Z. (2009) Biomechanics of whiplash injury, Chinese Journal of Traumatology (English Edition), Volume 12, Issue 5, pp 305–314.

Cullen E., Stabler K., Mackay G., Parkin S., (1996) Head restraint Positioning and Occupant Safety in Rear impact: The Case for Smart Restraints. IRCOBI Conference, Dublin, September.

Croft A.C, Randall Eldridge T.R. (2011) Human subject rear passenger symptom response to frontal car-to-car low-speed crash tests, Journal of Chiropractic Medicine, Volume 10, Issue 3, pp 141–146.

Davidson J., Deutscher C., Hell W., Linder A., Lövsund P., Svensson M., (1998) Human Volunteer Kinematics in Rear-End Sled Collisions. IRCOBI Conference, Göteborg, September.

Day, T. (2004) "Validation of the SIMON Model for Vehicle Handling and Collision Simulation - Comparison of Results with Experiments and Other Models," Society of Automotive Engineers International Congress, SAE Technical Paper 2004-01-1207.

Day, T. Roberts, S., and York, A., (2001) "SIMON: A New Vehicle Simulation Model for Vehicle Design and Safety Research," Society of Automotive Engineers Technical Paper 2001-01-0503, SAE 2001 World Congress.

Day, T. and Hargens, R., (1989) "Further Validation of EDCRASH Using the RICSAC Staged Collisions," Society of Automotive Engineers Technical Paper 890740, pp 139- 154.

Day, T. and Siddall, D., (1996) "Validation of Several Reconstruction and Simulation Models in the HVE Scientific Visualization Environment," Society of Automotive Engineers International Congress, Detroit February, SAE Technical Paper 960891 pp 221- 230.

Day, T. and Hargens, R., (1990) "Further Validation of EDSMAC Using the RICSAC Staged Collisions," Society of Automotive Engineers International Congress, SAE Technical Paper 900102, pp39-55.

Dehnera C, Schickb S, Krausa M, Hell W, Kramer M, (2013), Muscle Activity Influence on the Kinematics of the Cervical Spine in Rear-End Sled Tests in Female Volunteers, Traffic Injury Prevention, Volume 14, Issue 4, 2013 pp 369-377.

Deng B., Begeman P.C., Yang K.H., Tashman S., King A.I. (2000) Kinematics of Human Cadaver Cervical Spine Low Speed Rear-End Impacts, 44th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 00S-014.

Desapriya E.B.R, Pike I., Brussoni M. Han G. (2004) The Injury Severity Rate Differences in Passenger Cars and Pick Up Trucks Related Two Vehicle Involved Motor Vehicle Crashes in British Columbia, Canada, IATSS Research, Volume 28, Issue 2, pp 42–47

Dvorak J., Antinnes J., Panjabi M., Loustalot D., Bonamo M., (1992) Age and Gender Related Normal Motion of the Cervical Spine. Spine, 17: 393-398.

Evans R.W. (1992) Some Observations on Whiplash Injuries, Neurologic Clinics, Vol 10, No 4, November.

Ewing C.L., Thomas D.J., Lustik K.L. Muzzy W.H., Becker E., Willems G.C., (1975) The Effect of the Initial Position of the head and Neck on the Dynamics Response of the Human Head and Neck to –Gx Impact Acceleration, 19th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 751157.

Ewing C.L., Thomas D.J., Lustik K.L. Muzzy W.H., Willems G.C., Majewski P., Dynamic Response of the Human Head and Neck to +Gy Impact Acceleration , 21st Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 770928.

Faerber E. (1982) Interaction of Car Passengers in Frontal, Side and Rear Collisions, 26th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 821167.

Farmer C.M., Wells J., Werner J.V. (1999) Relationship of Head Restraint Positioning to Driver Neck Injury in Rear-end Crashes. Accident Analysis and Prevention. Vol 31 p 719-728.

Fildes B., Vulcan P. (1995) Neck and Spinal Injuries: Injury Outcome and Crash Characteristics in Australia. Institute of Engineers, Biomechanics Panel on Whiplash, Neck and Spinal Injuries, Adelaide.

Franklyn, M., Fildes, B., Zhang, L., King, Y., & Sparke, L. (2005). Analysis of finite element models for head injury investigation: reconstruction of four real-world impacts. *Stapp car crash journal*, *49*, 1.

Franklyn, M., Logan, D., Hillard, P., & Fildes, B. (2005). Full crash test reconstruction and analysis of four real-world impacts. In *Australasian Road Safety Research Policing Education Conference, 2005, Wellington, New Zealand*.

Franklyn, M., Fildes, B., Dwarampudi, R., Zhang, L., Yang, K., Sparke, L., & Eppinger, R. (2003, May). Analysis of computer models for head injury investigation. In *Proceedings of the 18th International Technical Conference on Enhanced Safety Vehicles*.

Galasko C.S.B., Murray P.M., Pitcher M., Chambers H., Mansfield S., Madden M., Jordon C., Kinsella A., Hodson M., (1993) Neck Sprains after Road Traffic Accidents: A Modern Epidemic, *Injury*, Vol 24(3). p155-157.

Gauccione S. J., Kaminski J. (2001) Human Head-Neck Kinematic Response to Impact Acceleration: Comparison of Oblique to Combined Frontal and Lateral Response. 17th International Conference on the Enhanced Safety of Vehicles. Amsterdam.

Gibson T., Bogduk N., MacPherson J., McIntosh A., (2000) Crash Characteristics of Whiplash Associated Chronic Neck Pain, Journal of Musculoskeletal Pain, Vol. 8 (1/2).

Gray H., Gray's Anatomy: A Facsimile (2001), Taj Books, Surrey, United Kingdom.

Gross J.M., Fetto J., Rosen E., (2009) Musculoskeltal Examination, 3rd Edition, Wiley-Blackwell.

Happee, R., Ridella S., Nayef A., Morsink P., de Lange R., Bours R., van Hoof J. (2000) Mathematical Human Body Models Representing a Mid Size male and a

Small Female for Frontal, Lateral and rearward Impact Loading, IRCOBI Conference, Montpellier (France), September.

.

Happee R., Hoofman M., van den Kronnenberg A.J., (1998) A Mathematical Human Body Model for Frontal and rearward Seated Automotive Impact Loading, 40th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 983150.

Harrison D.E., Harrison D.D., Cailliet R., Janik T.J., Troyanovich S.J. (2000) Cervical Coupling During Lateral Head Translations Creates a S-Configuration. *Clinical Biomechanics*, Vol. 15, pp. 436-440.

Hasija, V., Takhounts, E. G., & Ridella, S. A. (2007). Computational analysis of real world crashes: a basis for accident reconstruction methodology. In Proceedings of the 20th International Technical Conference on the Enhanced Safety of Vehicles, Lyon, France.

Hasija, V., Takhounts, E. G., & Ridella, S. A. (2009). Computerized crash reconstruction of real world crashes using optimization methodology. In Proceedings of the 21st International Technical Conference on the Enhanced Safety of Vehicles.(ESV), June (pp. 15-18).

Hell W., Langweider K., Waltz F. (1998) Reported Soft Tissue Injuries After Rear-End Car Collisions. IRCOBI Conference- Göteborg, September.

Horsch J., Schneider D.C., Kroell C.K., Raasch F.D., (1979) Response of Belt Restrained Subjects in Simulated Lateral Impact, 23rd Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 791005.

Identicar, Australia's Motor Vehicle Identification Guide, www.identicar.com.au

Insurance Institute for Highway Safety (2002) News Release, 22nd October.

Insurance Institute of Highway Safety 2007, Rear Crash Protection in Cars-Seat/head restraints in two of every three models are marginal or poor**,** IIHS News April 5**,** www.iihs.org.

Irwin, AL., Waliko, TJ., Cavanaugh, JM., Zhu, Y., King, AI. (1993), Displacement Responses of the Shoulder and Thorax in Lateral Sled Impacts, 37th Stapp Car Crash Conference, Warrendale, Society of Automotive Engineers, Paper No 933124.

Ivancic P.C. (2013) Neck injury response to direct head impact, Accident Analysis & Prevention, Volume 50, pp 323–329.

Ivancic PC1, Ito S, Tominaga Y, Carlson EJ, Rubin W, Panjabi MM. (2006) Effect of rotated head posture on dynamic vertebral artery elongation during simulated rear impact. Clinical Biomech, Volume 21(3), pp213-20.

ISO- International Organization for Standardization (2002) Road vehicles- Traffic Accident Analysis, Part 1 Vocabulary, ISO 12353-1.

de Jager M., Sauren A., Thunnissen J., (1994) A Three- Dimensional Head-neck Model: Validation for Frontal and Lateral Impacts. 38th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 942211.

de Jager (1996) Mathematical Head-neck Models for Acceleration Impacts, Ph.D. Thesis, Eindhoven University of Technology.

Jakobsson L., Lundell B., Norin H., Isaksson-Hellman I. (2000) WHIPS- Volvo's Whiplash Protection Study. Accident Analysis and Prevention Vol. 32, No 2, pp. 307- 319.

Jakobsson L., (1998)Automobile Design and Whiplash Prevention. *Whiplash Injuries: Current Concepts in Prevention, Diagnosis and Treatment of the Cervical Whiplash Syndrome.* Edited by Gunzburg R., and Szpalski M. Lippincot-Raven Publishes Philadelphia.

Jakobsson L., Norin H., Isaksson-Hellman I, (2000) Parameters Influencing the Risk of AIS 1 Neck Injuries in Frontal and Side Impacts, IRCOBI Conference, Montpellier (France), September.

Jenkins (1981) Hollinshead Functional Anatomy of the Limbs and Back. WB Saunders Company.

Jonsson B, Tingvall C, Krafft M, Bjornstig U, (2013) The risk of whiplash-induced medical impairment in rear-end impacts for males and females in driver seat compared to front passenger seat, International Association of Traffic and Safety Sciences Research, Volume 37, Issue 1, July 2013, pp 8–11.

Kallieris D., Mattern R., Schmidt G., Eppinger R., (1981) Quantifications of Side Impact responses and Injuries. 25th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, Paper No 811009.

Kallieris D., Schmidt G., (1990) Response and Injury Assesment using Cadavers and the US-SID for Far Side lateral Impacts of Rear Seat Occupants with Inboard-Anchored Shoulder Belts. 34th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, Paper No 902313.

Kallieris D., Mattern R., Schmidt G., Milter E., Stein K., (1991) Considerations for a Neck Injury Criterion. 35th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, Paper No 912916.

Kallieris, D., Rizzetti, A., Mattern, R., Thunnissen, J., and Philippens, M. (1996). Cervical human spinal loads during trauma to mechanical investigations, *Proc. IRCOBI Conf.*, pp. 89–106.

Kaneoka K., Ono K., Inami S., Hayashi K., (1999) Motion Analysis of Cervical Vertebrae During Simulated Whiplash Loading. WAD 99' Compendium/ Traffic Safety and Auto Engineering.

Kraft M. (1998) Non-Fatal Injuries to Car Occupants - Injury assessment and analysis of impacts causing short- and long-term consequences with special reference to neck injuries, Thesis, Karolinska Institutet, Stockholm, Sweden.

Kullgren, A., Krafft, M., Lie, A., & Tingvall, C. (2007, June). The effect of whiplash protection systems in real-life crashes and their correlation to consumer crash test programmes. In *Proc. 20th ESV Conf.* June*, Lyon (France)* pp. 1-7.

Kullgren A., Krafft M., Malm S., Ydendius A., Tingvall C., (2000) Influence of Airbags and Seatbelt Pretensioners on AIS 1 Neck Injuries for Belted Occupants in Frontal Impacts. 44th Stapp Car Crash Journal 44 (November 2000) 117-125.

Linder A , Schick S, Hell W, Svensson M, Carlsson A, Lemmen P, Schmitt K, Gutsche A, Tomasch E, (2013), ADSEAT – Adaptive seat to reduce neck injuries for female and male occupants, Accident Analysis & Prevention. Vol. 60, pp 334–343

Lord S., Barnsley L., Bogduk N., (1993) Cervical Zygopophyseal Joint Pain in Whiplash. *Spine: State of the Art Reviews.* Edited by Robert Teasell and Allen Shapiro. Vol 7/Number 3, September. Hanley and Belfus Inc, Philadelphia.

Lövsund P., Nygren Ä., Salen B., Tingvall C. Neck Injuries in Rear End Collisions Among Front and Rear Seat Occupants, Proceeding of International Research Council on the Biomechanics of Impacts pp 319-325, Bergisch, Gladbach, Germany.

Lundell B., Jakobsson L., Alfredsson B., Jernst Gm C., Isaksson-Hellman I., (1998) Guidelines for and the Design of a Car Seat Concept for Improved Protection Against Neck Injuries in Rear End Car Impacts; SAE Paper No. 980301, SAE International Congress and Exposition, Detroit, February.

MADYMO Reference Manual Version 5.4.1, 1999, TNO, Netherlands

MADYMO Theory Manual Version 5.4.1, 1999, TNO, Netherlands

McIntosh A.S., Kallieris D., Frechede B., (2007) Neck Injury Tolerance under Inertial Loads in Side Impacts, Accident Analysis and Prevention 39: 326-333.

Maher J. (2000) Report Investigating the Importance of Head Restraint Positioning in Reduced Neck Injury in Rear Impact. *Accident Analysis and Prevention*. Vol. 30, pp. 299-305.

Matsushita T., Sato T.B., Hirabayashi K., Fujimura S., Asazuma T., Takatori T., (1994) X-ray Study of the Human Neck Motion due to the Head Inertial loading, 38th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 9422080.

Mc Connell W. E., Howard R.P., Van Popel J., Krause R., Guzman H.M., Bomar J.B.

Raddin J.H., Benedict J.V., Hatseli C.P., (1995) Human Head and Neck Kinematics After Low Velocity Rear-End Impacts Understanding "Whiplash". 39th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 952724.

McIntosh A.S., Kallieris D.,. Frechede B (2007), Neck injury tolerance under inertial loads in side impacts, Accident Analysis & Prevention, Vollume 39, Issue 2, pp 326–333.

Mertz H.J., Patrick L.M. (1967) Investigation into the Kinematics of Whiplash. Society of Automotive Engineers, Warrendale. Paper No 670919.

Meyer S., Weber M., Castro W., Shilgen M., Peuker C., (1998) Implementing the Recommendations of the Québec Task Force on Whiplash Associated Disorders, *Whiplash Injuries: Current Concepts in Prevention, Diagnosis and Treatment of the Cervical Whiplash Syndrome.* Edited by Gunzburg R., and Szpalski M. Lippincot-Raven Publishes Philadelphia.

Minton R., Murray P., Pitcher M., Galasko C.S.B., (1997) Causative factors in Whiplash injury: Implications For Current Seat and Head Restraint Design. IRCOBI Conference-Hannover September.

Morris A., Kullgren A., Barnes J., Truedsson N., Olssan T., Files B., (2000) Prevention of Neck Injury in Frontal Impact. Institute of Engineers Australia, Biomechanics of Neck Injury Conference, Sydney.

Morris A.P., Thomas P. (1996) A Study of Soft tissue Neck Injuries in the UK, Enhanced Safety Vehicles Conference, Melbourne, Australia. Paper No 96-S9-O-08.

Nahum A.M. and Melvin J.W. (2001) Accidental Injury: Biomechanics and Prevention, Springer-Verlag New York Inc.

National Highway Traffic Safety Administration, (1996) Head Restraints- Identification of Issues Relevant to Regulation, Design, and Effectiveness. Office of Crashworhiness Standards, Light Duty Vehicle Division.

National Highway Traffic Safety Administration (1986), Crash 3 Manual.

Naval Biodynamics Laboratory/ National Biodynamics Laboratory (NBDL), 2002, www. nbdl.org . USA.

Nolet P.S., Côté P, Cassidy J.D., Carroll L, (2010) The association between a lifetime history of a neck injury in a motor vehicle collision and future neck pain: a population-based cohort study. European Spine Journal, Volume 19, Issue 6, pp 972-981.

Norkin C.C., Leavangie P.K. (1992) Joint Structure and Function: A Comprehensive Analysis. F.A. Davis Company, Philadelphia.

Nordhoff L.S., (2005) Motor Vehicle Collision Injuries: Biomechanics, Diagnosis and Management, Jones and Bartlett Learning, 2nd Edition.

Ono K., Koji K., Wittek A., Kajzer J. (1997) Cervical Injury Mechanisms Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck -Torso Kinematices During Low Speed Rear Impacts. 41th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 973340.

Ono K., Kaneoka K., Hattori S., Ujihashi S., Takhouts E., Haffner M., Eppinger R., (2002) Cervical Vertebral Motions and Biomechanical Responses to Direct Loading of The Human Head, IRCOBI Conference, Munich (Germany).

Ono K., Kaneoka K. (1997) Motion Analysis of the Human Cervical Vertebrae During Low Speed Rear Impacts by the Simulated Sled. IRCOBI Conference, Hannover (Germany).

O'Neil B. (2000) Head Restraints -the Neglected Countermeasure. *Accident Analysis and Prevention*. Vol. 32, pp. 143- 150.

Ordway N.R., Edwards W.T., Donelson R.G., Bosco M. (1993) The Effect of Head Position on the Analysis of Cervical Motion. *Head and Neck Injuries in Sport* Hoerner E.F. ATSM.

Otte D. (1995) Review of the Airbag Effectiveness in Real Life Accidents Demands-for Positioning and Optimal Deploymnet of Airbag Systems. 39th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 952701

Panjabi M.M., Grauer J.N., Cholewicki J., Nibu K., Babat L.B., Dvorák J. (1998) Whiplash Trauma Injury Mechanism: A Biomechanical Viewpoint. *Whiplash Injuries: Current Concepts in Prevention, Diagnosis and Treatment of the Cervical Whiplash Syndrome.* Edited by Gunzburg R., and Szpalski M. Lippincot-Raven Publishes Philadelphia.

Potula S.R. Solankia K.N, Oglesby D.L., Tschopp M.A., Bhatia M.A. (2012) Investigating occupant safety through simulating the interaction between side curtain airbag deployment and an out-of-position occupant, Accident Analysis & Prevention, Volume 49, pp 392–403

Patrick L.M. (1987) Neck Injury Incidence, Mechanisms and Protection. 31st Proceedings American Association for Automotive Medicine, September 28-30, New Orleans, Louisiana.

Rizzetti A., Kalleris D., Schiemann P., Mattern R.,(1997) Response and Injury Severity of the Head-Neck Unit During a Low Velocity Head Impact. IRCOBI Conference-Hannover, September.

Schuller E., Eisenmenger W., Beier G., (1999) A Forensic Sample of Low Speed Car

Accidents, WAD'99 Compendium/Traffic Safety and Auto Engineering, Vancouver, Canada.

Sekizuka M. (1998) Seat Designs for Whiplash Injury Lessening, Proceedings from the 16th International Technical Conference on ESV, Windsor, Canada.

Sim D.F. (2004) Biomechanics of Dysfunction and Injury Management of the Cervical Spine, Ph.D. Thesis. Queensland University of Technology.

Spense A.P. (1990) Basic Human Anatomy, Benjamin-Cummings Co, United Kingdom.

Spitzer, W. O., Skovron, M. L., Salmi, L. R., Cassidy, J. D., Duranceau, J., Suissa, S., and Zeiss, E. (1995). Scientific monograph of the Quebec task force on whiplash-associated disorders: Redefining "whiplash" and its management, *Spine* 20:3–73.

Storvik S. G., Stemper B. D., Axial head rotation increases facet joint capsular ligament strains in automotive rear impact, Medical & Biological Engineering & Computing, Volume 49, Issue 2, pp 153-161.

Styrke J, Stålnacke B, Bylund P, PhD, Sojka P, Björnstig U, (2012) A 10-Year Incidence of Acute Whiplash Injuries After Road Traffic Crashes in a Defined Population in Northern Sweden, Physical Medicine and Rehabilitation, Volume 4, Issue 10, pp 739–747.

Teamming J., and Zobel R., (1998) Frequency and Risk of Cervical Spine Distortion Injuries in Passenger Car Accidents: Significance of Human Factors Data. IRCOBI Conference, Goteborg, September.

Tidus (2008) Skeletal Muscle Damage and Repair: Mechanisms and Interventions, Human Kinetics, USA.

Tomasch, E. (2004). Accident Reconstruction Guidelines. *University of Technology, Graz*

Trempel, Rebecca E., David S. Zuby, and Marcy A. Edwards. (2016)"IIHS head restraint ratings and insurance injury claim rates." *Traffic Injury Prevention,* Insurance Institute of Highway Safety.

Watts A.J., Atkinson D.R., Hennessy C.J., (1999) Low Speed Automobile Accidents, Accident Reconstruction and occupant Kinematics, Dynamics and Biomechanics 2nd Edition. Lawyers & Judges Publishing.

Welch T. D. J., Bridges A. W., Gates D. H., Heller M. F., Stillman D., Raasch C C. and Carhart M. R. (2010) An Evaluation of the BioRID II and Hybrid III During Low- and Moderate-Speed Rear Impact, SAE International Journal of Passenger Cars- Mechanical Systems, vol. 3 no. 1, pp 704-733.

White AA III, Panjabi MM. (1990) Kinematics of the spine. *Clinical Biomechanics of the Spine*. 2nd ed. Philadelphia, Pa: JB Lippincott Co.

Whiting. , Zernicke RF., (2008) Biomechanics of Musculoskeletal Injuries, Human Kinetics, USA

Wismans J., Spenny C.H., (1983) Performance Requirements for Mechanical Necks In Lateral Flexion. 27thth Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 831613

Wismans J., V.Ororschot H., Woltring H.J., (1986) Omni-directional Head-Neck Response, 30th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 861893.

Wirklund K., Larsson H, (1998) Saab Active Head Restraint (SARH), Seat Design to Reduce the Risk of Neck Injuries in Rear Impacts. 42th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 980297.

Wirklund K., (1998) Saab Active Head Restraint System, Seat Design to Reduce the Risk of Neck Injuries. *Whiplash Injuries: Current Concepts in Prevention, Diagnosis and Treatment of the Cervical Whiplash Syndrome.* Edited by Gunzburg R., and Szpalski M. Lippincot-Raven Publishes Philadelphia.

van der Horst J. G. M., Thunnissen J.G.M., Happee R., Van Haaster R.M.H.P., Wismans J.S.H.M. (1997) The Influence of Muscle Activity on Head-Neck Response During Impact, Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 973346.

van der Horst J. G. M., Bovendeerd P.H.M., Happee R., Wismans J.S.H.M., Kingma H., (2001) Simulation of Rear End Impact with a Full Body Human Model with a Detailed Neck: Role of passive Muscle Properties and Initial Seating Posture, Enhanced Safety of Vehicles, Amsterdam.

van der Horst (2002) Human Head, Neck response in Frontal, Lateral and Rear End Impact Loading, Modelling and Validation, University of Eindhoven.

van den Kronnenberg A., Phillippens M., Cappon H., Wismans J., Hell W., Langwieder K.(1998) Human Head-Neck response During Low-Speed Rear End Impacts, 42th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale. Paper No 983158.

von Koch M., Kullgren A., Nygren A., Tingvall C. Soft Tissue Injury of the Cervical Spine in Rear-end and Frontal Car Collisions. Proceedings of IRCOBI Conference on Biomechanics of Impacts, Brunnen, Switzerland, 1995; 273-283.

Ziraknejad, N. Lawrence, P.D. Romilly, D.P. (2014) Vehicle Occupant Head Position Quantification Using an Array of Capacitive Proximity Sensors IEEE Transactions on Vehicular Technology, Volume:64 Issue:6, pp 2274 – 2287.