

Neck Response in Out of Position Rear Impact Scenarios

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Hamed Shateri

Abstract

Whiplash injuries occur in automotive crashes and may cause long term health issues such as headache, neck pain, and visual and auditory disturbance. Whiplash-Associated Disorders are very costly and can impair the quality of human lives. Most studies focus on whiplash injuries that occur in neutral position head postures, although there is some evidence in the literature that non-neutral head posture can significantly increase the persistence of symptoms on patients. Crash dummies have limited biofidelity particularly for out-of-position scenarios and the current neck injury criteria were not derived for situations at which the head motion is not through the sagittal plane. Therefore Finite Element Methods provide an important tool that can be used to predict injury in different impact scenarios.

The Finite Element model which was used for this study was previously developed at the University of Waterloo representing a 50th percentile male. The model had been previously validated at the segment level in extension, flexion, tension, and axial rotation. The full cervical spine model was validated in frontal and rear impact as well as tension. Since the final validation of the model, the ligament properties of the upper cervical spine and the muscle implementations had been improved to enhance the biofidelity of the model. To further improve the model, the addition of laxities to the ligaments of the upper cervical spine was studied.

Several studies were performed based on the experiments in the literature to determine appropriate laxities for the upper cervical spine model. First, the laxities of -2 to 4 mm on all the ligaments were studied on the segment level of the model to find their effect on the failure force and displacement to failure in extension, flexion, tension, and axial rotation. The model development then went through a series of iterations in order to achieve laxity values that satisfied the failure force and displacement to failure reported in the literature for the four loading cases. Finally the laxities were used on a full cervical spine model and tested in physiological range of motion in extension, flexion, axial rotation, and lateral bending. The laxities were optimized using an iterative process. The results of this study provided laxity values that were acceptable in both segments level failure study and full cervical spine physiological range of motion study.

The model was also validated against literature in impact scenarios. Using a cadaver experiment of 7 g rear impact, the global kinematics of the cervical spine was verified against the literature. The model provided good agreement with the head kinematics and relative rotations between the vertebrae for the cadaver tests. An 8 g rear impact cadaver test was used to validate the ligament strains and

disc shear strains. For the anterior longitudinal ligament, the capsular ligament, and the disc shear strains, the model results were within one standard deviation of the literature in the majority of cervical spine regions that were reported. The model was also validated against volunteer low severity rear impact to verify the active musculature in the cervical spine. The head kinematics was generally within the boundaries that were reported by the literature.

The model was compared to an experiment that used cadavers to investigate non-neutral rear impact scenarios. This experiment used cables and springs to replicate the passive behaviour of the musculature. The model showed good agreement with the extension and axial rotation results in both head kinematics and relative vertebrae rotations. The flexion and lateral bending results were not similar to the experimental data; attributed to the difference in muscle implementation between the two models.

A total of 24 simulations were completed to find the effect of impact severity, axial rotation, and muscle activations on ligament strains during out-of-position rear impacts. The results illustrated that in general, ligament strains increased with the severity of impact and decreased with muscle activation. In out-of-position scenarios, the strains increased in some of the ligaments. An increase to the ligament strain as a result of non-neutral posture was mostly visible in the capsular ligaments of the upper cervical spine. The alar ligament and the apical ligaments of the upper cervical spine may fail in out-of-position at high rear impact scenarios.

Recommendations for future work on the cervical spine Finite Element model includes the validation of the musculature and the usage of the muscles to rotate the head to a desired position to improve the biofidelity of the model and the results in out-of-position rear impacts. Further optimization of the laxities of the upper cervical spine can increase the biofidelity in this region. The modeling of the vertebral arteries into the FE model can help investigate whether out-of-position can increase the chance of injury of this region. The effect of flexion, extension, lateral bending, and their combination with axial rotation and the study of frontal and side impacts can be helpful in design of safer headrests for vehicles.

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Table of Content

Author’s Declaration	ii
Abstract	iii
Acknowledgements	v
List of Figures	viii
List of Tables.....	xi
Chapter 1 Introduction.....	1
1.1 Motivation for Research.....	1
1.2 Objectives and Approach	2
1.3 Outline	3
Chapter 2 Anatomy and Physiology of the Cervical Spine	5
2.1 Biomechanics Terms	5
2.2 Vertebra.....	6
2.3 Intervertebral Discs	11
2.4 Facet Joints	13
2.5 Ligaments	15
2.6 Muscles.....	20
Chapter 3 Background.....	26
3.1 Whiplash Injuries.....	26
3.2 Experimental Methods.....	32
3.3 Injury Evaluation Criteria.....	33
3.3.1 N_{ij} Criterion	33
3.3.2 Neck Injury Criterion (NIC).....	36
3.3.3 IV-NIC.....	37
3.3.4 Tissue Strain	38

3.4 Injury Locations.....	39
3.5 Neutral versus Out-of-Position Impact Scenarios	42
3.6 Out-of-Position Numerical Models	47
Chapter 4 Model Description	51
4.1 Vertebrae	51
4.2 Intervertebral Discs	54
4.3 Facet Joints	58
4.4 Ligaments	61
4.5 Muscle	66
4.6 Model Enhancements	71
Chapter 5 Model Validation	74
5.1 Segment Level Validation	74
5.2 Full Cervical Spine - Physiological Range of Motion Validation.....	84
5.3 Rear impact validations	89
5.4 Model Comparison to Simplified Out of Position Experiment	98
Chapter 6 Soft Tissue Strain Evaluation for Out-of-Position Impact.....	103
6.1 Modeling Out-of-Position Impact	103
6.2 Impact Severity.....	106
6.3 Active and Passive Musculature.....	107
6.4 Rotation Severity.....	108
Chapter 7 Summary and Recommendations	114
7.1 Summary	114
7.2 Recommendations	117
References	120
Appendix A Rear Impact Results	134

List of Figures

Figure 2-1: Anatomical planes and directions	5
Figure 2-2: Neck rotation terminologies	6
Figure 2-3: Vertebral column	7
Figure 2-4: Typical vertebra	8
Figure 2-5: Cervical spine vertebrae	9
Figure 2-6: C1 and C2 vertebrae	10
Figure 2-7: Intervertebral Disc	12
Figure 2-8: Detailed intervertebral disc	13
Figure 2-9: Facet Joints	14
Figure 2-10: Detailed facet joint.....	14
Figure 2-11: ALL, PLL, and Ligamenta flava.....	15
Figure 2-12: Lateral view of cervical spine ligaments	16
Figure 2-13: Upper cervical spine ligaments (Natter, 2006)	18
Figure 2-14: Normalized load-displacement curve for ligaments	19
Figure 2-15: The anterior view of deep neck muscles.....	21
Figure 2-16: Muscles of the neck (Lateral view)	22
Figure 2-17: Posterior view of neck muscles	23
Figure 2-18: The anterior view of superficial muscles.....	24
Figure 2-19: Deep muscles of the neck	25
Figure 3-1: Cervical spine motion after impact.....	27
Figure 3-2: Incident rate of all motor vehicle injuries by age and gender in Quebec (1987).....	28
Figure 3-3: Incidence rate of compensated whiplash injury by age and gender in Quebec (1987).....	28
Figure 3-4: Overall one year cumulative return to activity curve for cohort of whiplash subjects	29
Figure 3-5: Injury tolerance curves for axial neck compression force	34
Figure 3-6: Neck injury criteria for 50th percentile male dummy.....	35
Figure 3-7: N_{ij} risk curves (Eppinger et al., 2000)	36
Figure 3-8: Mean percentage (Left) and mean peak angles (Right) of the time outside neutral range of motion and rotation from left to right in degrees.....	43
Figure 3-9: Kallemeyn's ligament properties for the baseline (left) and calibrated (right) models.....	49
Figure 4-1: Skull and T1 vertebra	52
Figure 4-2: Cervical vertebrae geometries and mesh	53

Figure 4-3: Cervical vertebrae components	54
Figure 4-4: Length and initial posture of the cervical spine model	55
Figure 4-5: Intervertebral disc components	56
Figure 4-6: Stress versus strain along fibre directions of annulus fibres.....	57
Figure 4-7: Annulus fibrosus ground substance model fit (Panzer, 2006)	58
Figure 4-8: Facet joints.....	59
Figure 4-9: Cartilage in the upper cervical spine (Panzer, 2006)	59
Figure 4-10: Simple squeeze-film model	61
Figure 4-11: Pressure versus relative volume properties for synovial fluid (Panzer, 2006)	61
Figure 4-12: Sample ligament implementations in the upper cervical spine.....	62
Figure 4-13: Sample ligament implementations in the middle and lower cervical spine.....	62
Figure 4-14: Free-body diagram to extract preload data from a study by Heuer et al. (2007) (Fice, 2010).....	63
Figure 4-15: Force-deflection curves for upper cervical spine ligaments (Part 1) (Fice, 2010).....	64
Figure 4-16: Force-deflection curves for upper cervical spine ligaments (Part 2) (Fice, 2010).....	65
Figure 4-17: Force-deflection curves for middle cervical spine ligaments (Fice, 2010).....	65
Figure 4-18: Force-deflection curves for lower cervical spine ligaments (Fice, 2010).....	66
Figure 4-19: Neck muscles in the cervical spine model in flexion (left) and extension (right) (Fice, 2010).....	67
Figure 4-20: Hill Muscle Model Schematic	67
Figure 4-21: Parallel element response for Hill muscle model (Panzer, 2006).....	68
Figure 4-22: Active muscle force dependency on the length of muscle (Panzer, 2006)	69
Figure 4-23: Active muscle force dependency on contraction velocity (Panzer, 2006).....	69
Figure 4-24: Muscle activation function for a 100 ms step Neutral input starting at 74 ms (Fice, 2010)	70
Figure 4-25: Post-failure regression fit for the ALL	72
Figure 4-26: New muscle implementation (Left) & old muscle implementation (Right)	73
Figure 5-1: Validation process of segment level studies	77
Figure 5-2: Segment level axial rotation test apparatus	78
Figure 5-3: Segment level axial rotation simulation results	79
Figure 5-4: Torque comparison of axial rotation in range of motion	79
Figure 5-5: Comparison of C01/C12 ratio between the model and the average and range from the literature.....	80

Figure 5-6: Segment level extension and flexion test apparatus	81
Figure 5-7: Segment level extension simulation results	82
Figure 5-8: Segment level flexion simulation results	82
Figure 5-9: Segment level tension test apparatus	83
Figure 5-10: Segment level tension simulation results.....	84
Figure 5-11: Physiological range of motion simulation constrains	86
Figure 5-12: Axial physiological range of motion results from full cervical spine simulation.....	87
Figure 5-13: Extension physiological range of motion results from full cervical spine simulation.....	88
Figure 5-14: Flexion physiological range of motion results from full cervical spine simulation	88
Figure 5-15: Lateral physiological range of motion results from full cervical spine simulation	89
Figure 5-16: T1 prescribe motion for the 7 g rear impact (Deng, 1999).....	90
Figure 5-17: Head rotation of the model during 7 g rear impact.....	91
Figure 5-18: Head CG X acceleration of the model during 7 g rear impact	92
Figure 5-19: Head CG Z acceleration of the model during 7 g rear impact	92
Figure 5-20: Intervertebral rotation validation during 7 g rear impact.....	93
Figure 5-21: T1 acceleration profile to simulate the 8-g rear impact	94
Figure 5-22: Peak ligament strain comparison during 8 g rear impact.....	95
Figure 5-23: The peak disc shear strain during 8 g rear impact	96
Figure 5-24: T1 inputs for the 4 g simulation (Davidsson et al., 1998).	97
Figure 5-25: Model during the 4 g simulation.....	97
Figure 5-26: Head CG X displacement response to a 4 g rear impact	98
Figure 5-27: Head occipital condyle displacement response to a 4 g rear impact.....	98
Figure 5-28: Ivancic et al. (2006a) experiment setup	99
Figure 5-29: Applied acceleration to the T1 for 8 g rear impact.....	100
Figure 5-30: Head angles comparison with Ivancic et al. (2006a) experiment	101
Figure 5-31: Intervertebral rotation comparison with Ivancic et al. (2006b) experiment	102
Figure 6-1: T1 prescribed motion for 10 g rear impact (Deng, 1999).....	103
Figure 6-2: Application of 60.5° axial rotation on the cervical spine.....	104
Figure 6-3: Cervical spine motion during 60.5° out-of-position impact (7 g).....	105
Figure 6-4: Effect of impact severity on capsular ligaments.....	106
Figure 6-5: Effect of muscle activity on ligament strains of capsular ligament.....	108
Figure 6-6: Effect of rotation on ligament strains during 16 g rear impact with active musculature. .	110
Figure 6-7: C6-C7 relative rotation during extension in 16 g rear impact	112

List of Tables

Table 3-1: Clinical classification of Whiplash-Associated Disorders	30
Table 3-2: Symptoms after whiplash injuries of the neck at initial presentation and at final follow-up (percentages)	31
Table 3-3: Influence of features of accident mechanism on the long-term course.....	44
Table 4-1: Skull mass and inertial information	52
Table 4-2: Nucleus pulposus material parameters.....	56
Table 4-3: Material properties for Articular Cartilage (Panzer, 2006).....	60
Table 4-4: Model's ligament preloads of the middle and lower cervical spine.....	64
Table 4-5: Upper cervical spine ligament properties.....	71
Table 5-1: Study of laxity on individual ligaments	75
Table 5-2: Final laxity values	87
Table 6-1: Study simulations.....	105
Table 6-2: Impact levels that may cause injury to the cervical spine ligaments	107

Chapter 1

Introduction

1.1 Motivation for Research

The term whiplash injury is often referred to as strains and sprains in the neck. With it are the most common injuries treated in emergency departments in U.S. hospitals (Quinlan et al., 2004). Whiplash injury incidence rates range from 28 to 834 per 100000 each year (Cassidy et al., 2000; Osteremski et al., 1989). The highest incidence rates have been reported to be young females between the ages of 15-24 (Quinlan et al., 2004; Spitzer et al., 1995). The annual cost related to neck injuries is estimated to be 4.5-29 billion dollars in the United States and 5-10 billion Euros in Europe (Kinberger, 2000; Freeman, 1997; Schmitt et al., 2004). In addition to the cost, patients may suffer from chronic neck pains or other Whiplash-Associated Disorders (WAD) such as headache and visual weakness for the rest of their lives.

Many of the studies that have been performed on neck injuries during car collisions have focused on situations in which the head of the occupants are in a neutral position. Shugg et al. (2011) studied head posture of drivers in residential areas and reported that drivers had out-of-position posture during at least 19% of the time when driving and at least 40% of the time when stopped at a stop sign. Sturzenegger et al. (1995) studied various features that might influence neck injury in car collisions and reported that head position was the only factor that affected the persistence of symptoms in patients. There is not enough scientific evidence on cadaver specimen to support the claim that out-of-position head posture could increase the risk of neck injury during impact.

Crash tests have been developed to insure the safety of the vehicle for the consumers. Crash tests are necessary but expensive. The N_{ij} criterion uses the resultant forces of the neck of crash dummies to evaluate the severity of injury during frontal crashes. Unfortunately the N_{ij} criterion is not derived from rear impact scenarios. There have been other neck injury criteria developed, however none have been accepted by the National Highway Traffic Safety Administration to be used for crash tests. Until now only one neck injury criterion has been used for out-of-position head posture. Intervertebral neck injury criterion (IV-NIC) proposed that intervertebral motion beyond the physical range of motion in any directions may cause soft tissue injuries (Panjabi et al., 1999). This method is promising, but there is large variability in the physiological range of motion between individuals and to this date, and there is not enough evidence to support that this method could predict injury in all directions. Tissue

strains, however, can be used to locate the site of injury under any type of loading condition. Strains beyond the physiological limits of ligaments could damage the tissue. Measuring tissue strains is not currently possible on crash dummies and is very difficult for cadaver models. Finite Element (FE) models are a good method to create biofidelic cervical spine models and to predict injury using tissue strains. This thesis will focus on ligament tissue strains in order to predict injury in rear-impact collisions using Finite Element Analysis (FEA).

1.2 Objectives and Approach

The purpose of this research was to validate the existing University of Waterloo (UW) (Panzer, 2006; Panzer & Cronin, 2009) cervical spine model in out-of-position postures and to find the effects of various features on the severity of injury during impact. This research attempted to further validate the model to improve its behaviour in out-of-position impact scenarios. Once the model was validated, features such as severity of impact, effect of posture, and effect of active musculature on injury were investigated.

The UW model was developed to represent a 50th percentile male. The model was previously validated at the both segment level and full cervical spine. At the segment level, the model was validated in flexion, extension, shear, tension, and compression (Panzer, 2006; Panzer & Cronin, 2009). The full cervical spine model had been validated in tension without musculature, and in rear and frontal impact with passive musculature (Fice, 2010; Fice et al., 2011; Panzer et al., 2011).

Mattucci (2011) performed tensile testing for the cervical spine ligaments using young specimen (less than 50 years old). The ligaments were generally stiffer than the previous UW model because of the age difference. The ligament properties of the upper cervical spine measured by Mattucci (2011) were selected to be used in the UW model because they provided good overall results for the segment model validations. Laxities were applied to some of the ligaments to ensure proper rotational behaviour in the axial direction. Ligament failure was possible during the validation cases and therefore was included in the model. Disc failure in the full cervical spine was not previously validated and was not expected with the severities of the validation cases in out-of-position and therefore it was not included in the model. The segment models were used to validate the upper cervical spine against the literature in tension, flexion, extension, and axial rotation. The results were

then used to validate the range of motion of the full cervical spine model in extension, flexion, axial, and lateral rotations using a study by Ivancic et al. (2006b).

The validated model was used to investigate the effect of head posture, impact severity, and muscle activation on the ligament strains in out-of-position rear impacts. The head postures that were studied were the average \pm one standard deviation of the maximum head rotations reported by Shugg et al. (2011). The high impact severities that were investigated were similar to those studied by Fice (2011) in rear impact collisions with neutral head postures. The final effect was investigated by simply activating the muscle activation in the model and observing the changes in the ligament strains.

1.3 Outline

The second chapter of this thesis is written to provide readers with a deeper understanding of the anatomy and physiology of the cervical spine. This chapter starts with the introduction of basic biomechanical terms and then describes the main structures of the cervical spine. These structures include the vertebra, intervertebral discs, facet joints, ligaments, and muscles. This chapter is necessary for readers who are not familiar with the anatomy of the cervical spine; understanding this chapter is of great help in acknowledging the material in the following chapters.

Chapter three provides a background on whiplash and previous studies that can be related to this research. This chapter starts by providing a general description of whiplash and how it impacts society. The common experimental techniques and the methodologies that have been used to date to measure injury to the neck are presented. This chapter includes information on sources of whiplash pain, previous studies on out-of-position collisions, and non-neutral numerical models that have been developed by various authors.

Chapter four describes how the UW model was developed. This chapter goes into detail about geometry and material properties that have been used to model the different aspects of the cervical spine. This chapter follows the same categories as chapter two in order to help the readers make reference if more information on the specific anatomy is required. The model changes that have been adapted from other members are also included in this chapter.

Chapter five describes the validation cases that had been investigated to improve the behaviour of the model in out-of-position. Previous work on the segment level had suggested that addition of laxity to

the ligaments of the upper cervical spine can improve the biofidelity of the model. The model was firstly validated against literature at the segment level to avoid complexities that arise from the combination of the mid and lower cervical spine. The full cervical spine validation in physiological range of motion is described in the second portion of this chapter. In the third section, validations for neutral impacts are performed and the last section compares the current model relative to an out-of-position cadaver experiment.

Chapter six provides information on whether or not the model predicted higher strains in the ligaments as a result of out-of-position posture. The effect of impact severity, active musculature, and axial head rotation was investigated in this chapter. The results are discussed in more details and the final chapter provides summary from the research and recommendation that could help to improve this research in the future.

Chapter 2

Anatomy and Physiology of the Cervical Spine

This section provides an overview of the anatomy of the human cervical spine. The basic terminologies that are used in the biomechanics of the cervical spine are introduced. The various structures of the human neck and how they are interconnected to provide individuals with such great flexibility of movements are discussed. The material properties that are used to describe these structures in the literature are also mentioned to provide an understanding to the reader on how these parts will be modeled for the purpose of this thesis.

2.1 Biomechanics Terms

Three imaginary planes are used to describe the location of body parts (Figure 2-1). The sagittal plane divides the body into the left and right side. The medial direction refers to body parts that are close to the sagittal plane and the lateral or distal direction refers to parts that are away from this plane. The frontal plane divides the body into front and back sides. Any body part towards the front is in the anterior direction whereas any part towards the back is considered to be in the posterior direction. The Transverse plane divides the body into the upper and lower sections. The superior direction is designated for structures that are above this plane and the inferior direction describes body parts that are below this plane.

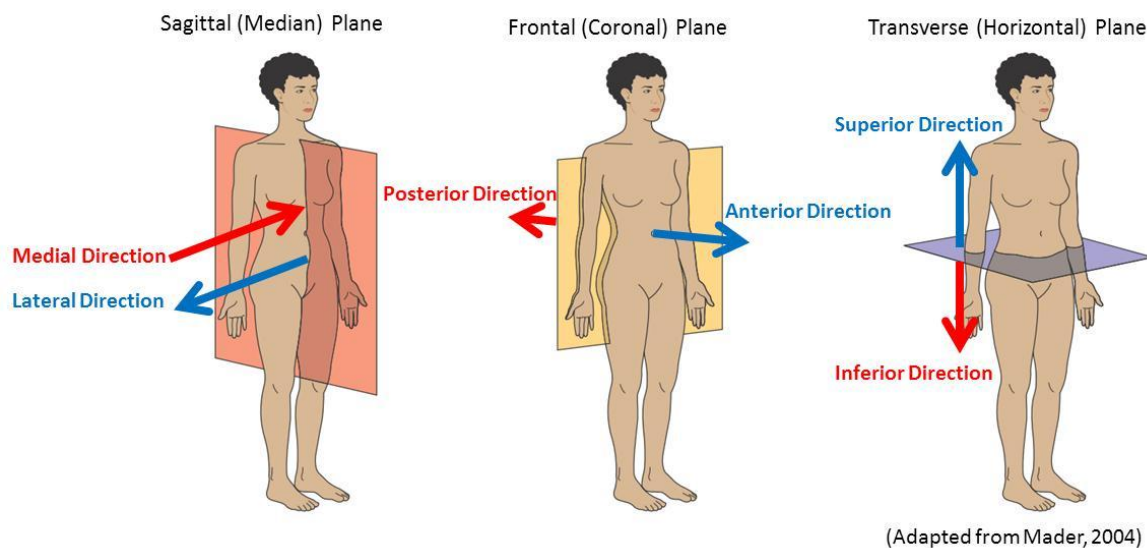


Figure 2-1: Anatomical planes and directions

Neck movement and a normal range of motion in the cervical spine are important for undertaking normal activities on a daily basis. Neck positions are measured from the neutral position. Neutral position is when an individual is looking forward. The terminologies used for head and neck movements are extension, flexion, axial rotation, and lateral rotation (Figure 2-2). Extension is tilting the head backward and flexion is tilting it forward, bending it towards the chest. Axial rotation refers to rotating the head to the left and right. In lateral bending, the neck bends so that the ear moves towards the shoulder as shown in the figure.

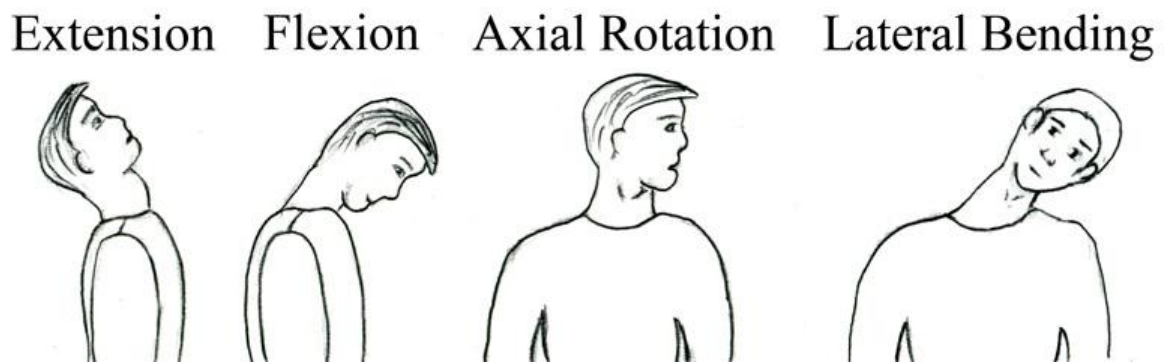
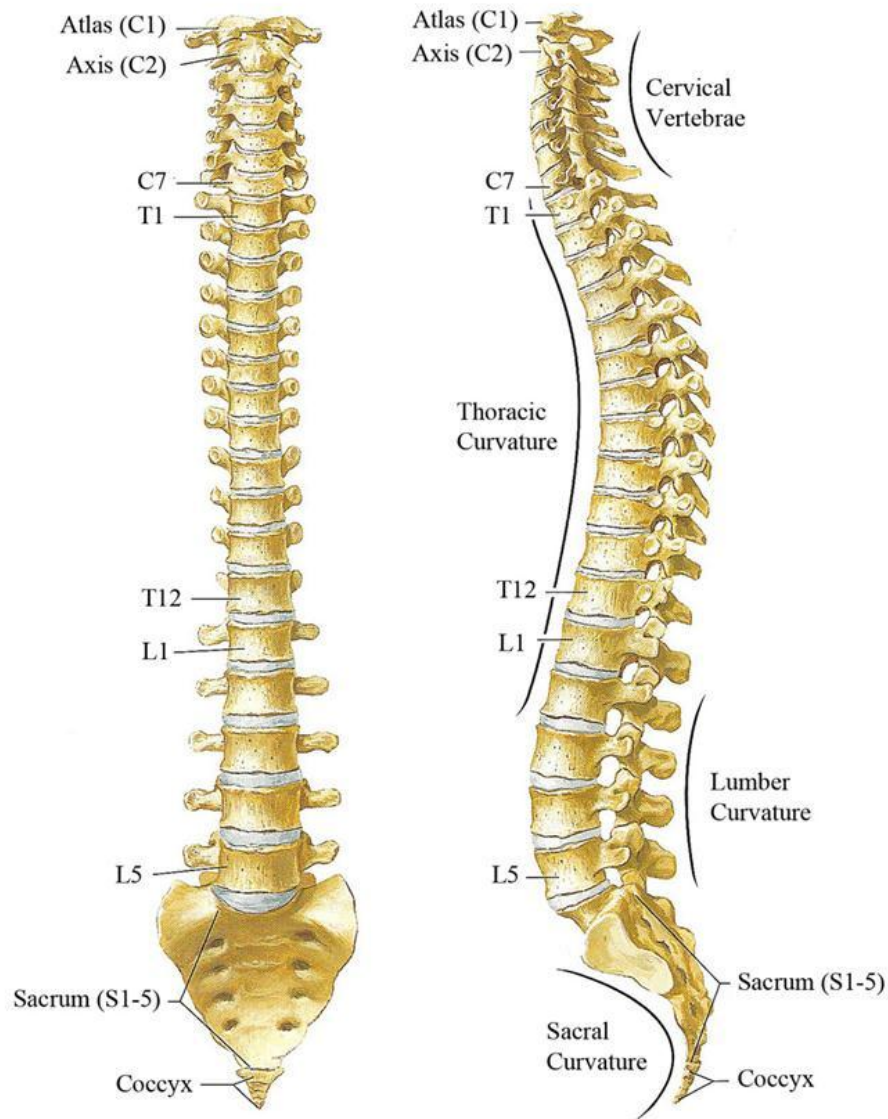


Figure 2-2: Neck rotation terminologies

2.2 Vertebra

The vertebral column extends from the skull to the tip of the coccyx (Figure 2-3) (Moore & Dalley, 2006). The vertebral column is a vital structure for humans as it supports the weight of the body, protects the spinal cord and spinal nerves, provides flexible axis for the body and a pivot for the head, and it is important to the role of posture and human movements. Most vertebral columns in adults are 72 to 75 cm in length; typically consisting of 33 vertebrae, spread through five regions. There are seven cervical, 12 thoracic, five lumbar, five sacral, and four coccygeal vertebral bodies in vertebral column. The inferior vertebrae are larger because they carry a larger portion of the body weight than the superior vertebrae. The superior 25 vertebral bodies are responsible for a large portion of the motion of the vertebral column.



(Adapted from Netter, 2006)

Figure 2-3: Vertebral column

The vertebral column consists of many bones called vertebrae which are separated by intervertebral discs (Moore & Dalley, 2006). The vertebral body comprise cancellous bone enclosed by a thin layer of compact bone. The characteristics and size of the vertebrae are different between regions of the column but are comparable within each region. The basic functions of all vertebrae are similar. Figure 2-4 shows the functional components of a typical vertebra. The vertebral body is shown in

light yellow, vertebral arch in red, and processes in blue and yellow. There are seven processes in each vertebra. The spinous process and two transverse processes are the locations where muscles are attached to the vertebra. These processes act as levers to help rotations between adjacent vertebrae. The other processes, known as articular processes, cooperate with adjacent articular processes to determine the allowable movement between vertebrae. Articular processes also help in alignment of adjacent vertebrae.

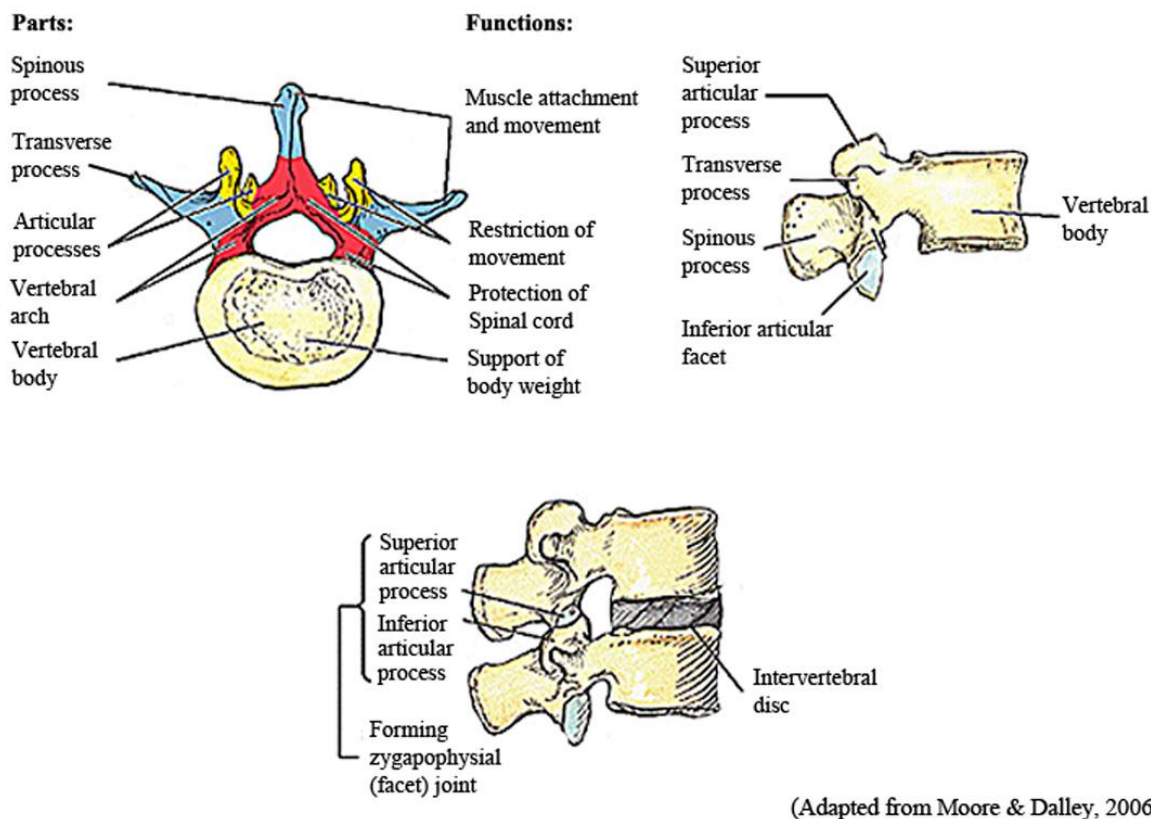


Figure 2-4: Typical vertebra

There are total of seven vertebrae in the cervical spine (Moore & Dalley, 2006). These vertebrae are smaller than those in thoracic and lumbar regions. The intervertebral discs in this region are also smaller; however, compared to other regions, the discs are relatively large. In addition, the small amount of surrounding body mass, the relative small vertebrae, and thick discs allow the cervical spine to have largest range of motion compared to the other regions of the vertebral column. Figure 2-5 shows the shape of all cervical spine vertebrae. C3 to C7 vertebrae have the same typical

structure as those in other regions of the spine. The vertebral foramen is larger in these vertebrae to accommodate the enlargement of the spinal cord. The C7 vertebra has the largest spinous process which can be felt by running a finger down the midline of the posterior portion of the neck.

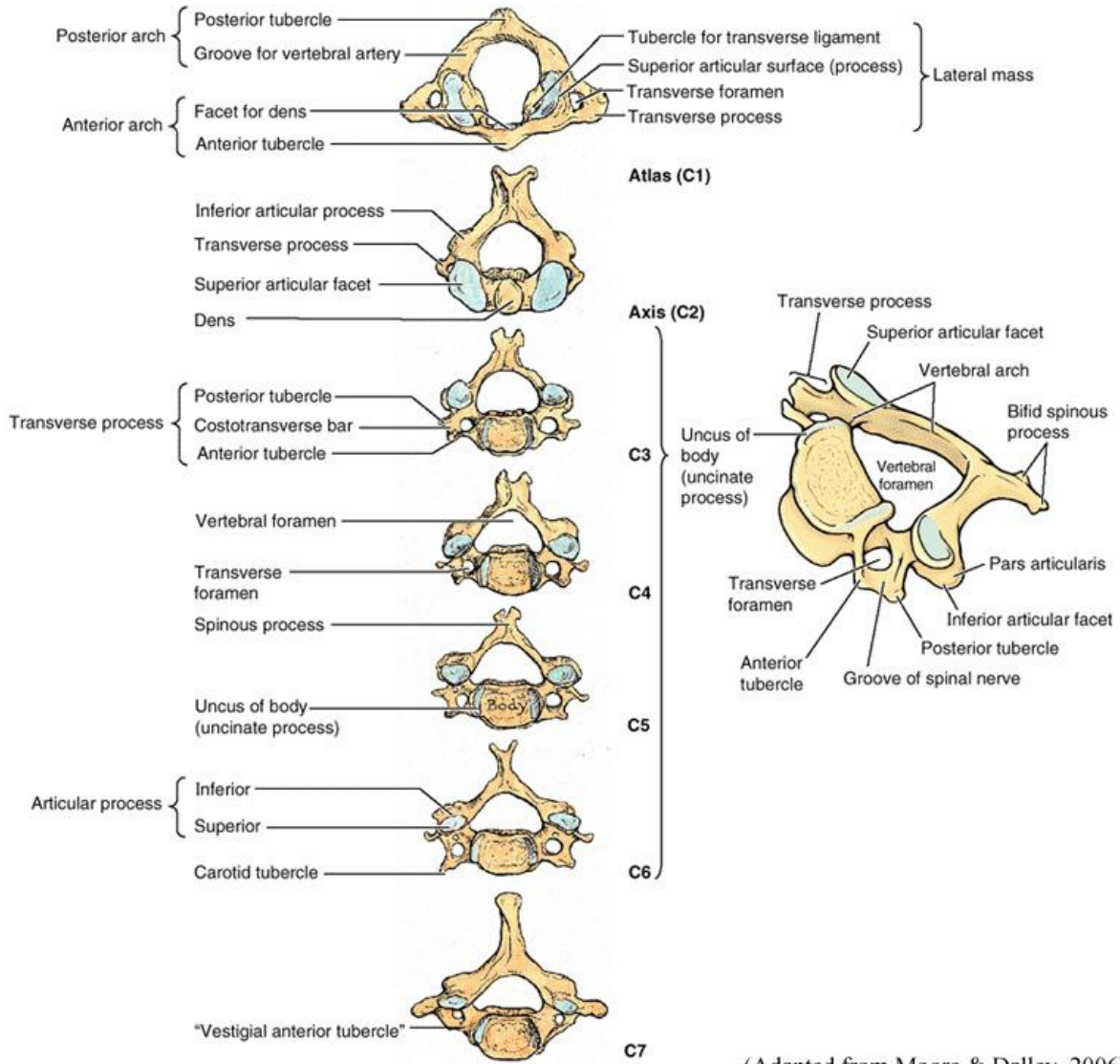


Figure 2-5: Cervical spine vertebrae

The two superior vertebrae are unique compared to the other vertebrae in the vertebral column (Figure 2-6) (Moore & Dalley, 2006; Mader, 2004). The C1 vertebra is called the atlas. The atlas does not have a body or spinous process. The transverse processes in the atlas are more laterally placed

than the inferior vertebrae which allows for increased leverage for the attached muscles. The occipital condyle articulates with the superior facet of the atlas. The C2 vertebra is called the axis. The axis is the strongest vertebra in the cervical spine and it contains a special structure which is the odontoid process. The odontoid process, also known as dens, provides an axis for atlas rotation. The anterior portion of the dens articulates with the anterior facet of the atlas and the posterior portion is held in place by the transverse ligaments of the atlas. The transverse ligaments prevent the anterior/posterior displacement between the atlas and the axis. The odontoid process allows for the skull to rotate axially on the cervical spine.

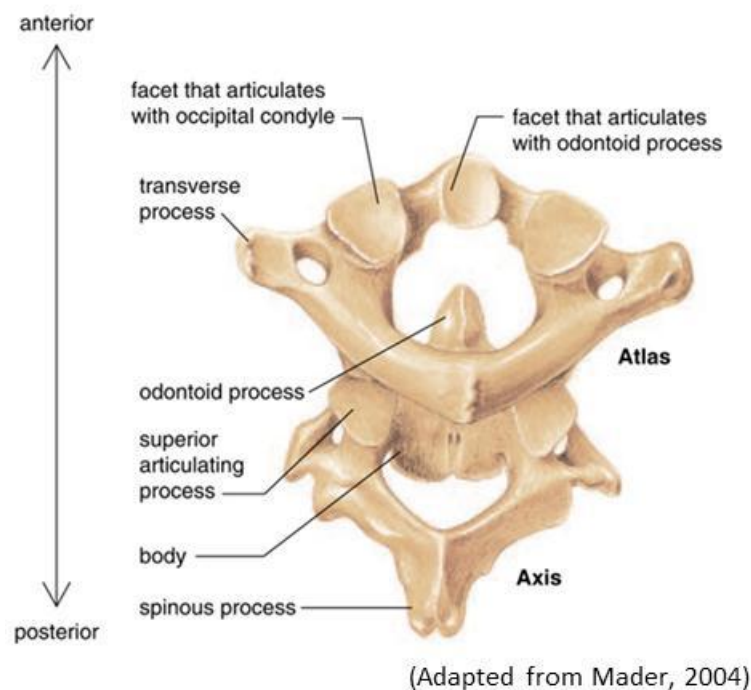


Figure 2-6: C1 and C2 vertebrae

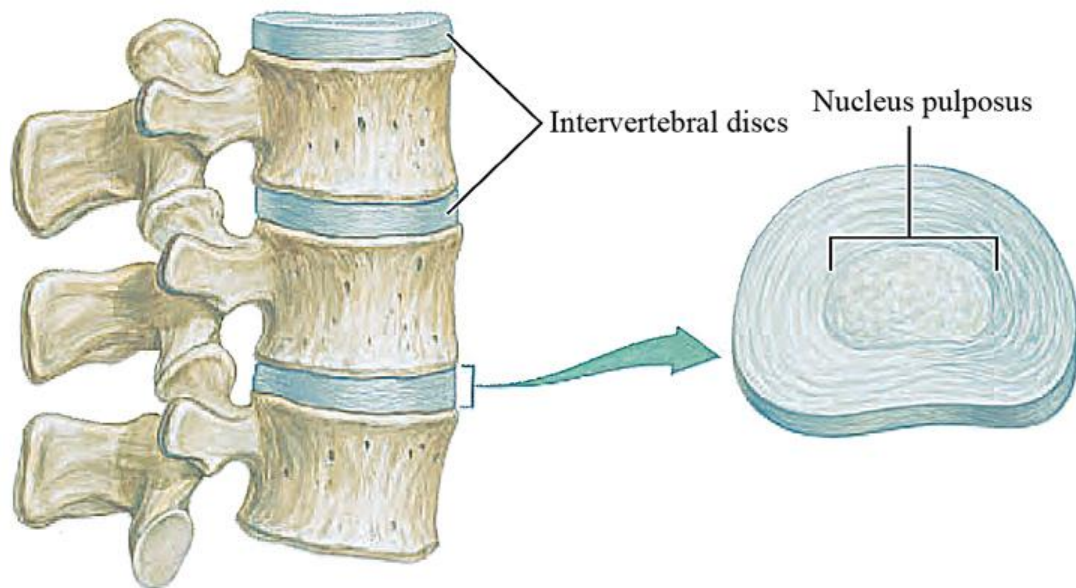
Cervical vertebrae are orientated horizontally and therefore are less tightly interlocked than the other vertebrae in the spine. As a result, forces can dislocate adjacent vertebrae and cause fracture. Slight dislocation without damage can occur in the cervical spine because of the large vertebral canal in this region. Large dislocations can cause impingement and injure the spinal cord. The ability of the vertebrae to slip back into place makes it difficult for radiographs to observe injury. Magnetic Resonance Imaging (MRI) may be used to reveal the resulting soft tissue damage (Moore & Dalley, 2006). In compression, the forces can cause a risk of fracture in the anterior and

posterior regions of the atlas. If the force is severe, it could also rupture the transverse ligaments which can cause spinal cord injury. A hangman's fracture is a fracture which is caused by hyperextension of the head on the neck. This causes the fracture of vertebral arch of the axis which is a common injury in the cervical vertebrae (Yochum and Rowe, 2004). In severe cases, the body of the C2 displaces anteriorly with respect to the C3 vertebra. This can damage the spinal cord and/or brainstem, resulting in paralysis or death. Fracture of odontoid process is also a common injury in the axis which may be caused from bone loss or horizontal impact to the head.

Cortical bone tissue has a composition of 65% mineral and 35% organic matrix (Cowin, 2001). The cortical bone is dense and often on the surface of the bone while the cancellous bone is porous and is surrounded by the cortical bone. A fluid mixture of blood and marrow fill the voids of cancellous bone (Carter & Hayes, 1977). Mineral density and apparent density of bone are two important factors that determine the properties of bone. The mineral density of the bone decreases with age resulting in reduction of compressive strength in the vertebrae (Carter & Hayes, 1997; Cody et al., 1991). The viscoelastic properties of cortical and cancellous bones are shown through creep, relaxation, and strain-rate stiffening effects (McElhaney, 1966; Carter & Hayes, 1977; Fondrk et al., 1988; Linde et al., 1991; Bowman et al., 1994).

2.3 Intervertebral Discs

The intervertebral discs are positioned between all adjacent vertebrae except C1 and C2. They form strong attachments and act as cushions to absorb shock and to prevent contact between vertebrae (Figure 2-7; Moore & Dalley, 2006; Mader, 2004). The intervertebral discs also allow adjacent vertebrae to move relative to one another while resisting compressive loads (Graaff, 2001). As humans age, discs become weaker and the possibility of disc rupture increases. Damaged discs, also called herniated discs, can cause pain if they come into contact with the spinal cord or spinal nerves. The removal of the damaged disc can fuse two adjacent vertebrae while limiting the flexibility between them. Disc herniation in the cervical spine region is less common than the lumbar region and the discs that are most vulnerable to damage are C5-C7 vertebrae (Snell, 2006).



(Adapted from Graaff, 2001)

Figure 2-7: Intervertebral Disc

Intervertebral discs contain nucleus pulposus at the centre, surrounded by annulus fibrosus on the outside (Figure 2-7; Moore & Dalley, 2006). The annulus fibrosus is composed of various layers with an arrangement which allows movement between adjacent discs and provides resistance to rotational and loading forces (Figure 2-8). The collagen fibres are oriented in the same direction in each layer but opposite in adjacent layers. The fibers in the posterior section of the vertebral column are thinner and smaller in numbers. This is why the nucleus pulposus is located posteriorly when looked from the sagittal plane. This causes a wedge shape in the intervertebral discs. The literature reports the thickness of the anterior side of the discs between 4.8-5.5 mm and the posterior side as 3.0-4.3 mm (Gilad & Nissan, 1986; Przybylsky et al., 1998). The nucleus pulposus is more moist and cartilaginous at birth (Moore & Dalley, 2006; Graaf, 2001). The semifluid structure allows for flexibility and resilience of the intervertebral discs. Figure 2-8 shows that under compressive loading, the annulus flattens and during flexion and extension, the nucleus acts as a pivot. During extension, flexion, or lateral motion of the neck, the annulus fibrosus are under compression on one side of the nucleus, and under extensions on the opposite side.

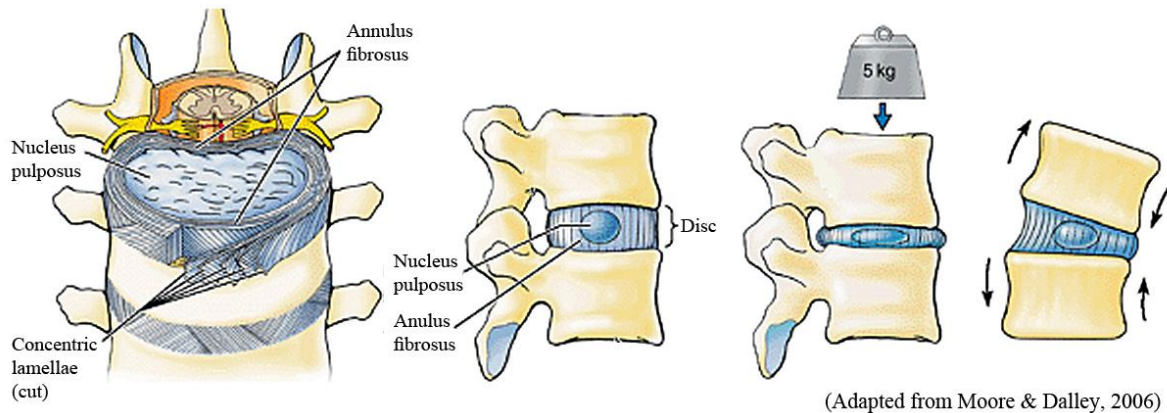


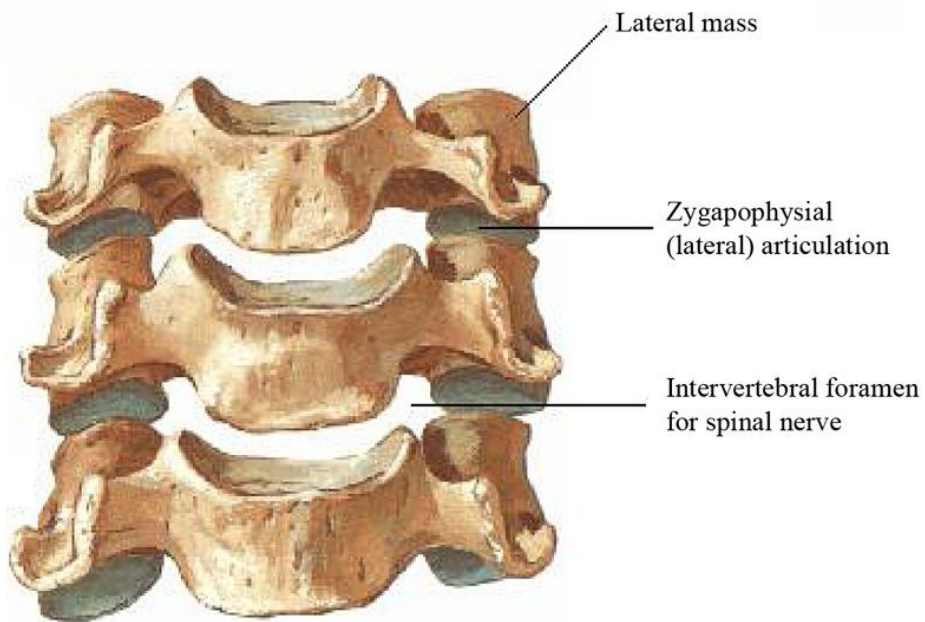
Figure 2-8: Detailed intervertebral disc

The annulus fibrosus tissue has non-linear tensile properties. The stress-strain behaviour starts in the toe region as the fibres stretch, and becomes linear when all the fibres have been straightened (Elliot & Setton, 2001; Pezowicz et al., 2005). Viscoelastic behaviour was found through relaxation testing of the annulus ground substance (Iatridis et al., 1998; Klisch & Lotz, 2000). The nucleus pulposus exhibits viscoelastic-solid characteristics in dynamic conditions and viscoelastic-fluid properties in relaxation-type loadings (Iatridis et al., 1996).

2.4 Facet Joints

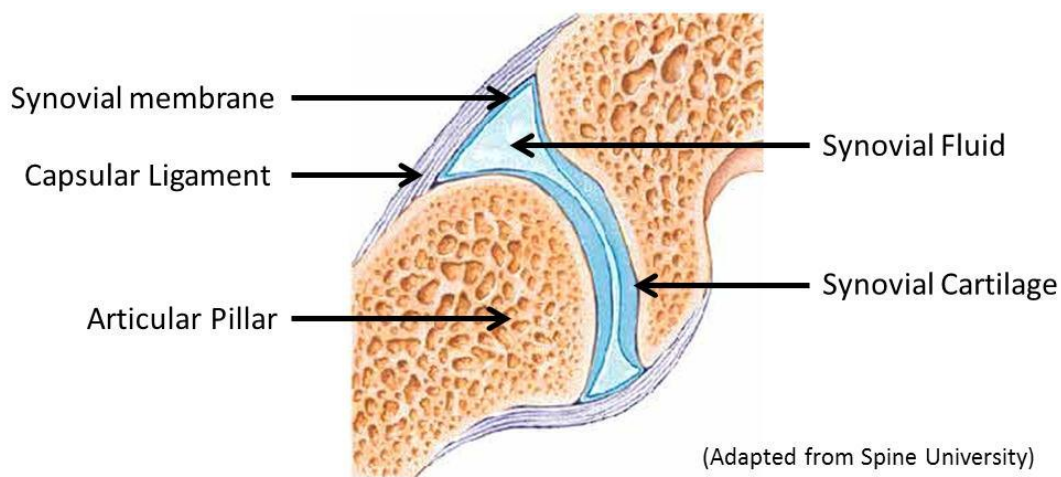
Facet joints or zygapophysial joints are synovial joints between superior and inferior articular processes of two adjacent vertebrae (Moore & Dalley, 2006; Figure 2-9). The shape of the articular surfaces determines the type of movements which is permitted by the facet joints. Gliding happens between superior and inferior articular surfaces of two adjacent vertebrae (Figure 2-10). Synovial fluid prevents wear by allowing the surfaces to glide with extremely low friction. The synovial fluid is enclosed by the synovial membrane and the capsular ligaments resist tensile loads to the joint. The relative size of intervertebral discs relative to the vertebral body determines the range of motion between vertebrae. In the cervical spine region, the facet joints help the intervertebral discs with load bearing. The spinal nerves emerge from the vertebral canal through the intervertebral foramina. Damage to the facet joints due to aging can affect these nerves and cause pain. To treat back-pain, denervation of lumbar facet joints, in which the nerves are sectioned or destroyed using radiofrequency percutaneous rhizolysis, is used. Cartilage tissues have been identified as having

viscoelastic behaviour which is largely dependent of the porosity of the tissue (Hayes & Mockros, 1971; Mow & Guo, 2002).



(Adapted from Netter, 2006)

Figure 2-9: Facet Joints



(Adapted from Spine University)

Figure 2-10: Detailed facet joint

2.5 Ligaments

The ligaments in the cervical spine will be discussed in this section. The anterior longitudinal ligament (ALL) connects the anterior part of adjacent vertebral bodies (Moore & Dalley, 2006; Figure 2-11). The ALL is a thick fibrous band and the only ligament which is responsible for limiting the extension motion. Therefore this ligament protects the cervical region from rear impact collisions and can be disrupted through hyperextension motion (Harris & Yeakley, 1992; Rao et al., 2005). This ligament runs from C2 to T1 in the cervical spine region. Anterior atlanto-occipital membrane (AAOM) and anterior atlanto-axial membrane (AAAM) are continuation of the ALL, located between the skull and atlas and atlas and axis, respectively. The posterior longitudinal ligament (PLL) starts at C2 and runs past the T1 vertebra in the cervical region. This ligament is weaker than the ALL and has a low contribution in resisting hyperflexion. Posterior atlanto-axis membrane (PAAM) is the continuation of PLL located between the atlas and the axis. Posterior atlanto-occipital membrane (PAOM) is the continuation that is attached between the atlas and the skull.

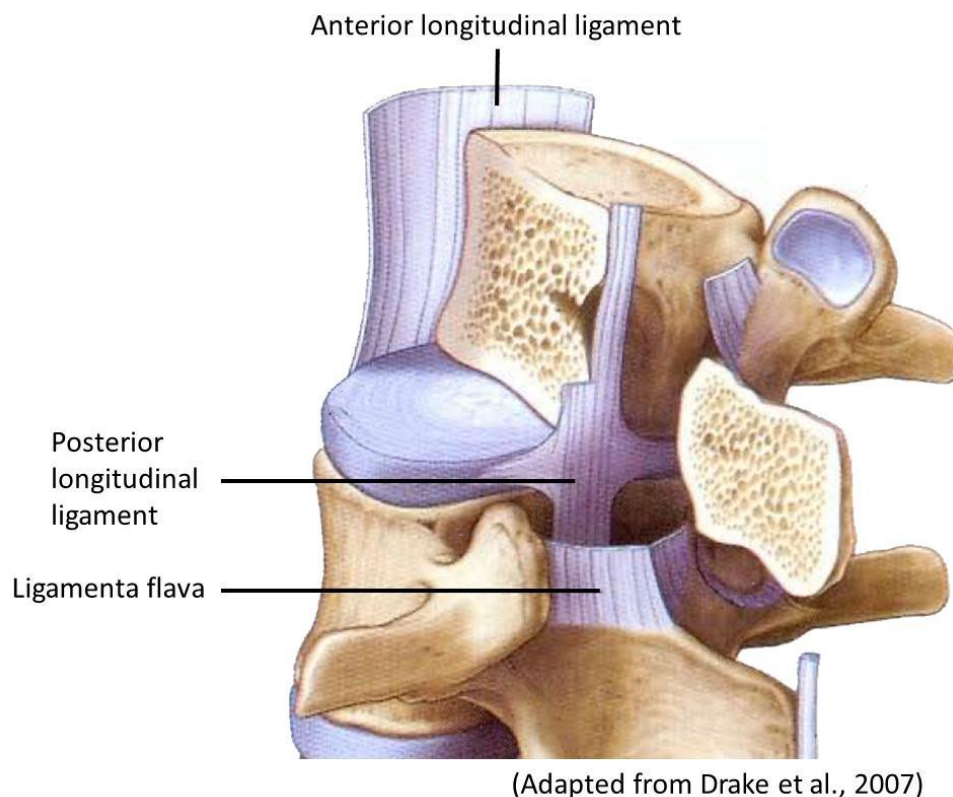
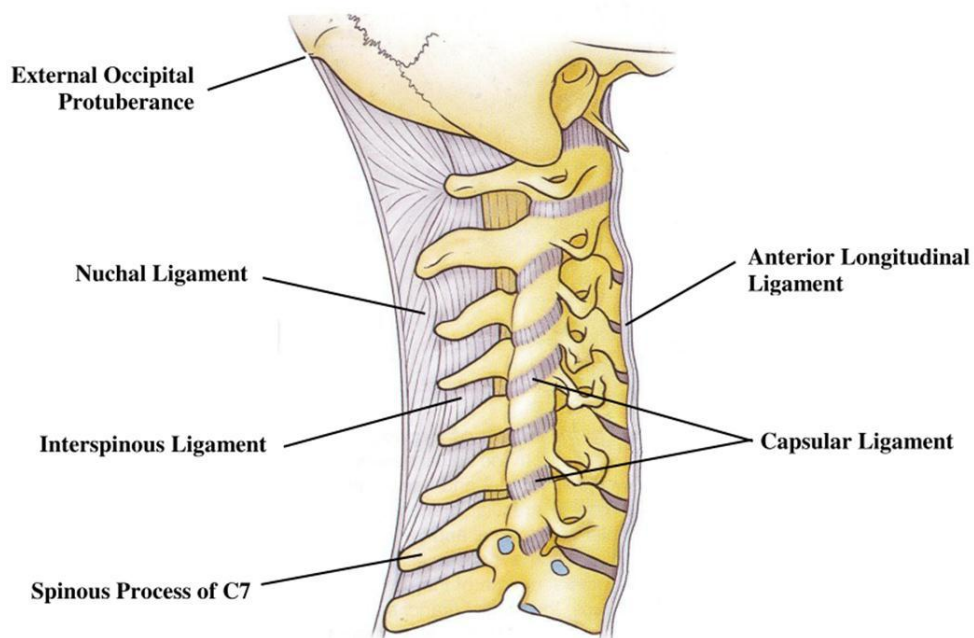


Figure 2-11: ALL, PLL, and Ligamenta flava

The ligamenta flava (LF) is a pale yellow elastic tissue which is aligned vertically and attaches the lamina of adjoining vertebrae. This long ligament is thinnest at the cervical region and thicker in the thoracic and lumbar regions. The ligamenta flava prevents injury in the intervertebral discs by supporting the vertebral column in sudden flexion. It also helps with keeping the proper curvature of the spinal column.

Interspinous ligaments (ISL) are weak fibrous that connect the spinous process of the vertebrae (Figure 2-13). The Supraspinous ligament is thicker and stronger than the ISL and it connects the tip of the spinous process from C7 to the sacrum and connects superiorly with the nuchal ligaments. Nuchal ligaments are thick and strong fibrous that are located in the back of the neck.



(Adapted from Moore & Dalley, 2006)

Figure 2-12: Lateral view of cervical spine ligaments

Capsular ligaments (CL) surround the facet joints with fibrous (Gray, 1918). These ligaments are attached between transverse processes of adjacent vertebrae. They exist in all sections of the cervical region of the spine and they stabilize all rotations.

Transverse ligament (TL) is a strong band which is located across the ring of the atlas and it keeps the anterior arch in contact with the odontoid process (Figure 2-13). Looking from the front, TL is concave in shape. It is thickest at the midline and it is attached to the lateral mass of the atlas on the

either sides. The superior crus (SC) connect the transverse ligament to basilar part of the occipital bone. Inferior crus (IC) connect TL to the posterior surface of the body of axis. The ring of the atlas is divided into two parts by the transverse ligament. The anterior section contains the odontoid process and the posterior side which is larger is used for transmission of accessory nerves and the medulla spinalis. The transverse ligament restricts the movement of the odontoid process and assists in preventing axial motion between the atlas and the axis.

The tectorial membrane (TM) is a broad and strong band which covers the odontoid process and its ligaments. The tectorial membrane is attached between the basilar groove of the occipital bone and the posterior surface of the body of C2. The posterior section of TM is in relation with the dura matter and the anterior section is with relation with TL and C1.

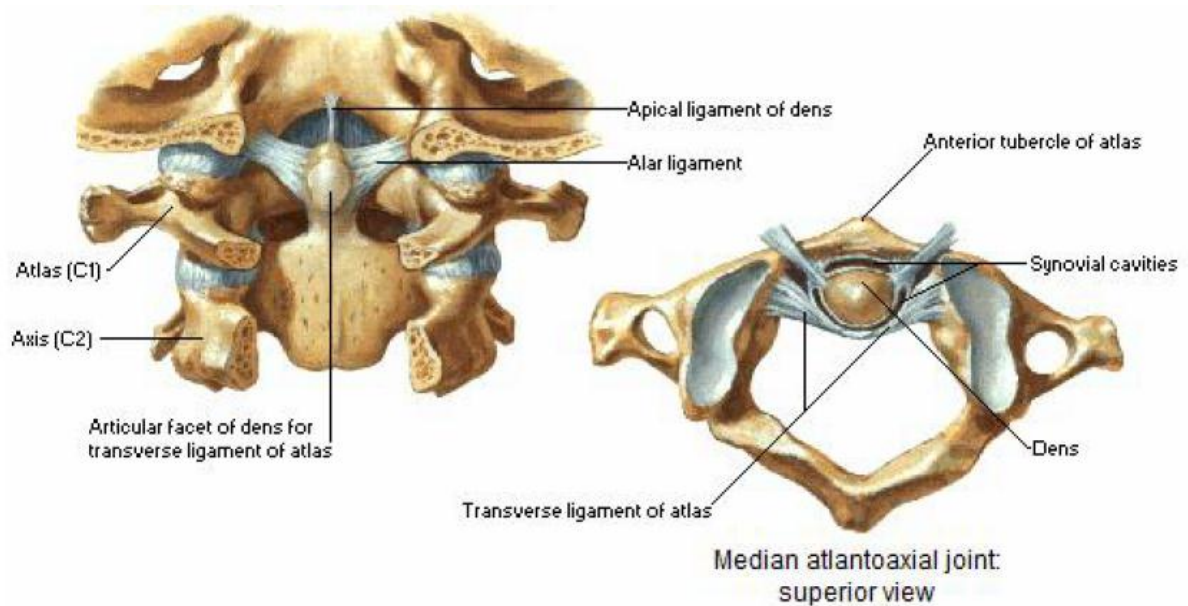
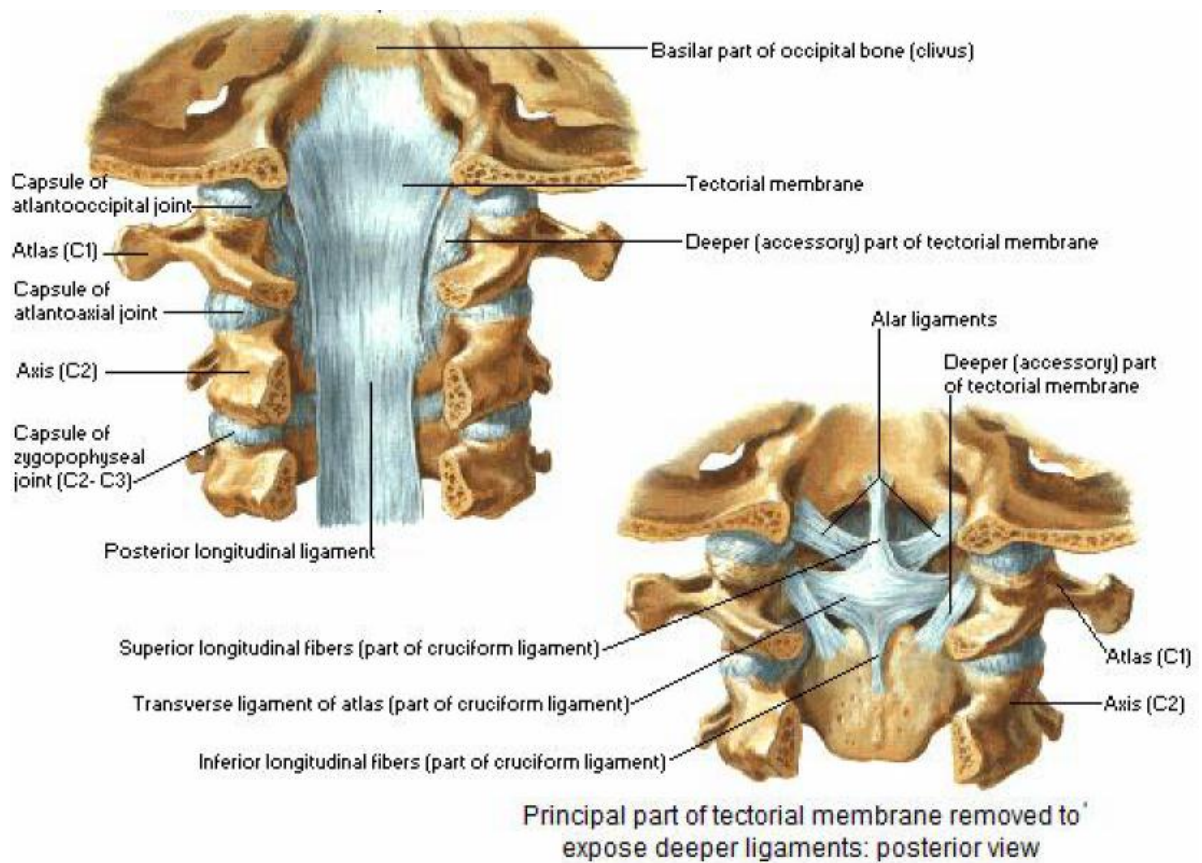


Figure 2-13: Upper cervical spine ligaments (Natter, 2006)

There are two pairs of alar ligaments. In the first pair, one side of the ligament is attached to the top of the odontoid process and the other side is attached on the medial sides of the condyles of the occipital bone. In the second pair, one side is attached to the inferior/lateral surfaces of odontoid process and the other is attached to the anterior lateral masses of the atlas. The purpose of the alar ligaments is to limit the rotation of the head. The apical odontoid ligament attaches from the tip of the odontoid process to the anterior margin of the foramen magnum. This ligament blends with anterior atlanto-occipital membrane and the superior crus.

Ligaments are a mixture of elastin and collagen fibres and do not support compressive loads (Yoganandan et al., 2001). Ligaments are viscoelastic and their mechanical response in tension is similar to a sigmoidal curve and can be represented by three distinctive regions (Chazal et al., 1985; Yoganandan et al., 1989; Shim et al., 2006; Figure 2-14). In tension, ligaments start to engage during the toe region. After engagement, the force of deflection response becomes linear. During the sub-traumatic region, the stiffness of the ligament softens to the point where it bears the maximum load and fibres then start to rupture. In a neutral posture, the ligaments in-vivo are preloaded to prevent joint laxity which is caused by the toe region of the ligament response (Nachemson & Evans, 1968; Viejo-Fuertes et al., 1998; Heuler et al., 2007).

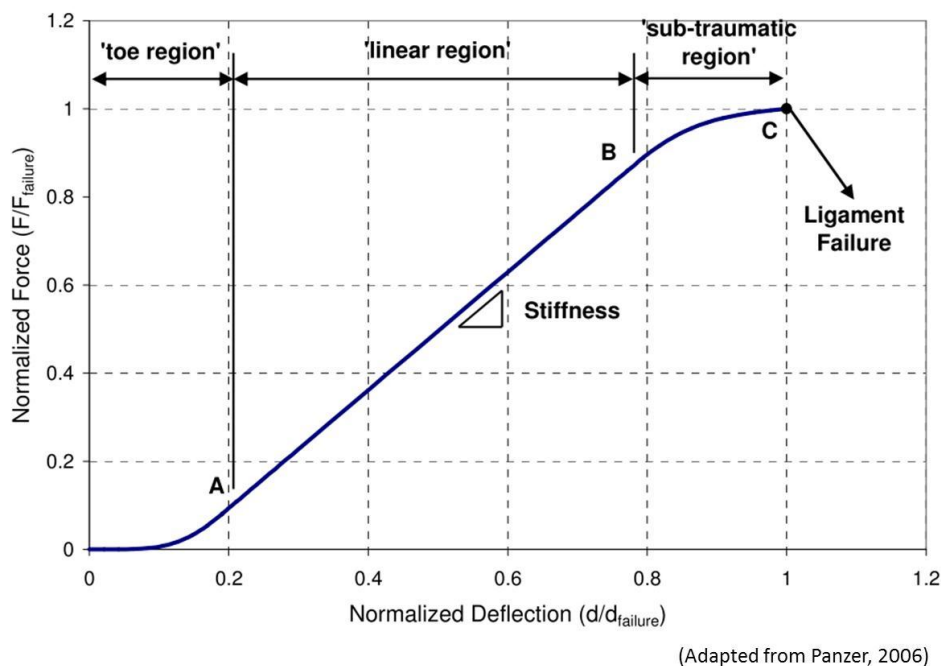


Figure 2-14: Normalized load-displacement curve for ligaments

2.6 Muscles

The main purpose of muscles is to enable movements in the cervical spine. There are 31 pairs of muscles in the neck (Knub & Myers, 1998) which are symmetrical about the sagittal plane. These muscles are divided into six groups which includes anterior, lateral, vertebral column, back, hyoid, and suboccipital muscles (Gray, 1918). The insertion point is referred to the side of the muscle that is attached to moving bone and the origin is the side that is attached to the stationary bone during contraction.

The anterior group consists of the Longus colli, Longus capitis, Rectus capitis anterior, and Rectus capitis lateralis (Figure 2-15). The Longus colli is narrow at either end and it is located between the third thoracic vertebra and the atlas, on the anterior surface of the vertebral column. This muscle is subdivided into a vertical portion, superior oblique, and inferior oblique. The vertical portion starts from front of the fifth, sixth, and seventh cervical vertebrae and first, second, and third thoracic vertebrae and insert into the front sections of the second, third, and fourth cervical vertebrae. The superior oblique portion starts from the transverse processes of the third to fifth cervical vertebrae and inserts into the anterior arch of the first cervical vertebrae. The inferior oblique portion is the smallest portion of Longus colli and it originates from the front of the first two or three thoracic vertebrae and is inserted into the fifth and sixth cervical vertebrae at the anterior of the transverse processes. The narrow side of the Longus capitis muscle originates from the anterior side of the transverse process of the third to sixth vertebrae and is inserted into the inferior surface of the basilar part of the occipital bone. The Rectus capitis anterior is a short muscle that originates from the anterior surface of the atlas and the root of its transverse process and is inserted into the inferior surface of the occipital bone. The Rectus capitis lateralis is also a short flat muscle. It originates from superior surface near the transverse process section of the atlas and is inserted into the jugular process of the occipital bone. The Longus capitis and Rectus capitis anterior help the head flex and restore it to its neutral position if it is bent in extension. These muscles can also assist in rotation of the skull. The Rectus lateralis is used to bend the head in a lateral direction. The Longus colli is used to rotate the cervical portion of the vertebral column.

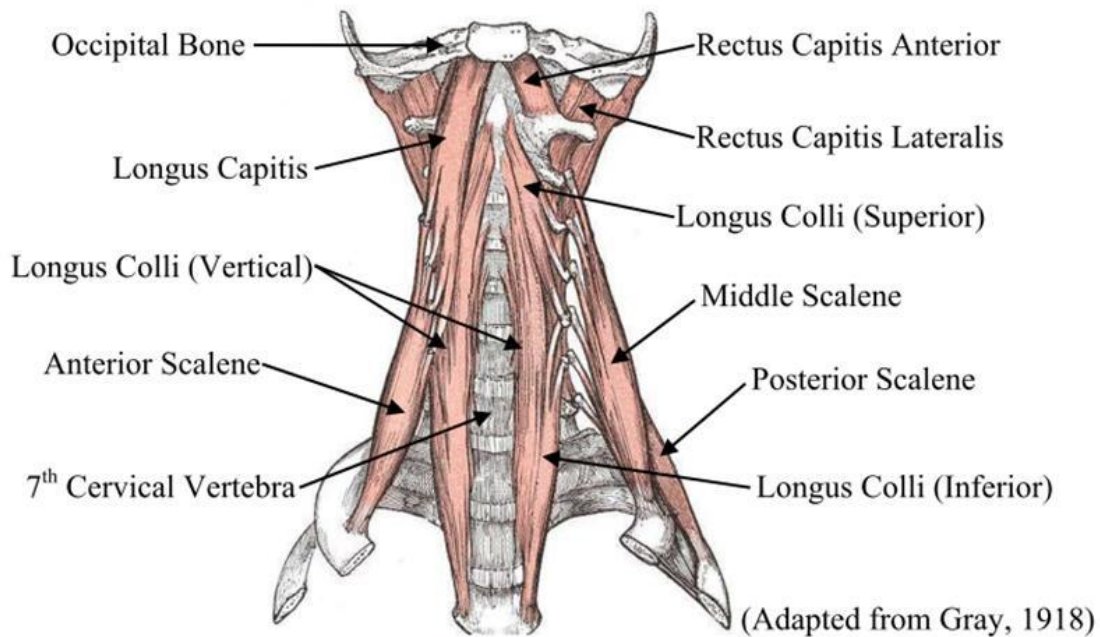


Figure 2-15: The anterior view of deep neck muscles

The lateral muscle group in the cervical spine includes the sternocleidomastoid and scalene anterior, middle, and posterior muscles (Figure 2-16). The sternocleidomastoid muscle is broad and thin on the either end and thick and narrow at the central area. It is attached to the sternum and clavicle at one end and to the mastoid process of the temporal bone of the skull at the other end. In combination with other muscles, the sternocleidomastoid muscle helps in head rotation, moving the head towards the shoulder, and flexing the cervical part of the vertebral column. The scalene muscles are inspiratory muscles and they originate from the transverse processes of the mid and lower cervical vertebrae and are inserted into the first two ribs.

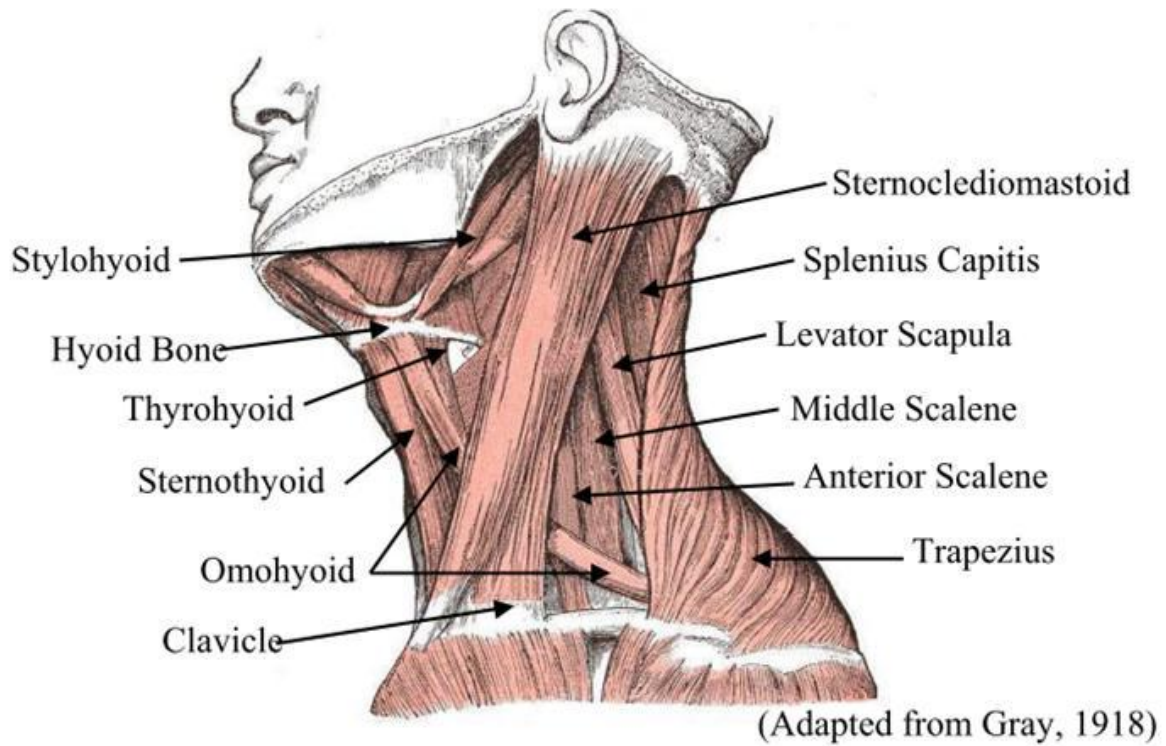


Figure 2-16: Muscles of the neck (Lateral view)

The vertebral column muscle group for the cervical spine consists of Trapezius, Rhomboideus minor, and Levator scapulae (Figure 2-17). These are muscles that connect the upper extremity to the vertebral column. These muscles have many functions but their main focus is producing the movement of the scapular towards the spine.

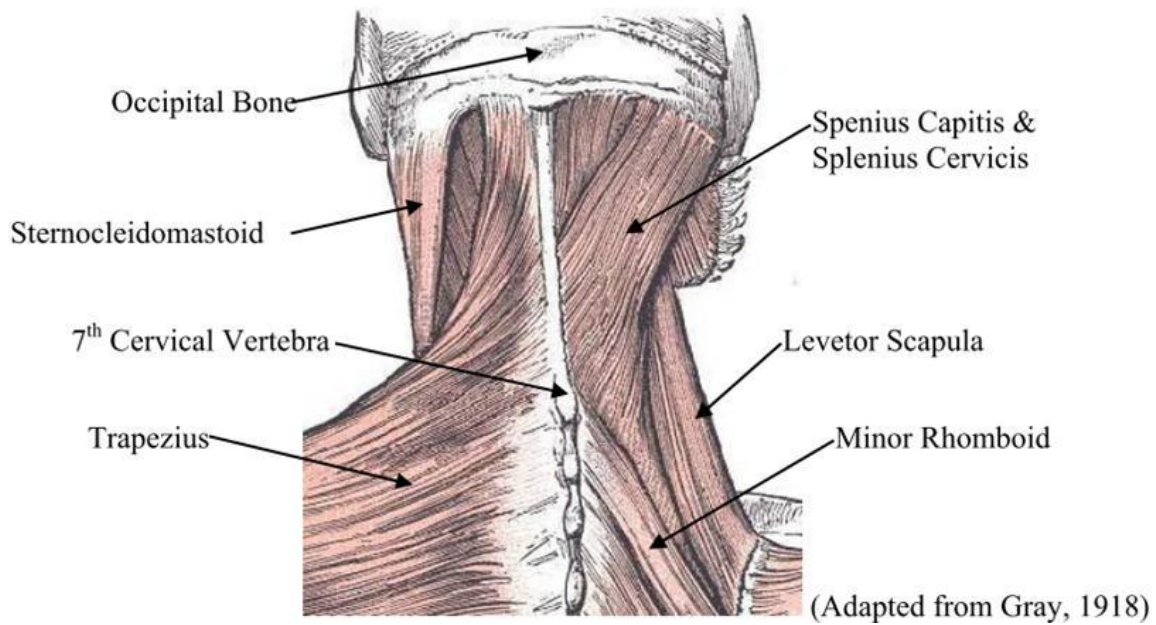


Figure 2-17: Posterior view of neck muscles

The hyoid muscle group is responsible for deglutition and it is subdivided into suprahyoid and infrahyoid muscles (Figure 2-16 and Figure 2-18). The suprahyoid group is consisted of diaphragm, mylohyoid, stylohyoid, and geniohyoid. The muscles in this group are positioned above the hyoid bone and are connected with the mandible or the skull. The infrahyoid group is consisted of omohyoid, sternohyoid, sternothyroid, and thyrohyoid muscles. These muscles are positioned inferior to the hyoid bone and are connected with the scapula, sternum, or the thyroid cartilage.

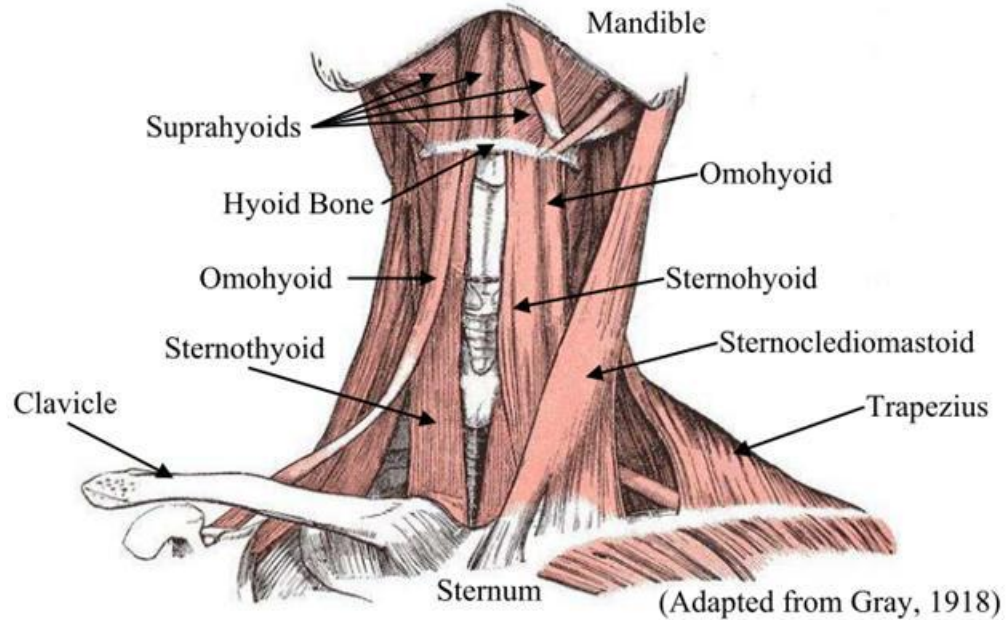


Figure 2-18: The anterior view of superficial muscles

The suboccipital muscle group consists of rectus capitis posterior major and minor and obliquus capitis inferior and superior (Figure 2-19). These muscles assist in the movement of the skull. The rectus capitis posterior minor and obliquus capitis superior muscles originate from atlas and are inserted into the occipital bone. The rectus capitis posterior major and obliquus capitis inferior originate from the spinous process of the axis and are inserted into occipital bone and the transverse process of the atlas, respectively.

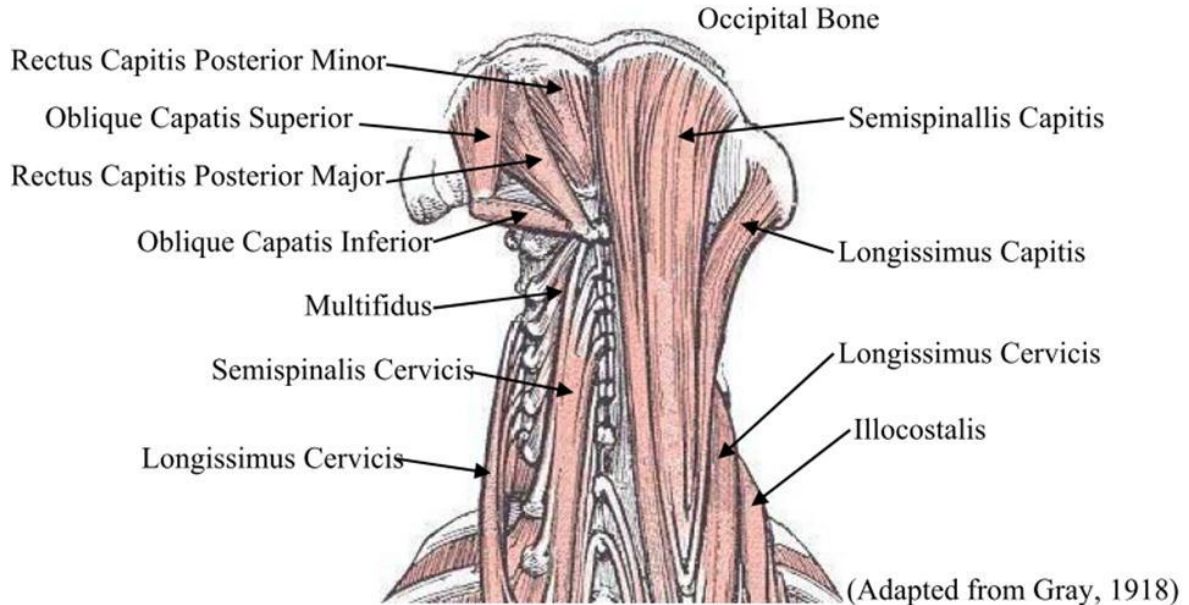


Figure 2-19: Deep muscles of the neck

The back muscles in the cervical spine are used for extension motion and consist of the splenius capitis, splenius cervicis, semispinalis, multifidus, longissimus, and iliocostalis muscles. Splenius capitis originates from the lower half of ligamentum nuchae, spinous process of upper three of four thoracic vertebrae, and the spinous processes of the seventh cervical vertebra. This muscle is inserted into mastoid process of temporal bone and into the rough surface of the occipital bone. The splenius cervicis originates from the spinous process of the third to sixth thoracic vertebrae and it is inserted into the posterior tubercles of the transverse process of the upper two or three cervical vertebrae. The semispinalis dorsi and cervicis originate from the transverse process the thoracic vertebrae and are inserted into the spinous processes of the cervical vertebrae. The semispinalis capitis originates from tendons at the tip of the transverse process of the upper six or seven thoracic and the seventh cervical vertebrae and the articular process of the C4-C6 vertebrae. The semispinalis capitis is inserted into the occipital bone. In the cervical region, the multifidus muscles originate from the articular process of the C4-C7 vertebrae and are inserted into the spinous process of one of the vertebrae above. The longissimus cervicis originate from the transverse process of the upper four or five thoracic vertebrae and are inserted into the transverse process of the second to sixth cervical vertebrae. The longissimus capitis originates in the similar region as the longissimus cervicis and it is inserted into the mastoid process. The iliocostalis cervicis originates from angles of third to sixth ribs and is inserted into the transverse process of the fourth to sixth cervical vertebrae.

Chapter 3

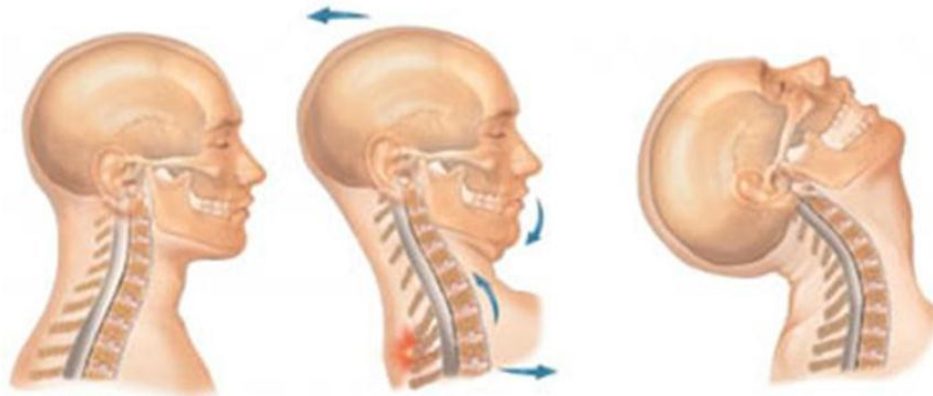
Background

The following section provides a background on previous research on whiplash, injury sites, and out-of-position injury. This section starts by providing information on whiplash injuries. Types of experiments that are used for finding relevant information to predict injury and tissue material properties are discussed. A summary of methods which are used to measure injuries and injury locations is presented. Finally, studies on out-of-position neck injury and some of the Finite Element models that have been developed will be covered.

3.1 Whiplash Injuries

The term whiplash was first introduced by Harold Crowe in 1928 at the Western Orthopaedic Association Meeting (Crowe, 1928). The description was based on an acceleration extension flexion injury during the First World War relating to take-offs from aircraft carriers. Gay et al. (1953) used the term whiplash injury for the first time in a medical journal in 1953 and from 1951 to 2009 there have been approximately 6200 scientific articles on the term whiplash (Kaale, 2009).

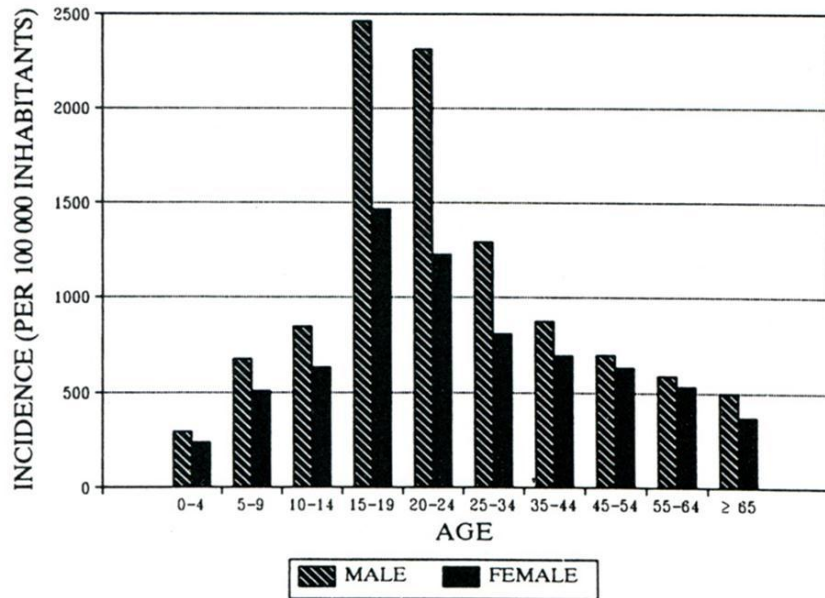
The definition of whiplash has been unclear and literature has used different descriptions in various studies. Spitzer et al. (1995) described whiplash as an acceleration-deceleration mechanism of energy transfer to the neck. Whiplash can occur during driving, in rear-end or side-impact collisions, or during other mishaps. The impact could result in bony or soft-tissue injuries which are referred to as whiplash injuries. Whiplash injuries could lead to different clinical manifestations which are referred to as Whiplash-Associated Disorders (WAD). Kaneoka et al. (1999) defined three distinct patterns of cervical spine motion after rear-end collisions. In the first stage the seat pushes the torso in the anterior direction while the head remains stationary, causing flexion in the neck. This motion causes an S-shaped motion in the neck, shown in Figure 3-1. The skull then starts rotating in extension and the neck might become subjected to hyperextension before it moves into the flexion motion again. The duration of whiplash can vary depending on the severity of the impact.



(Adapted from Creemore Chiropractic)

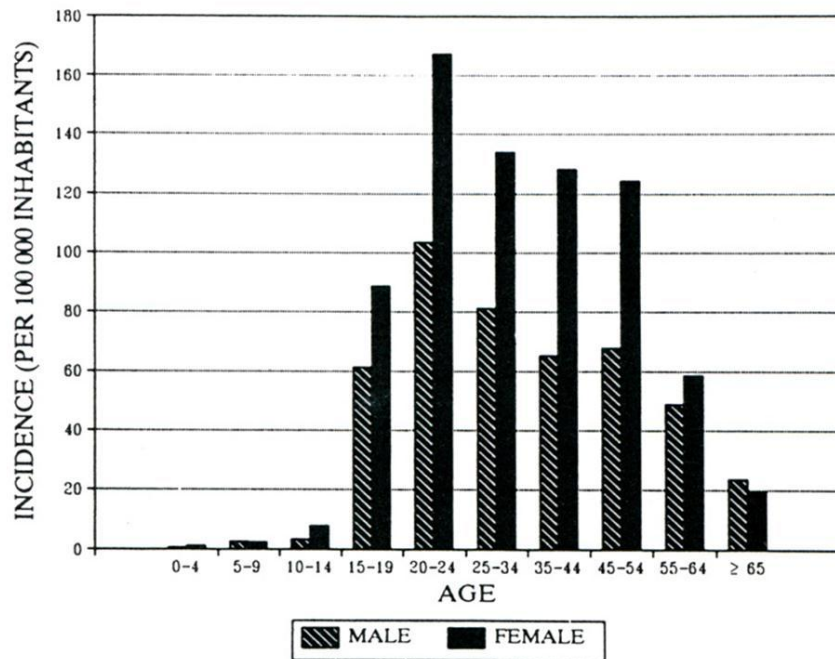
Figure 3-1: Cervical spine motion after impact

Spitzer et al. (1995) reported that in the province of Quebec, approximately 5000 whiplash cases accounted for 20% of annual insurance claims. In addition, the average period for compensation had increased from 72 to 108 days between 1987 and 1989 (Girard, 1989; Giroux, 1989). It was also reported that whiplash injuries represented 68% of insurance claims in British Columbia and 85% of the claims in Saskatchewan (Giroux, 1989; Dieck et al., 1985). Figure 3-2 shows the distribution of automotive injuries in the province of Quebec in 1987. The figure shows an almost normal distribution with the peak around the ages of 15-24. The data also shows that the rate of motor vehicle injuries was higher for the male population. Figure 3-3 shows that most of whiplash injuries in Quebec during 1987 happened to individuals between the ages of 20-54 with the peak being around the age of 20-24. This figure also shows that females had a higher number of reported whiplash injury as compared to males which is consistent with the study by Quinlan et al. (2004).



(Adapted from Spitzer, 1995)

Figure 3-2: Incident rate of all motor vehicle injuries by age and gender in Quebec (1987)



(Adapted from Spitzer, 1995)

Figure 3-3: Incidence rate of compensated whiplash injury by age and gender in Quebec (1987)

Whiplash injuries have a large impact on society. The study by Spitzer et al. (1995) showed that the duration of absence from normal activities for more than 50% of the 2810 subjects that had undergone whiplash injuries during 1987 in Quebec was longer than 28 days (Figure 3-4). The total cost for insurance companies to replace the regular income of all subjects that had undergone whiplash injuries was approximately thirteen million dollars. The addition of medical expenses brought this number to 18.3 million dollars during that year. This number is very small when compared to the overall cost in the United States and Europe. The total annual cost related to neck injuries had been estimated to be 4.5-29 billion dollars in the United States and 5-10 billion Euros in Europe (Kinberger, 2000; Freeman, 1997; Schmitt et al., 2004).

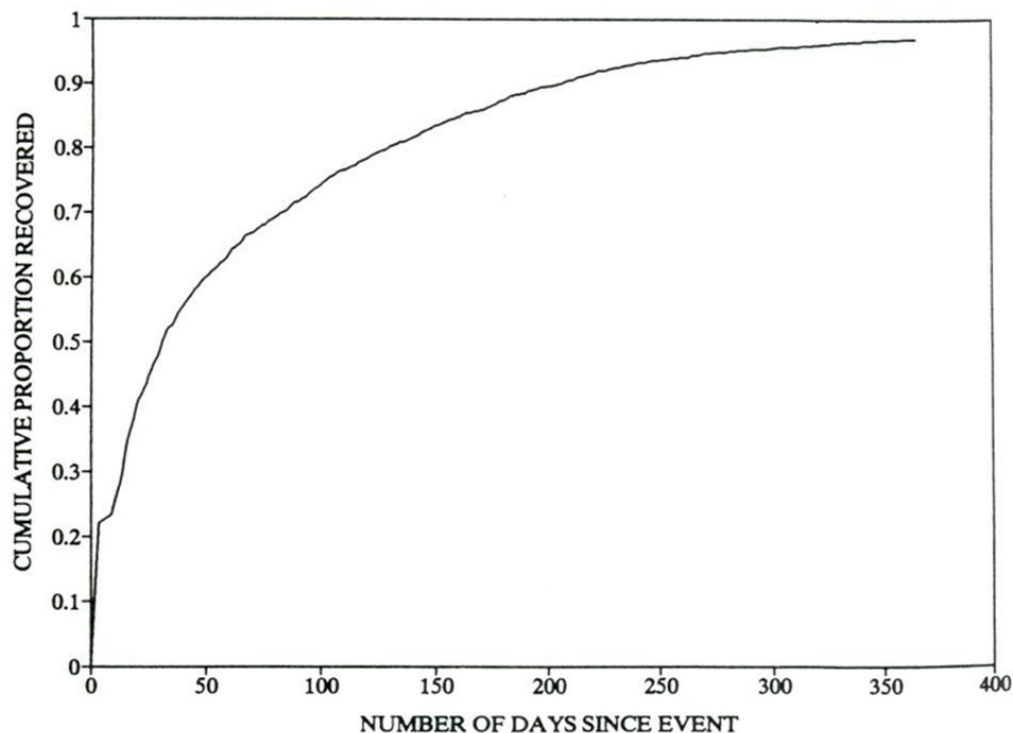


Figure 3-4: Overall one year cumulative return to activity curve for cohort of whiplash subjects

Whiplash symptoms are divided into two time frames. Acute injuries are those that last for a short period of time while chronic injuries last longer and may even last forever. Chronic pains were defined as lasting longer than 3-6 months by the European Foundation of International Association for the Study of Pain (Niv & Devor, 2007). Whiplash symptoms include neck pain, neck stiffness, headache, shoulder pain, arm pain/numbness, paresthesias, nausea, weakness, dysphagia, visual and auditory disturbances, and dizziness (Haldorsen et al., 2003; Klein et al., 2001; Norris & Watt, 1983).

Spitzer et al. (1995) classified the whiplash-associated disorders into five sets of groups (Table 3-1). It has been shown that a significant number of whiplash patients exhibit Abbreviated Injury Scale (AIS) 1 injuries (Ono & Kanno, 1996) which are classified as minor injuries and can often be associated with grade 1-3 disorders. Study by Norris & Watt (1983) showed that whiplash symptoms can be apparent on patient follow-ups that had been conducted approximately two years after the incident (Table 3-2). In this study, the patients were put in three distinctive sections. The group number described the clinical presentation of grade numbers which were classified by Spitzer et al. (1995). The results showed that neck pain at follow up was higher in grade 2 and 3 classifications. It is also apparent that whiplash injuries could cause symptoms that can last a long time. One of the most important findings of this study was that approximately 44% of patients with no physical signs of damage were suffering from neck pain two years after the incident.

Table 3-1: Clinical classification of Whiplash-Associated Disorders

Grade	Clinical presentation
0	No complaint about the neck No Physical sign(s)
1	Neck complaint of pain, stiffness or tenderness only No physical sign(s)
2	Neck complaint and musculoskeletal sign(s) ^a
3	Neck complaint and neurological sign(s) ^b
4	Neck complaint and fracture or dislocation

a Musculoskeletal signs include decreased range of motion and point tenderness
b Neurologic signs include decreased or absent deep tendon reflexes, weakness,

(Adapted from Spitzer, 1995)

Table 3-2: Symptoms after whiplash injuries of the neck at initial presentation and at final follow-up (percentages)

Symptoms	Percentage of patients					
	Group 1		Group 2		Group 3	
	At presentation	At final follow-up	At presentation	At final follow-up	At presentation	At final follow-up
Neck pain	100	44	100	81	100	90
Headache	48	37	78	37	80	70
Dysphagia	18.5	0	9.5	0	30	0
Paraesthesiae	33	37	43	29	100	60
Weakness	15	0	9.5	0	50	0
Visual symptoms	7.5	18.5	0	9.5	30	10
Auditory symptoms	7.5	11	0	14	30	20
Dizziness	0	0	4.5	0	10	0

Group 1: Neck complaint of pain, stiffness or tenderness only, no physical sign(s)

Group 2: Neck Complaints and musculoskeletal sign(s)

Group 3: Neck complaint and neurological sign(s)

(Adapted from Norris & Watt, 1983)

The usage of seatbelts has not proven to reduce the chance of whiplash injuries. Allen et al. (1985) studied the effect of seatbelt legislation on injuries by car occupants and found that wearing seatbelts increased the chance of neck injuries. However, seatbelts decrease the chance of head injuries and overall fatality rates and have been found to be important for the occupants. The effect of headrests on preventing neck injuries has also been found to be challenging. In theory, the existence of headrests should limit extension of the head in rear impacts and reduce the chance of injury. A study of 106 patients by Morris (1989) concluded an increase in the incidence of whiplash with the unrestrained neck. Other studies have shown that there is no significant improvement on whiplash injuries due to headrests (Hildingsson & Toolanen, 1990; Olney & Marsden, 1986). The study by Viano & Gargan (1996) showed that most vehicle occupants do not set the position of their headrests to prevent injury. The result of this study, along with the experiment by Ivancic et al. (2009) showed that proper positioning of the headrest could reduce the chance of whiplash injuries.

Whiplash injuries are problematic to society. These injuries can happen to all genders and all ages. Whiplash-associated disorders lead to a huge burden to the insurance companies as the cost exceeds billions of dollars every year. Up until now, there is no way to prevent this type of injury and many

patients that are exposed to whiplash will never recover from the chronic symptoms. It is very difficult to identify physical damage from whiplash injuries and medical imaging cannot detect damage in many cases. The study of whiplash will continue in hope of reducing and eliminating such problems in the future.

3.2 Experimental Methods

The mechanical properties and the tolerance limits of the human body, found through experimental methods, help with the creation and analysis of human models. The methods that are used to find mechanical properties of the cervical spine include human volunteers studies, animal studies, cadaver studies, segment studies, and isolated ligament experimentations. Human volunteer testing represents the most realistic results as compared to the other methods. The limitation of human volunteers is that the experimentation should be performed at a low severity and injury cannot be predicted through this method.

Animal testing can be used to find injury levels in the spine. An example includes using pigs in experiments and relating their data to a three year old child dummy (Mertz & Weber, 1982; Prasad & Daniel, 1984). Since many animals can no longer be subjected to severe impact, animal studies are more challenging to interpret in recent years. The experimental data previously gathered is still being used in neck injury evaluation

Cadavers or Post-Mortem Human Subjects (PHMS) are used to study the *in situ* mechanics of the neck. PHMS provide a very similar response to live humans; the only disadvantage being the lack of active musculature. Unlike human volunteers, cadaver specimens can be put through severe impact scenarios to predict ligament failure. PHMS are also used in finding the range of motion in the neck. Cadaver specimens are vital in finding interactions between the cervical spine vertebrae and validating numerical models.

Segment level testing are usually performed by fixing the inferior vertebra and displacing the superior vertebra. Segment studies are simpler to perform than PHMS and they allow scientists to get an understanding of how each segment of the cervical spine behaves. Segment studies are performed *in vitro* and even though they cannot fully represent *in vivo* behaviour, they are extremely helpful in understanding the localized behaviour of in the neck.

The properties of ligaments is usually measured based on the force and deformation in the ligaments. This is because the cross sections of ligaments are complicated to measure and might vary along the length, making it unsuitable to report stress values. Specific ligaments are dissected and put through tension tests. Varying the speed of the tension test can allow for extraction of rate effects within the ligaments. This is extremely important because ligaments exhibit viscoelastic properties. The different testing techniques have both strengths and weaknesses. The knowledge from the experimental data can be used to model biofidelic human structure with improved kinematics as compared to crash dummies.

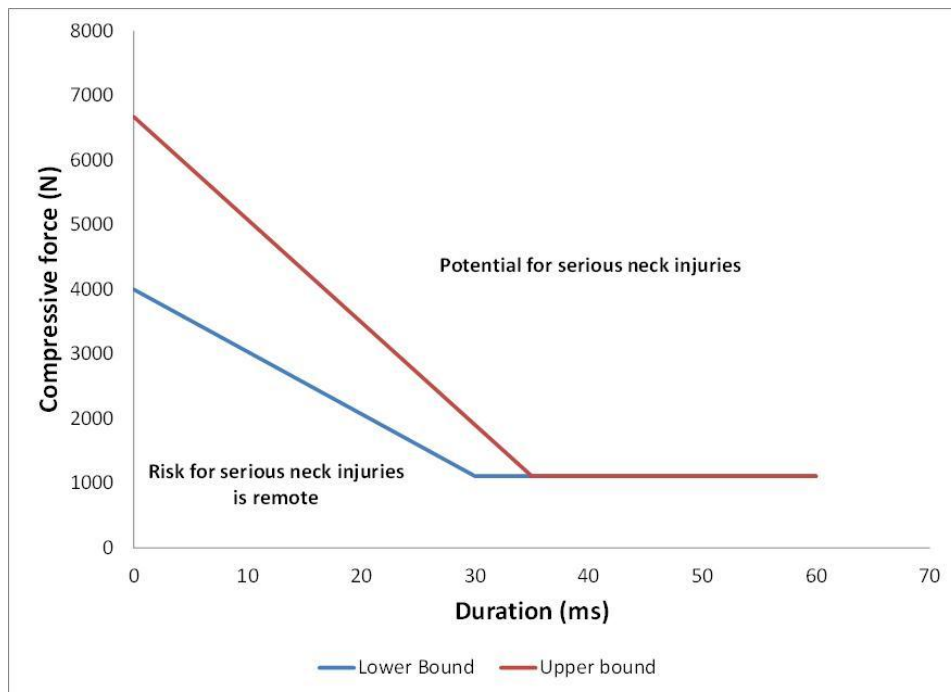
3.3 Injury Evaluation Criteria

This section provides a summary of some of the most typical techniques that are used in the evaluation of the severity of injury. Methods that will be covered include N_{ij} criterion, NIC criterion, IV-NIC, and tissue strains. The advantages and disadvantages of these criteria will be discussed.

3.3.1 N_{ij} Criterion

The N_{ij} criterion was proposed by the US National Highway Traffic Safety Administration (NHTSA) (Klinch et al., 1996). This criterion was developed for frontal impacts and included as part of Federal Motor Vehicle Safety Standards and Regulations (FMVSS 208). This section provides a background to the development of this criterion.

The axial compression tolerance levels for the N_{ij} criterion was developed by Mertz et al. (1978). The hybrid III 50th percentile male dummy was used to investigate neck reaction loads that were produced by spring-loaded tackling blocks that caused serious head and neck injuries in high school football players. In the test, the load was applied on top of the hybrid III dummy. The configuration was chosen to minimize head rotation and to produce a maximum value of neck compression force for the given impact velocity. The result of the study in Figure 3-5 shows the compressive force with the duration which may cause neck injury.



(Adapted from Mertz, 1978)

Figure 3-5: Injury tolerance curves for axial neck compression force

Nyquist et al (1980) used Hybrid III 50th percentile male dummy to develop tolerances for tension and shear loads. The dummies were put into a real-world collision situation and the responses were correlated with field injuries. The limits for shear and tension were reported as 3000N and 3300N, respectively. These values were unique to the Hybrid III dummies.

Mertz & Patrick (1971) used volunteers and cadaver subjects to find the tolerance levels for flexion and extension bending moments. The volunteers were tested to pain threshold and the cadavers extended the limits for serious injuries. In extension, ligamentous damage occurred in one of the cadavers at an equivalent 50th percentile male level moment of 57 Nm. In flexion, no injury was produced and the maximum measured value of 190 Nm was selected as the limit. These tests were based on human subjects rather than dummies and therefore the biofidelity of the dummies during bending moments become important in interpreting the results.

In a study done by Prasad and Daniel in 1984, the authors judged that based on a number of anatomical and developmental factors, pigs can represent a 3-year-old dummy. Measured responses in the child dummy were correlated with injuries sustained by the surrogate. Prasad and Daniel

concluded from their results that axial tension loads and extension bending moments should be linearly combined to form a composite neck injury indicator. Critical values proposed at the time for tension and extension of a 3-year-old dummy were 2000 N and 34 Nm, respectively. These data were later reanalyzed and the critical values were changed to 2120 N for tension and 26.8 Nm for extension. A scaling method was used to come up with the critical values for other types of dummies (Figure 3-6; Eppinger, 1999). If the combinations of the axial and bending moments lie outside the gray area of the figure, then there is potential for injury.

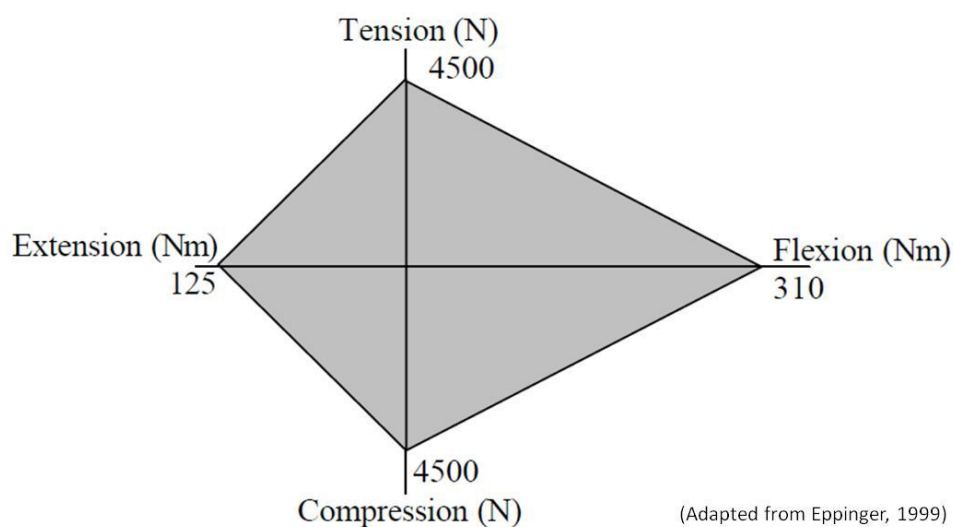


Figure 3-6: Neck injury criteria for 50th percentile male dummy

The data in Figure 3-6 were normalized and the N_{ij} was calculated as the combination of the axial and bending moments on the neck (Equation 3-2). In this equation the subscript i represents tension or compression and j represents extension or flexion. F_z represents the tension or compression load, M_y represents extension or flexion moments, and subscript int is the critical intercept value of loads which are shown in Figure 3-6. The critical intercept values based on FMVSS 208 for 50th percentile dummy in tension, compression, flexion, and extension are 6806 N, 6160 N, 310 Nm, and 135 Nm respectively (National Highway Traffic Safety Administration, 2011).

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}} \quad \text{Equation 3-1}$$

The N_{ij} injury curves are shown in Figure 3-7. The values that are used to curve fit data for AIS 3+ come from Mertz et al. (1982) and Prasad et al. (1984). In these experiments porcine subjects were

placed in the same impact conditions as a 3-year-old child dummy and the probability of their injury had been reported. The rest of the curves were based on injury rates predicted using N_{ij} calculations from experimental dummy test data and real world injury rates (Eppinger et al., 1999); hence, the difference between the curves. One of the limitations of curve fitting the data is the existence of risk of injury at zero load (Figure 3-7). Another disadvantage is that this method has not been derived for out-of-position scenarios.

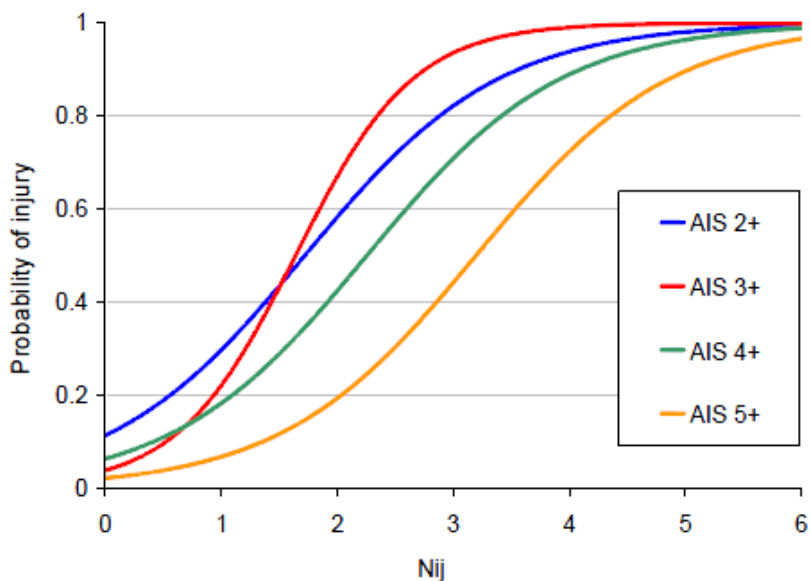


Figure 3-7: N_{ij} risk curves (Eppinger et al., 2000)

3.3.2 Neck Injury Criterion (NIC)

The neck injury criterion (NIC) is often used to evaluate injury in rear impacts. This criterion was developed by Bostrom et al. (1996) and assumes that the sudden change of the fluid flow inside the fluid compartments of the cervical spine are related to injury. NIC is evaluated by the acceleration and velocity difference between the centre of mass of the head relative to the first thoracic vertebra in the anterior-posterior direction (Equation 3-2).

$$NIC(t) = 0.2a_{rel}(t) + v_{rel}^2(t)$$

Equation 3-2

The value of $15 \text{ m}^2/\text{s}^2$ was set as the threshold above which significant risk of minor injury would be inherent. There is a significant error introduced into the equation once the head is going through extension and is no longer parallel to the T1 vertebra. The equation uses kinematics instead of fluid flow on pressure and it is unable to predict injury in out-of-position scenarios.

3.3.3 IV-NIC

The Intervertebral Neck Injury Criterion (IV-NIC) hypothesizes that intervertebral motion beyond the physical range of motion may cause soft tissue injuries (Panjabi et al., 1999). IV-NIC has a few advantages over the previously described methods for the evaluation of neck injury. The major advantage lies in the fact that this method can be used in flexion, extension, lateral bending, and axial bending. This method can also predict the severity, time, intervertebral level, and mode of cervical spine soft-tissue injury (Ivancic et al., 2006b). The IV-NIC value is evaluated based on Equation 3-3. The dynamic intervertebral rotation is divided by the quasi static physiological range of motion. The subscript i represents the intervertebral level and j represents the plane of motion (Panjabi et al., 1999). IV-NIC values above one are hypothesized to cause soft tissue injuries.

$$IV - NIC_{i,j}(t) = \frac{\theta_{dynamic,i,j}(t)}{\theta_{physiological,i,j}} \quad \text{Equation 3-3}$$

A bench-top apparatus was used by Ivancic et al. (2005a) to simulate 4, 6, 8, and 10 g frontal impacts on biofidelic whole human cervical spine model with muscle force replication. In this study soft tissue injury was defined as statistically significant increase in neutral zone, range of motion, or both during the frontal impact, above the baseline values obtained after 2 g dynamic preconditioning. It was concluded IV-NIC was an effective tool in determining soft tissue neck injury. Similar setups have been used to evaluate IV-NIC in rear and side impacts (Panjabi, 2005a; Panjabi, 2005b). The author concluded that IV-NIC had correlated well with neutral zone and total range of motion increases in the cervical spine.

Hybrid III and BioRID II are two common test dummies that are used in prediction of injury in rear impact crashes (Schmitt et al., 2002; Zuby et al., 1999). Between these dummies, the BioRID II has been shown to be more biofidelic in rear impact scenarios when validated against human volunteers and cadaver results (Davidsson et al., 1999; Kim et al., 2001; Linder et al., 2002; Philippens et al., 2002; Siegmund et al., 2001a). However, the BioRID II's hinge system does not allow it to move in axial or lateral rotations. Hybrid III dummy has similar limitations and therefore

IV-NIC cannot be used with these dummies. Designing a more biofidelic cervical neck in dummies will be advantageous in predicting injury in three-dimensions.

The ability of this method to predict injury in situations where the head is not at a neutral posture makes this method a good candidate for numerical model analysis. One of the disadvantages of this criterion includes the fact that during the validations, muscle forces were replicated by cable and spring systems instead of cadaver passive muscles. This method may be able to predict which level of the cervical spine is injured, but it cannot provide the exact tissue that has been injured. There have been no physical injuries reported during the validations of this criterion.

3.3.4 Tissue Strain

Automobile collisions may result in soft tissue injuries in the neck and may result in chronic symptoms such as headache and neck pain (Barnsley et al., 1994). Tissue strains are one way to evaluate the possibilities of injury during collision. Ligament strains above failure threshold values reported in the literature can be used to predict soft tissue injuries in the neck. Hybrid III dummies do not include soft tissues and therefore this method is not currently used in automotive industries for crash tests. Ligaments in the cervical spine are very small and therefore it is very difficult to observe their strains during *in situ* full cervical spine experimentations. Finite Element models provide a good tool to predict intrinsic biomechanical responses in the neck, such as ligament strains, facet joint strains or stresses, and end plate stresses (Yoganandan et al., 1996a).

The strain on the ligaments can simply be calculated by the change of length over the initial length of the ligament. Panjabi et al. (1999 & 2004a) and Ivancic et al. (2004) have investigated the length of the ligaments in the cervical spine. In numerical models, the change of length of the ligaments during impact can be evaluated using the simulation software. This data can be used to compute the strains in the ligament which can later be compared to injury threshold which have been previously reported in the literature. Winkelstein et al. (2000) and Siegmund et al. (2001b) had studied failure in the capsular ligaments and Yoganandan et al. (2001) had experimented on the failure on other cervical spine ligaments in the neck. Studies that had been performed on the failure of ligament of the upper cervical spine can also be used to predict injury in this region of the neck.

3.4 Injury Locations

The sources of initial symptoms from whiplash are often uncertain (Binder, 2007). The anatomical sites that have been assumed to relate to pain and symptoms in the neck include facet joints, dorsal root ganglia, spinal ligaments, intervertebral discs, vertebral arteries, and neck muscles (Siegmund et al., 2009). This section provides an overview of how these anatomical sites can be responsible for whiplash symptoms.

Literature suggests that the most common source of neck pain is caused by the cervical facet joints (Aprill & Bogduk, 1992; Barnsley et al., 1994). Anatomic and histologic studies have identified pain fibers in the facet joints which carry nociceptive signals and therefore any injury in the joints could result in pain (Cavanaugh, 2000; Cavanaugh et al., 1989; Giles & Harvey, 1987; Inami et al., 2001; Kallakuri et al., 2004; McLain, 1994; Ohtori et al., 2003). The pinching of the synovial fold and excessive strain of the capsule were identified as two mechanisms of facet joint injury (Siegmund et al., 2009). Ono et al. (1997) and Kaneoka et al. (1999) observed high compressive force on the facet joints during whiplash and proposed that the forces could lead to the pinching of synovial fold. There has been no other study which supports this idea and therefore the evidence for the cause of pain from the pinching of the synovial fold is incomplete. Excessive peak strains of 29-40% have been observed on C6-C7 which is larger than the average 6 percent observed during normal bending (Panjabi et al., 1998a; Pearson et al., 2004). Partial rupture of facet capsules have been observed prior to soft tissue failure in tension and shear loading of the facet joint (Siegmund et al., 2001b; Winkelstein et al., 2000). The excessive elongation on the capsule could be a potential for injury. Joint distraction during impact could also cause distraction in the fiber arrangements in the capsular ligaments which could lead to pain.

The vertebrae in the neck are connected through ligaments. Ligament damage may cause acute pain and chronic spinal instabilities (Siegmund et al., 2009). Ligaments also have mechanoreceptive and nociceptive nerve endings and their damage may lead to abnormal sensory signals and a decrease in neck mobility and proprioception (Panjabi, 2006). Ligament rupture could be caused when it is stretched beyond its physiological limits. Pearson et al. (2004) tested cervical spine cadaver models with muscle force replication in rear impact and concluded that capsular ligaments in the lower cervical spine, especially in C6-C7 are at risk of injury. Ivancic et al. (2004) used a similar setup and observed that anterior longitudinal ligaments experience high strains in the lower cervical spine with

the highest strain at C6-C7 during 8 g rear-impact simulation. In frontal impact, Panjabi et al. (2004) observed significant increases in the strain at the supraspinous and interspinous ligaments and the ligamentum flavum. During 10 g frontal impact, Panjabi observed high strains of capsular ligaments but the posterior longitudinal ligaments were not at risk of injury. In out-of-position, Kaale et al. (2005) has observed higher than normal deformation in the alar and transverse ligaments with MRI studies. Cadaver studies up to 8 g have not been able to confirm injuries in out-of-position scenarios.

The results of ligament studies have shown that ligaments in the cervical spine are at a risk of injury during impact and damage to ligaments may cause instabilities on the cervical spine, resulting in chronic pain. As discussed previously, facet joints are the most common site of injury and the distraction of capsular ligaments that connect these joints could be very influential in chronic pain. The cadaver study by Ivancic et al. (2008) revealed that capsular ligaments that were previously exposed to whiplash had significantly higher elongation than those of control ligaments at tensile forces of zero and five Newtons. Ivancic concluded that the increased laxity could be a cause of instability and chronic pain in whiplash patients. The physiological range of strains for the ligaments, observed in the literature, could be used along with numerical models to predict the sites of injury in the cervical spine due to impact.

The dorsal root ganglion is the combination of rootlets from the anterior and posterior of the spinal cord. Cell bodies of sensory neurons are contained inside the dorsal root. Damage to this structure can cause impaired local sensory processing (Greening et al., 2005; Kasch et al., 2001; Scott et al., 2005; Sterling et al., 2003&2006; Sterner et al., 2001). Extension, flexion, and lateral bending can cause pressure gradients in the cervical spine. In low level impacts, blood flow can compensate for the pressure changes. However, during whiplash motion, the pressure gradient between inside and outside the spinal canal can cause loading to the dorsal root ganglia and may lead to WAD (Siegmond et al., 2009). Whiplash experiments on anesthetized pigs (Svensson et al., 2000) had shown transient pressure drop inside spinal canals and nerve cell membrane dysfunction in pigs with dynamic loads. Eichberger et al. (2000) had observed transient pressure gradients in cadavers but did not perform histological investigation on the dorsal root ganglia. There is limited knowledge on whether or not the transient pressure gradients could cause damage to the dorsal root ganglia in humans. Another potential for injury could be caused by the deformation of the nerve roots. Panjabi et al. (2006a) and Tominega et al. (2006) found that C5/C6 intervertebral foramen in cervical spine cadavers narrowed

by 1.8mm during simulated rear impacts and concluded that the intervertebral foramen narrowing could cause a high risk of dorsal root ganglia damage in the lower cervical spine.

Vertebral arteries supply blood to the brain, head, and neck tissues. The vertebral arteries enter the C6 transverse processes bilaterally and run superiorly through all transverse foramen up to C1 (Gray, 1918). They then run along the C1 posterior arch and enter the skull. The stretch of the vertebral arteries could decrease the vessel diameter due to Poisson's ratio, limiting the blood flow to the tissues (Siegmund et al., 2009). Stretching or pinching of the vertebral arteries along a turn of its course can also lead to WAD. Cervical spine cadavers with muscle force replication have shown higher strains in the vertebral arteries during side and head-turned rear impacts (Carlson et al., 2007; Ivancic et al., 2006a). The peak strains of 30.5 and 17.4 mm in side impact and head-turned rear impacts significantly exceeded the physiological elongation limits of 7.1 mm. Evidence on chronic pain due to vertebral arteries is insufficient and the cadaver studies were performed with muscle force replication systems which may have provided non-physical muscle response. However, the studies do suggest that vertebral arteries could potentially be damaged due to side and head-turned rear impacts and therefore should be further studied in the future.

Muscles can be damaged during impact and may cause acute or chronic pain in the neck. The injury mechanism is caused by eccentric contraction which is imposed from lengthening of the muscle during active contraction (Siegmund et al., 2009). The threshold for muscle strains has been shown to be 5-20% (Macpherson et al., 1996; McCully & Faulkner, 1985). Simulated impacts by Vasavada et al. (2007) showed that during rear impact collisions, the average strain on the sternocleidomastoid, splenius capitis, and the semispinalis capitis exceed the thresholds. There is no supportive evidence to suggest that muscle injury can cause long term injuries, but interactivity with other anatomic sites may contribute to chronic pain. Some of the muscles are directly connected to the capsular ligaments and their early activation during rear-impact collisions may increase the strains in the capsular ligaments, leading to chronic pain (Siegmund et al., 2008). Neck muscle activation can change the kinematics of the cervical spine during impact and this may cause excessive strains on the tissues of the neck and may lead to chronic pain (Siegmund et al., 2009). Interaction between muscles and the nervous system may also cause chronic pain. Altered neuromuscular patterns have been observed in patients with chronic pain (Fella et al., 2004; Nederhand et al., 2002), however, these abnormalities may be a protective strategy to avoid pain rather than a physiological deficit in motor control (Siegmund et al., 2009). Future research is required in order to provide enough evidence that

musculature could contribute to chronic pain and therefore muscles strains will not be investigated in this thesis.

3.5 Neutral versus Out-of-Position Impact Scenarios

The study of whiplash injuries in the cervical spine has been mostly performed in situations where the neck is in a neutral position. In reality the neck is not always in its neutral position during impact. In many situations impact occurs when the driver is looking out to the side for traffic or pedestrians. The rotation of the head can introduce strains in some of the ligaments in the neck. As a result of impact, the stretched ligaments can undergo large deformation that can increase the severity of injury after impact.

Shugg et al. (2011) studied the posture of neck in driving. In this study the subjects drove in the city of Guelph, Ontario through specific routes. The routes included residential (Res), thruway (Thr), and highway (Hi) driving; minor driving tasks such as lane changes were also included. The results from the study are summarized in Figure 3-8. RHLC and LHLC refer to right hand and left hand lane changes, respectively. The results show that on average drivers spent a longer time in an out-of-position posture in tasks that were performed in residential areas with the exception of lane changes. The average peak angles of the neck were 42.5 (SD=18) and 35.7 (SD=14.2) degrees to the right and left, respectively. The study also noted that on average drivers routinely adapted to out-of-position head postures 13% of the time. The result of this study is helpful in identifying the range of motion and the limits of axial rotation of the head during driving.

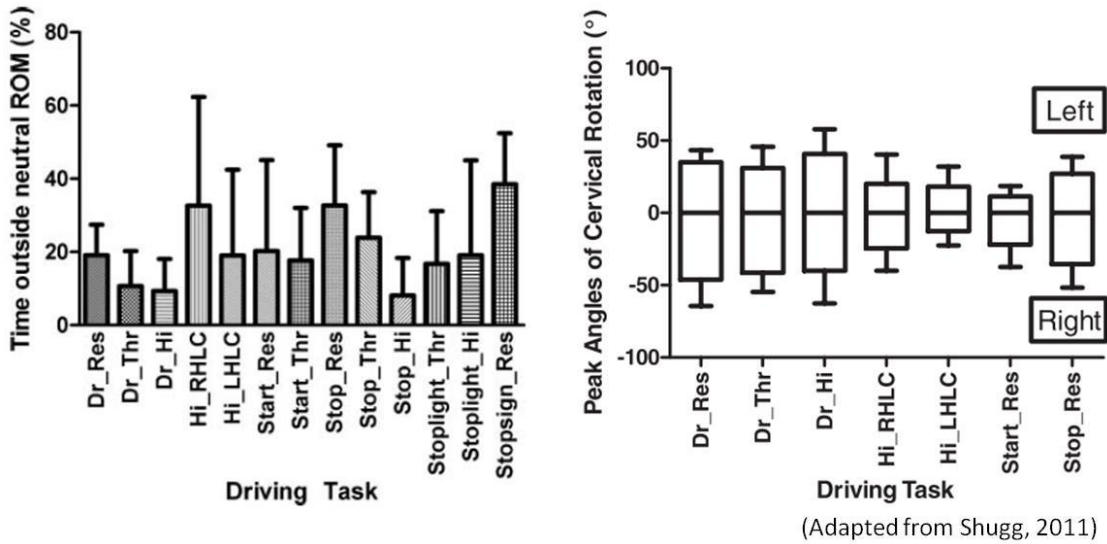


Figure 3-8: Mean percentage (Left) and mean peak angles (Right) of the time outside neutral range of motion and rotation from left to right in degrees

Sturzenegger et al. (1995) performed the first study that was influential in the relation between head rotation and whiplash-associated disorders. The study included 117 patients with a mean age of 30.8 years (SD=9.5) who did not have any history of pre-existing neurological dysfunction, symptoms, head trauma, fracture or dislocation in the cervical spine, and injuries to the other parts of the body. In this work, accident features including passenger position, head restraints, use of seatbelt, damage to seat, head position at the moment of impact, state of preparedness, car stationary when hit, and injury type (rear-end, frontal, or side impact) were studied. The initial examination was performed on average at 7.4 days after the accident and included neurological and radiological examinations. In this baseline examination, the whiplash-associated disorders were reported along with the features that affected the accident. Regular follow-up examinations were performed every three months until the patients were completely free of symptoms. The results showed that after one year, 24% of the patients were symptomatic. From the features of the accident mechanism, head position was the only one that was significantly associated with the persistence of symptoms (Table 3-3). The findings of this study are very valuable; however, even though the results show that there can be a higher persistence of injury due to head rotation, there is no scientific indication that the severity of injury is increased due to head rotation.

Table 3-3: Influence of features of accident mechanism on the long-term course

		Symptomatic at 1 year (n=28) n(%)	Asymptomatic at 1 year (n=89) n (%)	Statistical analysis	
				X ²	P
Passenger Position	Driver	22 (79)	69 (78)	0.0	1.0
	Front passenger	6 (21)	16 (18)	0.0	1.0
	Back passenger	1 (4)	3 (3)	0.0	1.0
Head restraints	Yes	26 (93)	81 (91)	0.0	1.0
Point of head contact	Occiput	16 (62)	59 (73)		
	Posterior neck	7 (27)	11 (14)	1.55	0.21
	Unknown	3 (12)	11 (14)		
Use of seat belt	Yes	24 (86)	80 (90)	0.369	0.54
Damage to seat ^a	Yes	5 (18)	22 (25)	0.244	0.62
Head position at the moment of impact ^b	Straight	8 (29)	56 (63)	7.87	0.005
	Rotated to side	13 (46)	21 (24)	4.33	0.037
	Inclined	7 (25)	7 (8)	4.48	0.034
	Unknown	0	17		
State of preparedness	Prepared ^c	7 (25)	24 (27)	0.0	1.0
Car Stationary when hit		17 (61)	42 (47)	1.06	0.302
Injury type	Rear-end only	14 (50)	30 (34)	1.76	0.183
	Rear-end + Frontal ^d	5 (18)	21 (24)	0.141	0.706
	Frontal only	7 (25)	25 (28)	0.005	0.938
	Side impact	2 (7)	13 (15)	0.122	0.726

a Seat upright broken, for example

b A combination of different head position is possible

c Foreseeing the collision, for example, in the rear-view mirror

d Patients car pushed into a car standing ahead

(Adapted from Sturzenegger, 1995)

Kaale et al. (2005) used magnetic resonance imaging (MRI) to study the soft tissue structure of the upper cervical spine. In this study, a selected group of ligaments in the upper cervical spine were compared between normal and whiplash-associated disorder patients. The examination was performed after 2-9 years of the accident. There were a total of 30 control persons and 92 whiplash patients from which 47 had their head/neck rotated during the impact. The ligament and membrane were classified into four possible categories of grade 0-3. With this classification, grade 0 reflected normal structure and grade 3 reflected severe damaged tissue. It must be noted that MRI has not yet been proven to visually identify ligament injury. The damage to the tissues was evaluated by

increased signal intensity or reduction in the length or cross sectional of the tissue. The result of the study showed that whiplash patients had more high-grade lesions than the control persons. Patients who had their head/neck rotated to the side during the impact more often had high-grade lesions of the transverse and alar ligaments. The results of this study can be helpful in pinpointing the ligaments that might be damaged due to non-neutral position impacts. The measurement of injury in this study was evaluated based on MRI signals and imaging processing and there was no real scientific method to prove that the classification of high-grade lesion is related to injured ligaments. The examinations were also performed a very long time after the accident. Therefore the damage to the ligaments might have been caused by the fatigue of the ligaments.

Kumar et al. (2004a, 2004b, 2005a, 2005b, 2005c & 2005d) conducted studies with similar setups to investigate the effect of impact direction, head rotation, and trunk position on the Electromyography (EMG) of response of the cervical spine muscles. In his studies twenty healthy volunteers were subjected to impact accelerations, ranging between 4.3-13.9 m/s². In the setups, head positions of volunteers were either 45° to the left or right and the trunks were flexed 45° to the left or right. The results of head kinematics and muscle EMG were reported. Kumar et al. (2004b) tested the effect of trunk position in rear impact and found that the EMG responses were greatly reduced and concluded that muscle injury seemed less likely in out-of-position. Kumar et al. (2004a) investigated the effect of head rotation in right posterolateral impacts. He concluded that there was an asymmetry in the paired sternocleidomastoid which could affect the risk of injury of this muscle. Kumar et al. (2005a) investigated the effect of head rotation in right lateral impact. He concluded that the rotation of the head reduced the muscle activity and therefore the risk of muscle injury. Kumar et al. (2005b) studied the effect of head rotation in left posterolateral impact. The result suggested that there is an increased EMG generation mainly in the contralateral sternocleidomastoid due to the direction of the impact. The head rotation reduced the EMG response of the cervical muscles and therefore the risk of muscle injury was reduced. Kumar et al. (2005d) investigated the effect of head position in rear impact. He concluded that the risk of injury tends to be great for the sternocleidomastoid muscle during out-of-position posture. Kumar et al. (2005c) investigated the effect of head and trunk positions in 8 impact directions. The directions of the impact that were investigated were anterior, posterior, and left and right anterolateral, lateral, and posterolateral. It was concluded that EMG and kinematic measures results did not show an increase in muscle response. It was noted that having the head rotated or trunk flexed, can reduce the apparent result of the perturbation.

The studies by Kumar provide some understanding to the level of muscle activity in the cervical spine. The big limitation to all the studies is that the tests were volunteer based and none of the studies tested accelerations that can result in any type of injury. Therefore, even though the results provide a good understanding on the muscle activities during low-speed impacts, there is no indication or proof that the same type of behaviour could be achieved in high severe impacts. The series of studies focus on muscle injury and therefore there is no knowledge on whether out-of-position postures may cause ligament injury. The studies reported the head acceleration during the impact but there was no information on the thoracic acceleration and therefore the information could not be used to validate Finite Element model kinematics during out-of-position impacts.

The following experiments were used by Panjabi et al. (2006b) and Ivancic et al. (2006b) to investigate the effect of head-turned rear impacts on cervical spine soft tissue injury threshold acceleration, correlate IV-NIC with multiplanar injury, and determine IV-NIC injury thresholds at the cervical spine intervertebral levels. Six human cervical spine specimens with an average age of 80.2 years were used. A surrogate head was rigidly attached to the occipital mount. A system of cables and spring were attached to the cervical spine to replicate passive muscles that influence the cervical spine. The head was rotated in flexion, axial rotation, and lateral bending. A three-plane flexibility test was performed to find the neutral zone and the range of motion of the neck in flexion, extension, lateral bending, and axial rotations. A horizontal impact of 2 g was used to dynamically precondition the specimen and impacts of 3.5, 5, 6.5, and 8 g were applied to the T1 for analysis. The three-plane flexibility test was performed after each impact scenario and soft tissue injury was defined as a statistically significant increase in the neutral zone or range of motion of any intervertebral level, above the baseline values obtained after the preconditioning simulation.

The results from Panjabi et al. (2006b) showed that injury threshold was at 5 g. At this acceleration, injuries happened in extension neutral zones or axial rotation range of motions between vertebral bodies of C3-T1, with the exception of C6-C7. In the impact levels of 8 g, 3-plane injury happened in C5-C6 and 2-plane injury occurred at C7-T1 in extension and axial rotation. Injuries also occurred in C0-C1 and C3-C5 in axial rotation and C6-C7 in extension. Panjabi concluded that head-turned rear impact significantly increases the chance of injury. Ivancic et al. (2006b) found that the range of IV-NIC injury thresholds were between 1.1 at C0-C1 and C3-C4 to 2.9 at C7-T1. Based on the correlation between IV-NIC and multi-planar injuries, it was concluded that three-plane intervertebral instability was caused primarily by dynamic extension during head-turned rear impact.

There are some limitations to the studies by Ivancic et al. (2006b) and Panjabi et al. (2006b). The muscle force replication system which was created with cables and springs may not be representative of passive muscles in the neck; therefore influencing the kinematics. Injury was defined as a significant increase in neutral zone or range of motion in the cervical spine. Even though increase in range of motion can be associated with injury, visual presence of injury in the soft tissue is a much stronger indication to identify injury in the neck. Finally, the average age of the specimens represented an elderly population and the weight of the surrogated head was lower than a 50th percentile male population. Because of the old age of the specimen, the soft tissue of the neck might have been degenerated or weaker than an average adult. The lower weight on the surrogate head also did not represent an average adult and therefore would influence the kinematics of the system.

Currently, there is limited information on out-of-position neck injuries due to impact. The studies that were mentioned in this section showed that when an occupant is in a non-neutral posture, impacts can cause more severe injuries to the neck. Scientists have used different procedures to identify the location of injuries but none have been able to successfully show injuries in the ligaments or soft tissues due to impact.

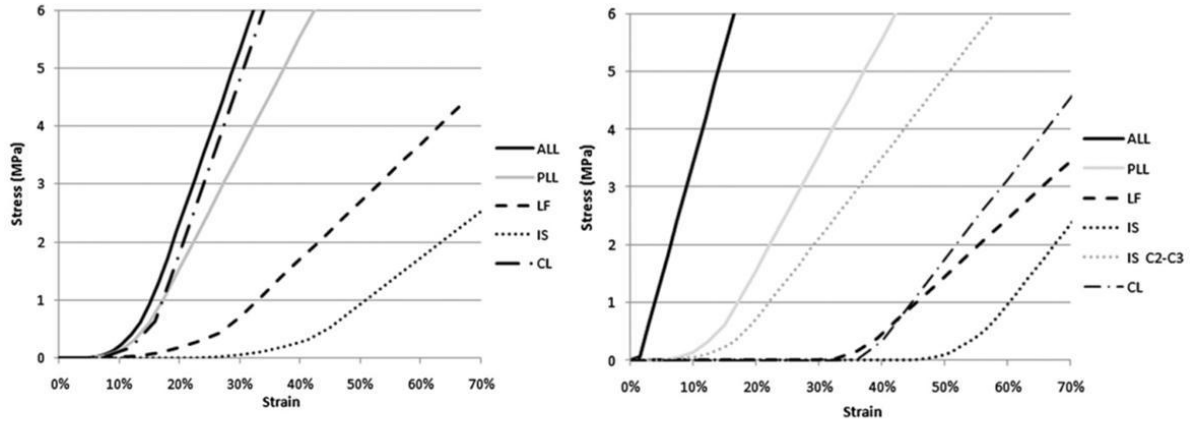
3.6 Out-of-Position Numerical Models

Finite Element models provide an alternative method to study the dynamic behaviour of the cervical spine. FEA could be used to study both the kinematics and the stress and strain on tissues resulting from the impact. The ability to change loading and boundary conditions to understand the effect of different parameters on the cervical spine makes FEA a very effective method in studying injuries due to impact. Yoganandan et al. (1996a) specified that simulation models should have representative geometry, material properties, and loading and boundary conditions. In order to be able to use FE models to simulate various impact situations, the model must also be validated against appropriate experimental data. This section provides the studies that have been found which are related to out-of-position FE models.

Brolin and Halldin (2004) developed a FE model of the upper cervical spine and used the literature for validation. Later, the effect of ligament properties on spinal kinematics was analyzed. The ligaments were modeled with nonlinear spring elements and the model was validated for axial rotation, flexion, extension, lateral bending, and tension. In order to have

similar results as compared to the literature, the stiffness and the toe region of some of the ligaments were altered. The stiffness of the PAOM and CL-01 ligaments were increased and their toe regions were reduced. The stiffness of the CL-12 ligament was decreased and the toe region and the deformation to failure were increased. This allowed for most of the simulations to be within the standard deviation of the validation studies. Using a parameter study, they showed that changes in material property of any ligament would have an effect on the motion of the upper cervical spine. It was concluded that capsular ligaments are the most sensitive factor that can affect motion in the upper cervical spine.

Kallemeyn et al. (2010) worked on the validation of C2-C7 cervical spine FE model. In this study, Kallemeyn experimentally applied moments of up to 1 Nm/s to a C2-C7 cervical spine (age: 74 years) in flexion, extension, lateral bending, and axial rotation. The motion of the intervertebral bodies were noted and used for the validation of the FE model. The FE model was initially assigned baseline material properties from the literature. In order to fit the results of the FE model to the behaviour of the experimental data, the material properties were calibrated, while keeping the properties within the range of properties which were used in literature. Figure 3-9 shows the baseline and the calibrated ligament properties that were used in the FE model. To obtain the experimental data from the FE model simulations, both the laxities and the stiffness of the ligaments were altered. Bone properties and annulus ground properties of the spine were also changed in order to obtain the experimental results. Following calibration, the model provided a good estimate of the experimental results in flexion, extension, lateral bending, and axial rotation. This study shows the challenges in creating a representative cervical spine model that can mimic the proper behaviour of the human neck. The advantage was that the experiment and the Finite Element model were created by the same group of individuals and therefore the authors had a solid understanding of the boundary condition required for the FEA. The limitations to the study were that only one cervical spine was used to obtain experimental data and the results were not compared to past studies.



(Adapted from Kallemeyn, 2010)

Figure 3-9: Kallemeyn's ligament properties for the baseline (left) and calibrated (right) models

Zhang et al. (2006) developed a nonlinear FE model of the cervical spine. To validate, a moment of 1.0 Nm was applied incrementally to the skull in flexion, axial rotation, and lateral bending and the motion of each vertebral body was recorded. Most ligaments were modeled with nonlinear elements and the material properties of the head and spine components were assumed as linear elastic material. The results were compared to the literature and the rotation values were mostly within the standard deviations in flexion and axial rotation. The rotations in lateral bending were too low between C2-C7 vertebrae when compared to the literature. The author concluded that the FE model offered potential for biomedical and injury studies. This was a good study but the author did not provide the nonlinear material properties of the ligaments and therefore the reader has limited information on the toe region of the ligaments and no information on whether there were added laxities. The assumption that other parts of the neck are elastic could reduce the accuracy of the results in impact scenarios because most tissue behaviours in the neck are viscoelastic.

Storvik and Stemper (2010) studied the effect of axial head rotation on the facet joint capsule strains during rear impact. The FE model of the cervical spine was previously validated in neutral position and not in axially rotated conditions due to a lack of experimental data in literature. Hill-type muscle elements were used to represent musculature. Axial rotations between 0 and 60° and impact severities between 8 and 24 km/h were studied. The results of the study showed an increase in ligament strains from C3 through C7 due to an increase in axial rotation. The capsular ligament strains between C0-C2 and C7-T1 were not computed. In this study, the severity of impact did not have a significant effect on the capsular ligament strains. The study concluded that axial head rotation can increase the risk of

facet joint injury. There are a few limitations with this study that should to be noted. The model was not previously validated in range of motion or failure. The author used two arguments to validate the model against literature. The first one was that his study showed that approximately 86-94% of the total cervical spine axial rotation was between C1-C2. This value is very high as compared to experimental data of cadaver studies with the average of 42.5% (Ivancic et al., 2006b). The second point of validation was that the C0-C1 level demonstrated rotation in an opposite direction to the applied rotation at C1-C2. Even though this behaviour was observed by Iai et al. (1993), others have not made similar observations (Dvorak et al., 1987; Penning et al., 1987; Ishi et al., 2004). Another deficit to the study is excluding the strain on C0-C2 and C7-T1 capsular ligaments.

The studies in this section show that there are many limitations to the current models of the cervical spine for out-of-position impacts. The studies proved the importance of material properties and proper validation of the models. The next two chapters provide an understanding of a recent neck model (Fice, 2010) and the enhancements to improve its performance for out-of-position rear impact scenarios.

Chapter 4

Model Description

This section provides a basic understanding of how the model was developed. Detailed information can be found in Panzer (2006). The model was designed to represent a 50th percentile male. The model was developed at University of Waterloo using Hypermesh for meshing and LS-Dyna for dynamic analysis. The model has been previously validated in various loading conditions. The segment level of the model has been previously validated in tension, flexion, extension, lateral bending, and translation (Panzer, 2006; Panzer & Cronin, 2009; DeWit & Cronin, 2012). In rear impact, the full cervical spine model showed good agreement with the head kinematic response and ligament strains (Fice et al., 2009; Fice et al., 2011). In tension, and frontal impact, the model has been validated using ligament strains and head kinematic response from volunteers (Panzer et al., 2011). Dorsal root ganglia and vertebral artery are not included in the model and will not be discussed in this thesis.

4.1 Vertebrae

The geometry of the vertebrae was taken from a model by Deng et al. (1999). This model was constructed from a commercial database of 3D surfaces (Viewpoint DataLab, Orem, UT). The dimensions of the geometries were within the standard deviations of 50th percentile males which have been measured by Gilad and Nissan (1986).

The skull and the first thoracic vertebra (T1) were modeled for inertial effects and boundary condition applications. Since injury was not being evaluated on the skull and T1, these geometries were modeled as rigid materials with shell elements and coarse meshes (Figure 4-1). This modeling technique allowed for the simulations to run in a time efficient manner. The skull mass and inertial properties were based on research from Walker et al. (1973) and are outlined in Table 4-1. The out-of-plane inertia ratios that have been used in the model were based on values that have been reported by Robbins (1983).

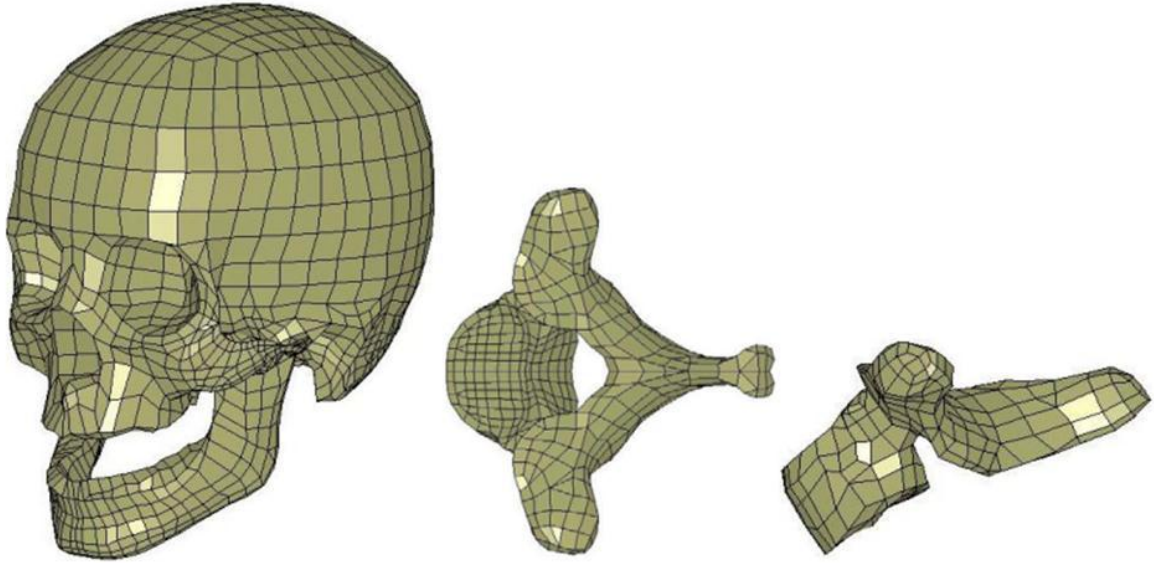


Figure 4-1: Skull and T1 vertebra

Table 4-1: Skull mass and inertial information

	Mass	Moment of Inertia		
		I_{xx}	I_{yy}	I_{zz}
Skull	4.376 kg	21060 kg mm ²	23300 kg mm ²	15200 kg mm ²

(Adapted from Panzer, 2006)

The cervical vertebrae (C1-C7) were modeled with fine mesh and rigid material properties (Figure 4-2). The vertebrae are capable of being switched into deformable bodies in the future to predict bone failure during impact; however, this feature has not been validated in the full cervical spine model. The cancellous bone of the vertebrae was modeled with solid elements (Figure 4-3). The relatively thin properties of the cortical bone and the bony plates allowed the use of computationally efficient shell elements. All dimensions of the vertebrae in the cervical spine are available in Panzer (2006).

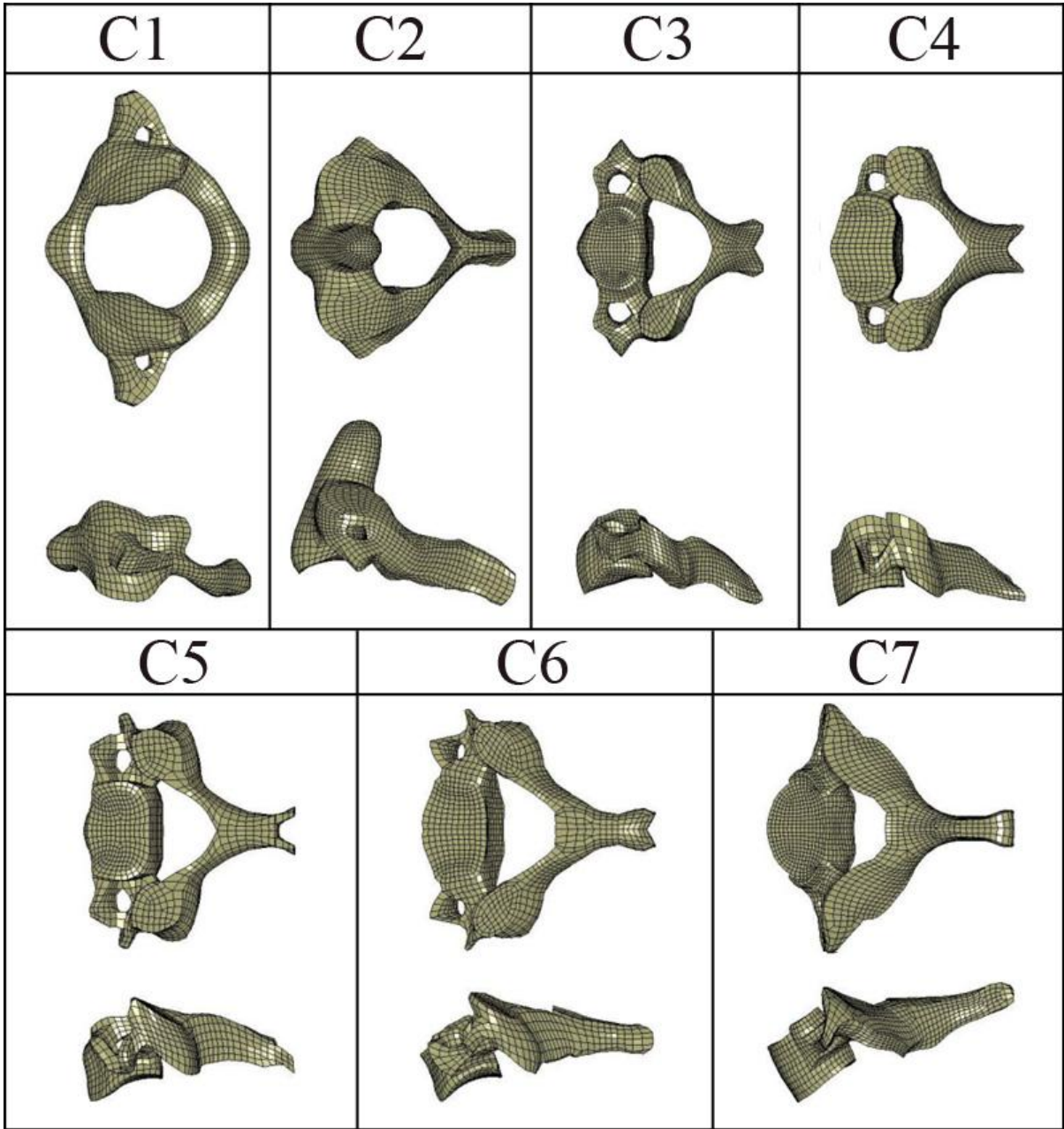


Figure 4-2: Cervical vertebrae geometries and mesh

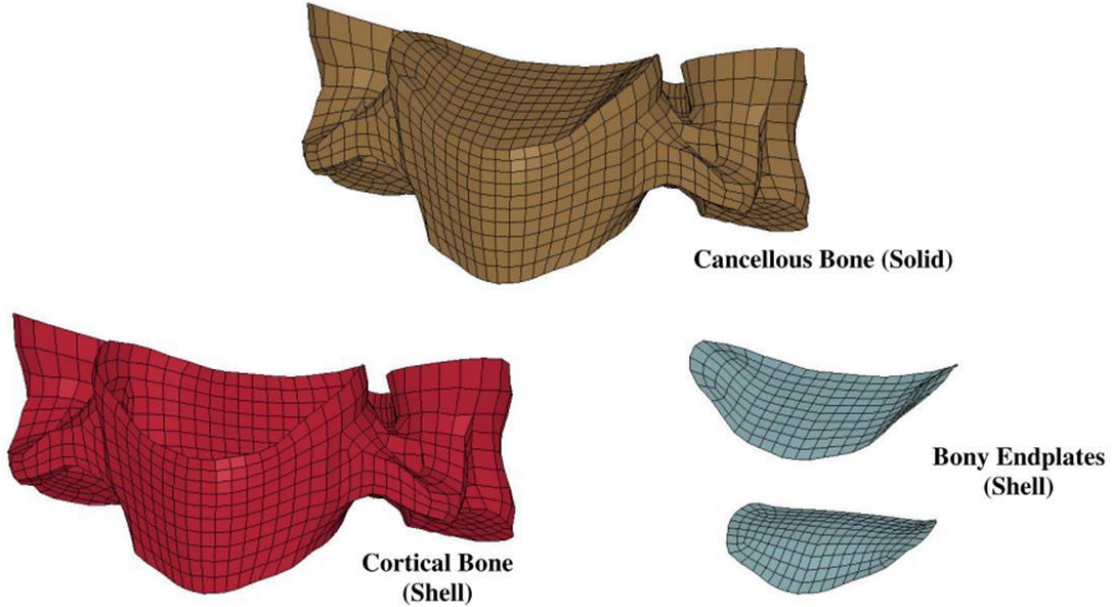


Figure 4-3: Cervical vertebrae components

4.2 Intervertebral Discs

The intervertebral discs were positioned between adjacent vertebrae. The geometry of the discs was dependent on the positions of the vertebrae. The heights between the vertebrae were based on 50th percentile male from data by Gilad and Nissan (1986). The transverse cross-sectional areas of the discs were between 200-400 mm² based on the data provided by Pooni et al (1986). The ratio of 1:2 was used between nucleus pulposus area and the total disc area based on data from Pooni et al (1986) and Iatridis et al (1996). The vertebrae and the intervertebral discs in cervical spine model are shown in Figure 4-4. The initial posture of the spine model was based on a 50th percentile male and the angles in the model agreed with anthropometric data reported by Klinich et al (2004).

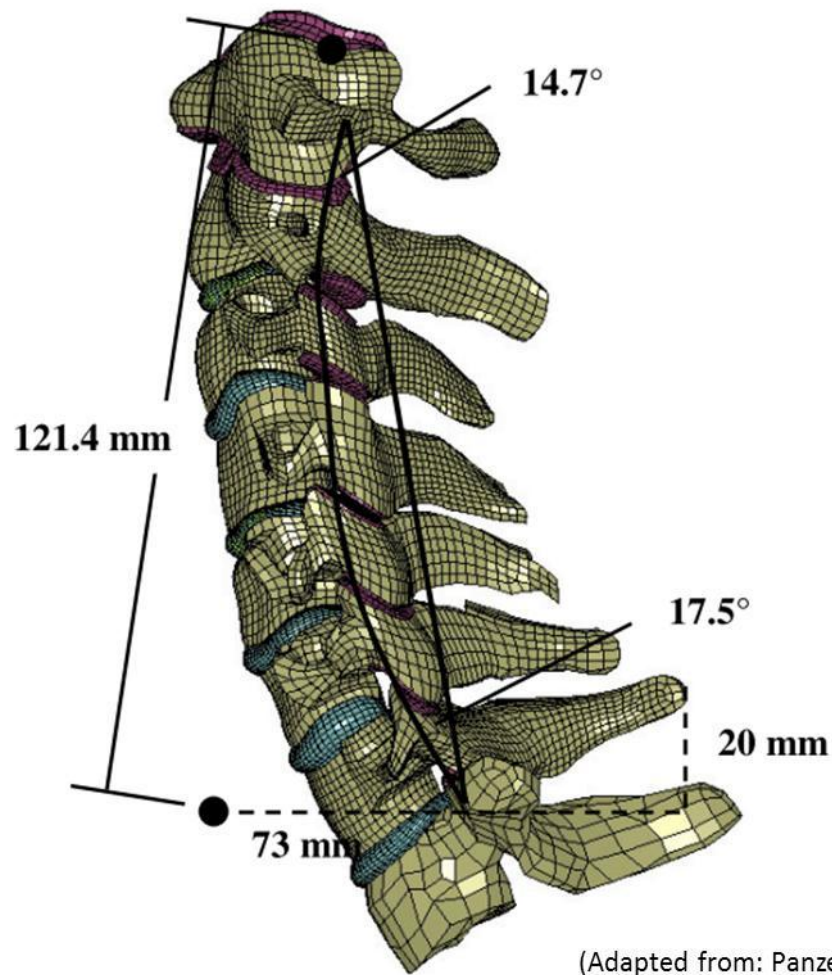


Figure 4-4: Length and initial posture of the cervical spine model

Figure 4-5 shows the intervertebral disc components. Solid elements were used to model the nucleus pulposus and the soft ground substance. The annulus fibrosus was modeled with five layers which were meshed using shell elements. The transmissions of load between the intervertebral discs happen through shared nodes. Tied contact interfaces were used between the discs and the vertebrae due to dissimilar mesh sizes. These contacts could be used in the future to allow for high stress failures.

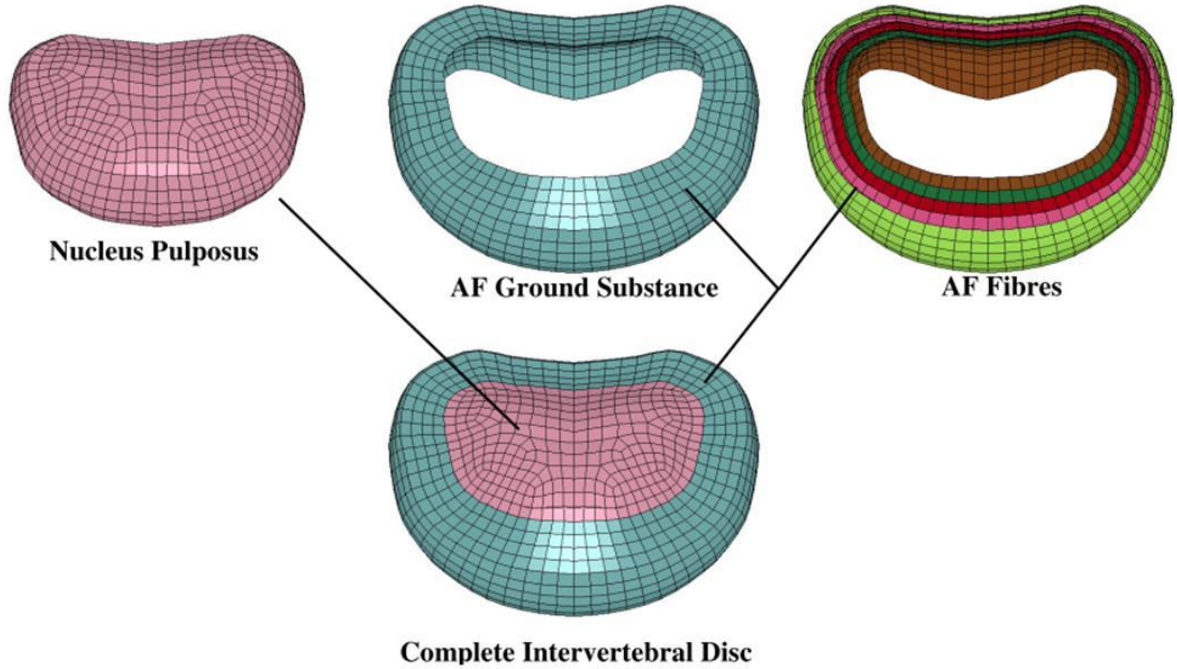


Figure 4-5: Intervertebral disc components

General linear viscoelastic material properties were used to model the fluid characteristics of the nucleus pulposus. The properties were based on a study on nucleus pulposus in relaxation by Iatridis et al. (1996). The parameters that were used to fit the viscoelastic model to the data are shown in Table 4-2 (Panzer, 2006).

Table 4-2: Nucleus pulposus material parameters

Material Model	Material Parameters	
	N = 4	K = 1.720 Gpa
Linear Viscoelastic	$G_1 = 0.5930$ kPa	$\beta_1 = 0.001477$ 1/s
	$G_2 = 0.6763$ kPa	$\beta_1 = 0.061524$ 1/s
	$G_3 = 0.9516$ kPa	$\beta_1 = 1.017893$ 1/s
	$G_4 = 2.0384$ kPa	$\beta_1 = 13.20041$ 1/s

(Adapted from Panzer 2006)

The mechanical properties of the annulus fibrosus are anisotropic and non-linear. Five layers of annulus were modeled in each disc. The angle of annulus varied radially from $\pm 45^\circ$ in the inner layers to $\pm 25^\circ$ on the outer layers (Cassidy et al., 1989; Marchand & Ahmed, 1990; White & Panjabi, 1990;

Wanger & Lotz, 2004). The material properties were based on cadaver studies (Holzapfel et al., 2005). In the study, single lamina samples were tested along the fibre directions and the average curves for the inner and outer lamina were reported for up to 4% strain. The method of continuity of slope was used by Panzer (2006) to extrapolate the data for higher strains (Figure 4-6).

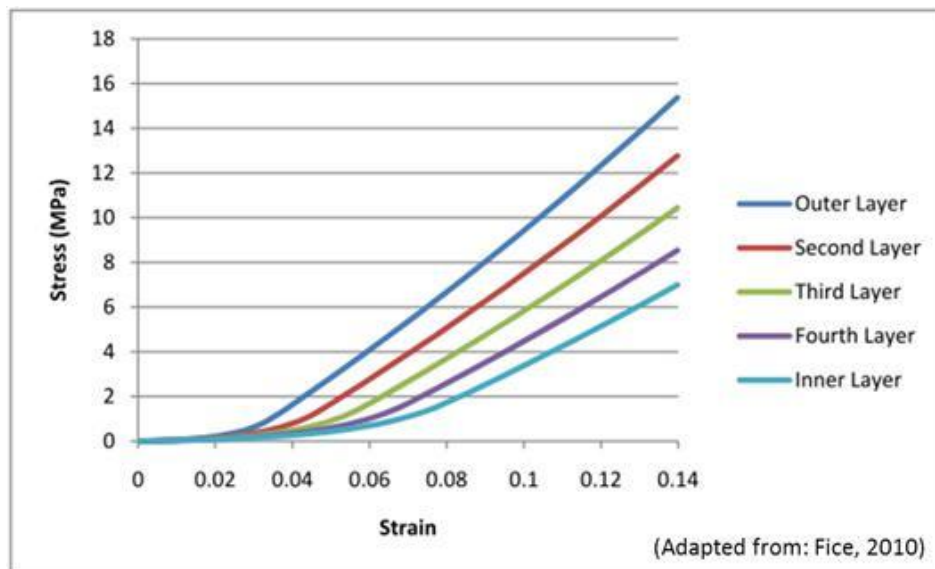


Figure 4-6: Stress versus strain along fibre directions of annulus fibres

The annulus fibrosus ground substance material properties could be obtained through experiments where the mechanical testing has been performed in the direction perpendicular to the fibre layers in order to minimize the influence from the fibres (Fujita et al., 1997; Elliott & Setton, 2001). The material model of Ogden-rubber was selected for the annulus fibrosus ground substance. The constants for the model were calculated based on the reported data from confined compression (Iatridis et al., 1998), unconfined compression (Wagner & Lotz, 2004), and uniaxial tension (Fujita et al., 1997). The method of least squares was used to fit the experimental data to the material model (Figure 4-7; Panzer, 2006). It has been reported that annulus fibrosus ground substance is viscoelastic at low strain rates (<10/s) (Iatridis et al., 1998; Iatridis et al., 1999; Holzapfel et al., 2005). There is limited information on the viscoelastic properties of the annulus fibrosus ground substance at high strain rates. The material model did not include viscoelasticity.

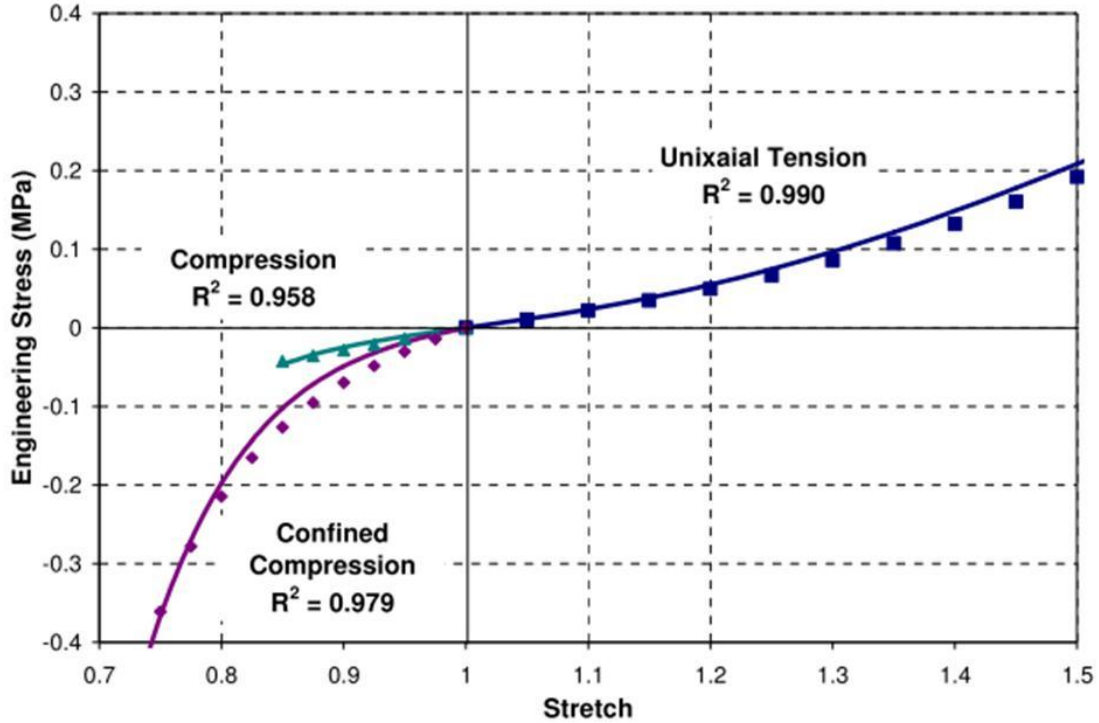


Figure 4-7: Annulus fibrosus ground substance model fit (Panzer, 2006)

4.3 Facet Joints

The dimensions of the facet joints were in agreement with literature (Francis, 1995; Panjabi et al. 1993; Pal et al., 2001; Yagamandan et al., 2003). The orientation of the facet joints have been measured and found to be in reasonable agreement with Pal et al. (2001) and Panjabi et al. (1993).

The facet joints between C2 to T1 were modeled using solid elements for the articular cartilage and simple pressure airbags for the synovial fluid (Figure 4-8). Simple pressure airbags were selected because they are more computationally efficient than fluid elements and provide the necessary material response. The airbags apply a hydrostatic pressure within defined pressure volume segments as the articular cartilages get closer to each other. Capsular ligaments connect the articular cartilages on the outside surface.

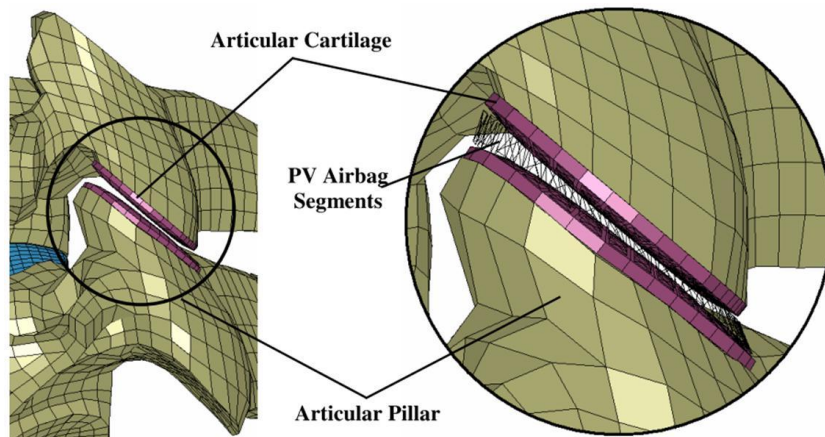


Figure 4-8: Facet joints

The cartilage in the upper cervical spine is illustrated in Figure 4-9. The contacting surfaces in the cartilages of upper cervical spine have more complex geometries compared to the middle and lower cervical regions. Due to this complexity, simple pressure airbags were not implemented for this region. The zero coefficient of friction between these surfaces should be sufficient to represent the synovial fluid.

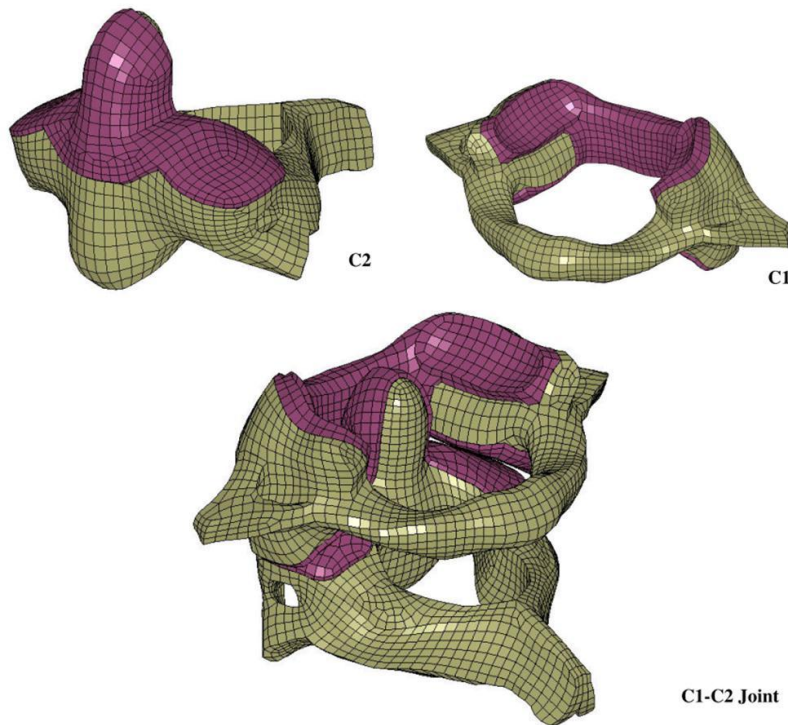


Figure 4-9: Cartilage in the upper cervical spine (Panzer, 2006)

The cartilage is a porous material with nonlinear mechanical properties and strong dependency on internal fluid-flow at low strain-rates. Fluid loss is not expected in strain rates seen in automotive crashes and quasi-linear viscoelastic material can be used to model the cartilage behaviour. This model is incompressible and accounts for strain rate effects including creep and relaxation. The constants for the material were found by curve fitting reported data by DiSilvestro and Suh (2001) (Table 4-3; Panzer, 2006). This model is limited since the strain rates from the study (0.001/s) are well below automotive impacts which are 100/s or higher.

Table 4-3: Material properties for Atricular Cartilage (Panzer, 2006)

Material Parameters	
N = 4	K = 2.0 Gpa
$G_1 = 0.2100$ kPa	$\beta_1 = 0$ 1/s
$G_2 = 0.0243$ kPa	$\beta_1 = 0.000303$ 1/s
$G_3 = 1.0824$ kPa	$\beta_1 = 0.080807$ 1/s
$G_4 = 1.9984$ kPa	$\beta_1 = 0.012927$ 1/s

(Adapted from Panzer, 2006)

The airbag material model uses pressure versus relative volume properties. These properties were calculated based on a simple squeeze-film Finite Element model (Kumaresan et al., 1998; Panzer, 2006). This model used rigid elliptical plates (1 mm apart), synovial fluid which was modeled with a bulk modulus of 2.2 GPa (equivalent to water), and enclosing synovial shell membrane with a Poisson's ratio of 0.4 and Young's modulus of 10 MPa (kumaresan et al., 1998) (Figure 2-1; Panzer, 2006). The force that was used to displace the fluid was converted to pressure using the surface area of the endplates and the results were used as the material properties for the airbags (Figure 4-11; Panzer, 2006).

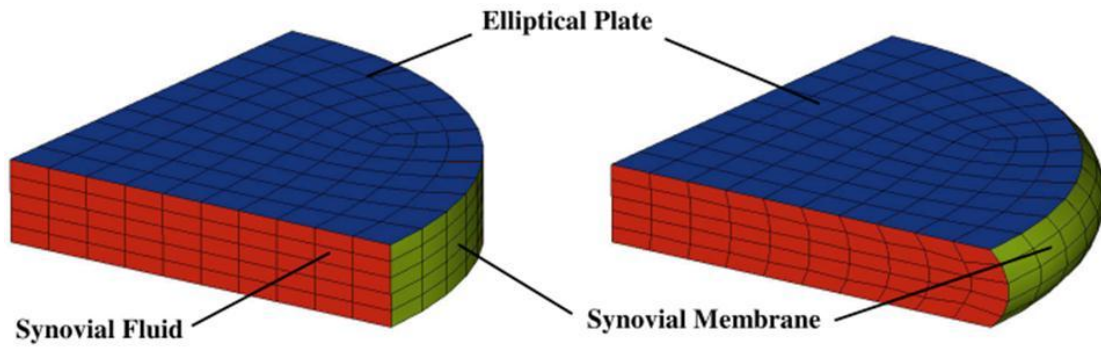


Figure 4-10: Simple squeeze-film model

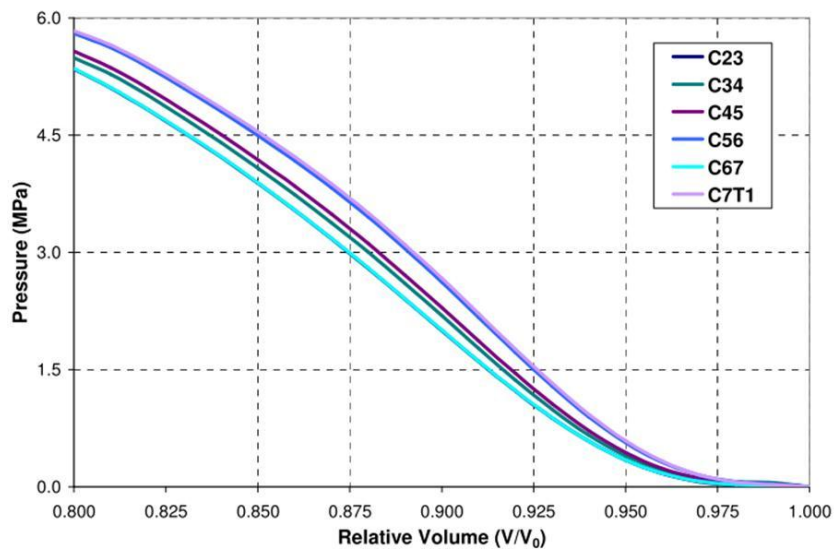


Figure 4-11: Pressure versus relative volume properties for synovial fluid (Panzer, 2006)

4.4 Ligaments

Ligaments have non-linear, viscoelastic, and orthotropic material properties. The literature reports ligament properties in force-deflection because measuring the cross-sectional area and the length of the ligaments is difficult (Przybylski et al., 1998). Ligaments were modeled using 1D discrete elements which allowed input of force versus displacement data. Discrete elements have previously been used in cervical spine ligaments (Yoganandan et al., 1996b; Clausen et al., 1997; Halldin et al., 2000; Ng & Teo, 2001; Meyer et al., 2004).

Each ligament was composed of a set of evenly spaced discrete elements. Each element was attached to the appropriate bone by sharing common nodes. The samples of the ligaments in the upper, middle, and lower cervical spine are shown in Figure 4-12 and Figure 4-13

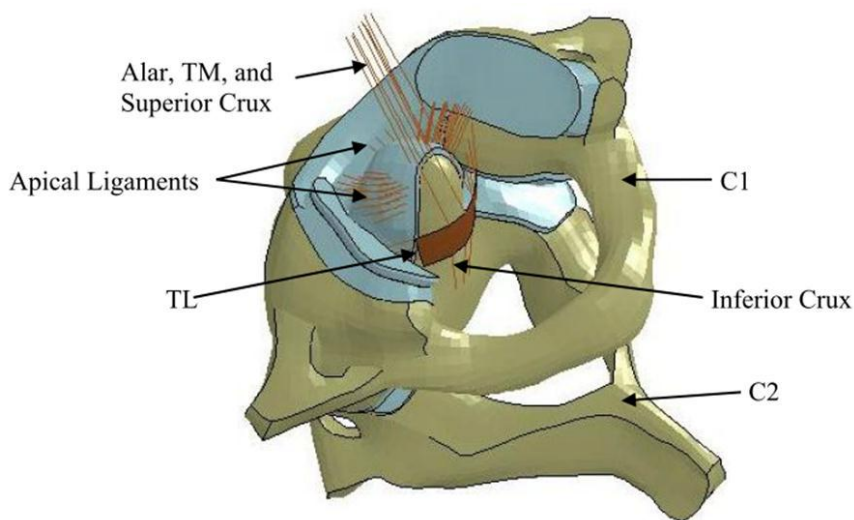


Figure 4-12: Sample ligament implementations in the upper cervical spine

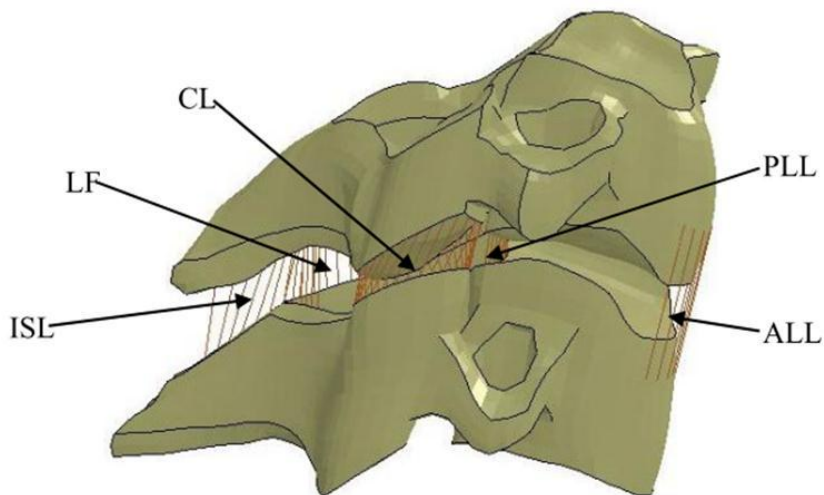


Figure 4-13: Sample ligament implementations in the middle and lower cervical spine

The ligaments were positioned using origin and insertion data which have been reported in various studies and anatomy textbooks. Since the literature reported data in force versus deflection, the proper length and cross-sectional dimensions were not necessary for the model implementation

(Panjabi et al., 1991a & 1991b; Przybylski et al., 1998; Yoganandan et al., 2000). The lengths of all ligaments excluding capsular ligaments were in agreement with the literature. The capsular ligaments were modeled shorter than the reported lengths for ease of attachment.

The force which is generated from the discrete elements has a force-deflection curve (nonlinear) and a force-velocity curve which allows rate dependent properties to be modeled. The normalized force-deflection curves for the cervical spine ligaments were based on data from Chazal et al. (1985). This data was used to find the force and deflection for end of toe and linear regions of the curve. The failure force and failure deflection data were taken from various studies of cadaver ligament response under tension (Yoganandan et al., 2001; Myklebust et al., 1988; Dvorak & Panjabi, 1987; Panjabi et al., 1998b; Chazal et al., 1985). The material properties for the hyothroid ligament, which connect hyoid bone and the thyroid cartilage, were calculated based on data from Vilkmann and Karmma (1989).

Heuer et al. (2007) had shown that ligaments in the spine were under preload in-vivo. Fice (2010) used a free body diagram shown in Figure 4-14 to mimic Heuer's test to solve for the unknown forces using compatibility equations. The resulting preloads are summarized in Table 4-4. The calculated preload displacement from the LF agreed with the reported values in the literature (Nachemson & Evans, 1968; Viejo-Fuertes et al., 1998).

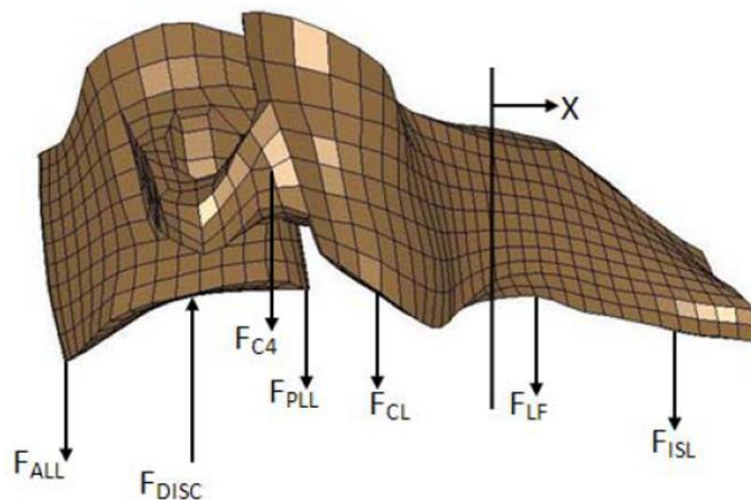


Figure 4-14: Free-body diagram to extract preload data from a study by Heuer et al. (2007) (Fice, 2010)

Table 4-4: Model’s ligament preloads of the middle and lower cervical spine

		C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
ALL	Force (N)	20.82	18.76	19.00	19.67	18.35	18.13
	Displacement (mm)	1.72	1.63	1.64	1.50	1.45	1.45
PLL	Force (N)	5.62	6.46	6.46	5.69	7.22	7.40
	Displacement (mm)	0.81	0.84	0.85	1.12	1.19	1.19
LF	Force (N)	2.61	3.12	2.94	3.41	3.10	3.52
	Displacement (mm)	1.04	1.08	1.07	1.45	1.42	1.45
ISL	Force (N)	0.53	0.69	0.62	0.62	0.55	0.56
	Displacement (mm)	1.13	1.19	1.17	1.25	1.22	1.23
CL	Force (N)	1.77	2.41	2.21	2.02	2.17	1.81
	Displacement (mm)	1.46	1.59	1.59	1.06	1.07	1.04

(Adapted from Fice, 2010)

The resulting quasi-static force-displacement responses with the addition of the preloads for the ligaments of the cervical spine are shown in Figure 4-15 to Figure 4-18 (Fice, 2010). The dynamic scale factor for the model was curve fitted to experimental data by Yoganandan et al. (1989). In this experiment, Yoganandan tested the ALL and LF ligaments from cadavers at rates of 9, 25, 250, and 2500 mm/s. The implementation allowed the ligaments to exhibit rate effects in high strain rate vehicle crash simulations.

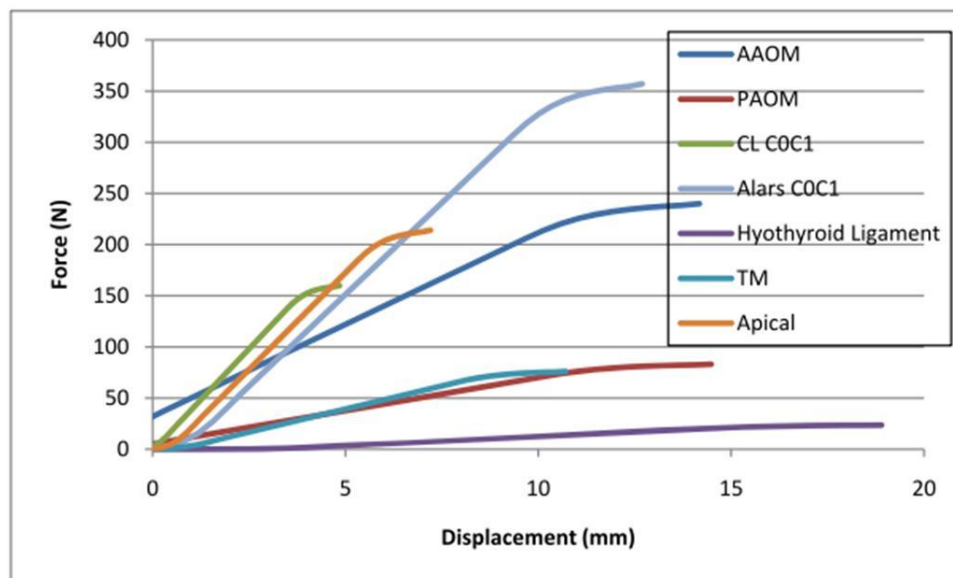


Figure 4-15: Force-deflection curves for upper cervical spine ligaments (Part 1) (Fice, 2010)

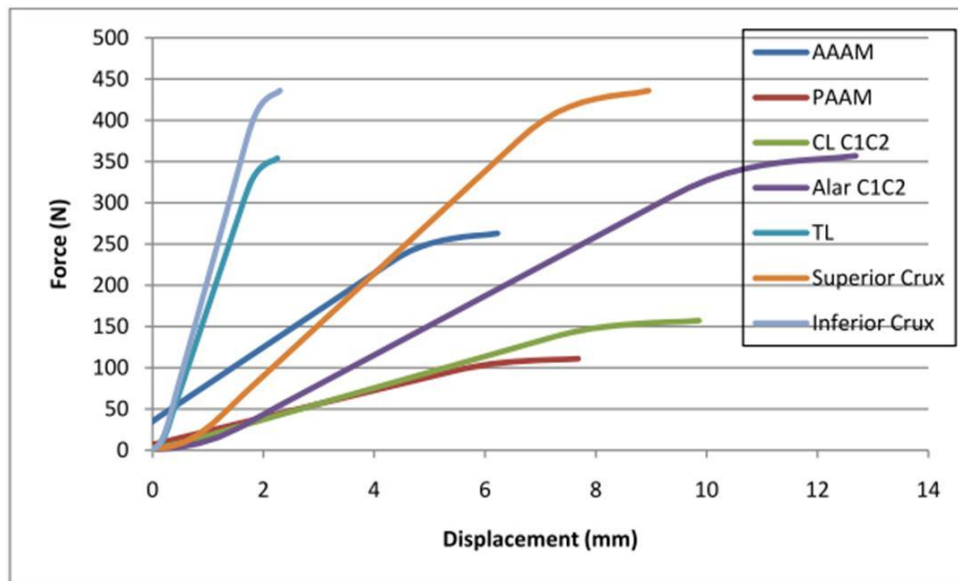


Figure 4-16: Force-deflection curves for upper cervical spine ligaments (Part 2) (Fice, 2010)

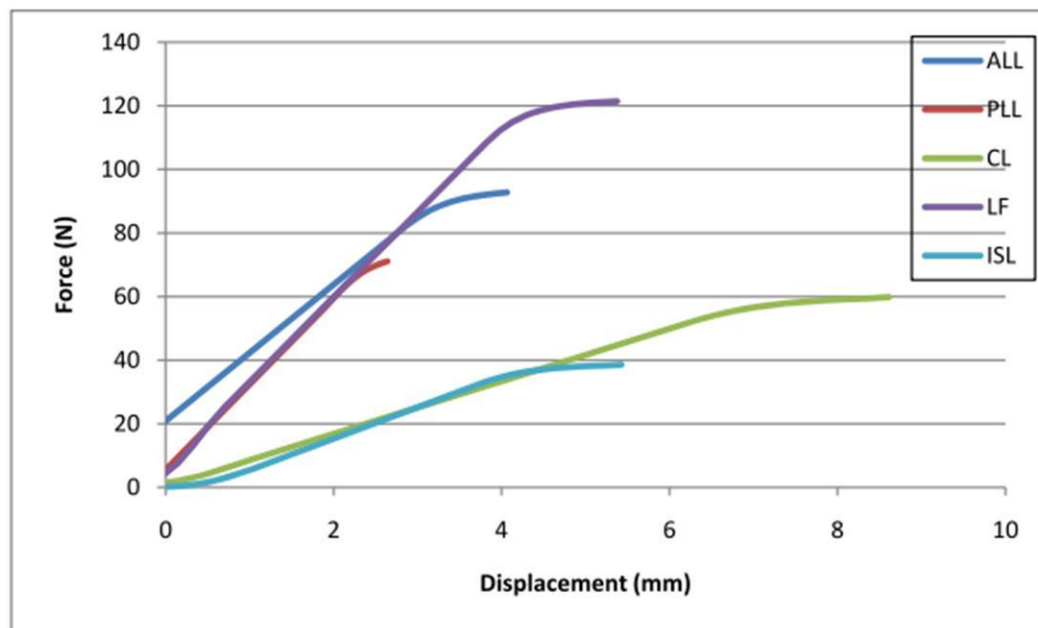


Figure 4-17: Force-deflection curves for middle cervical spine ligaments (Fice, 2010)

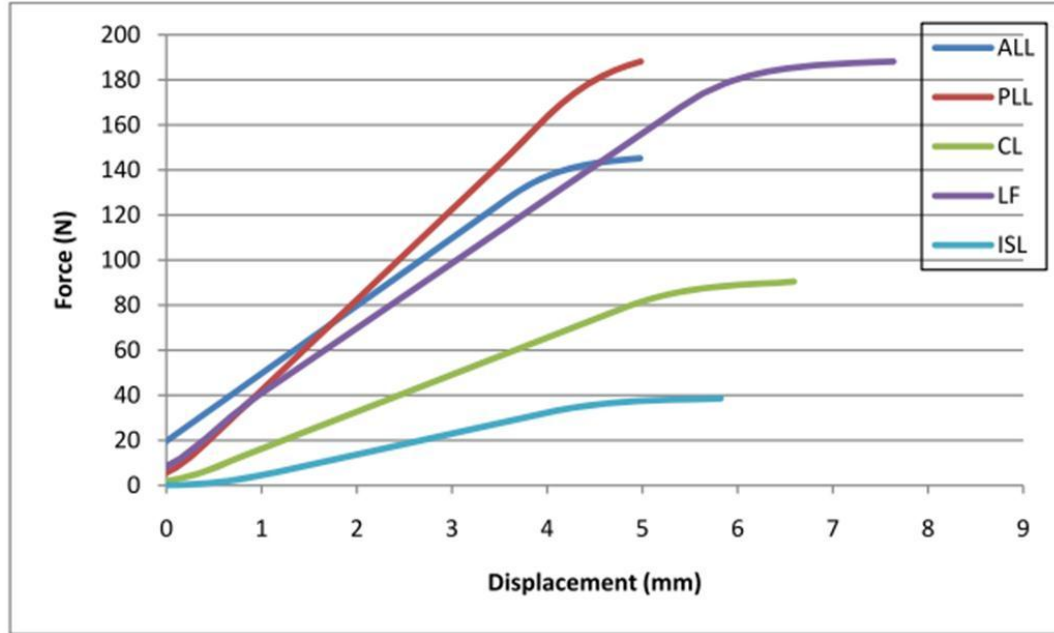


Figure 4-18: Force-deflection curves for lower cervical spine ligaments (Fice, 2010)

4.5 Muscle

The muscles in the cervical spine were modeled using 1D discrete elements. The origin and insertion points were determined based on anatomy textbooks (Gray, 1918; Agur & Dalley, 2004). Each muscle was divided into segments in order to model the origin and insertion points. This resulted in 90 separate muscle pairs which represented the 27 different muscles in the cervical spine. It must be noted that some of the muscles in the neck which primarily help with swallowing were not modeled because they have very small effect on the behaviour of the neck for the impacts under study. Each muscle segments was divided into a series of equal length elements. To model the proper curvature of the muscles, the intermediate nodes in each segment were constrained to a vertebra. This allowed a physical movement of the muscles as shown in Figure 4-19 (Fice, 2010). The disadvantage is that the elements within the segments may have different loads. This becomes important when the relative rotation between two adjacent vertebrae is significantly different. Since 1D discrete elements do not have mass properties, the mass of the muscles were added to the end nodes of the muscle elements. The mass properties were calculated through neck muscle volumes and densities in the literature (Knaub & Mayers, 1998; Ward & Lieber, 2005).

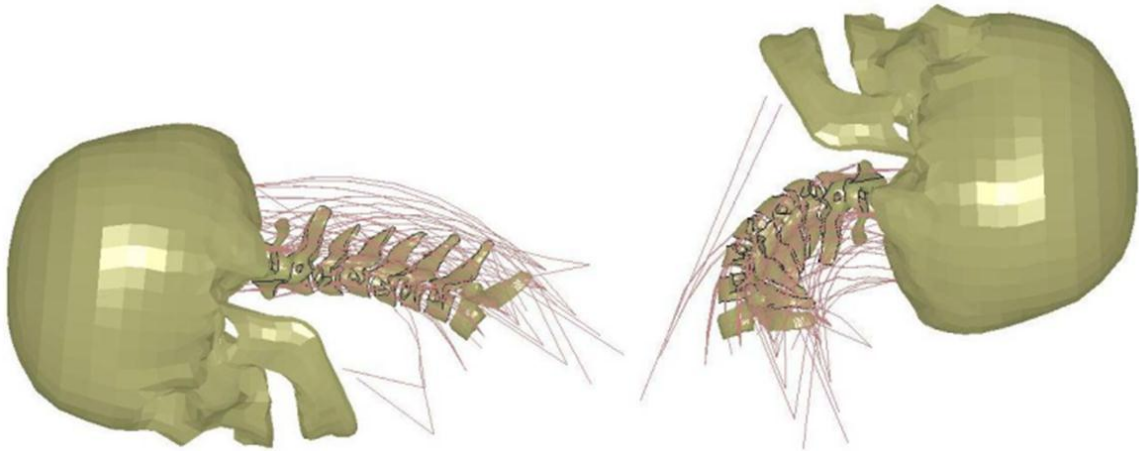
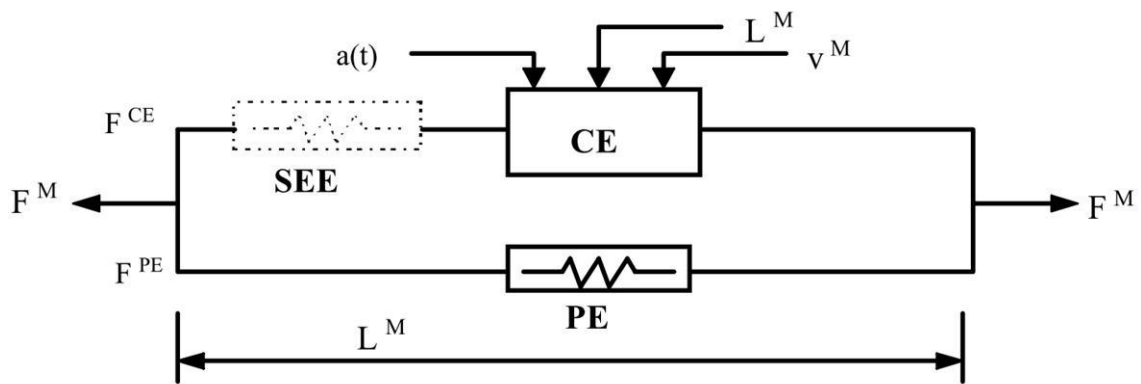


Figure 4-19: Neck muscles in the cervical spine model in flexion (left) and extension (right)
(Fice, 2010)

The muscles in the cervical spine were modeled using a Hill-type muscle model. This model allows for both passive and active muscle responses to be modeled. The model consists of parallel elastic elements (PE), contractile elements (CE), and series of elastic elements (SEE) (Figure 4-20). The parallel elements simulate the passive muscle behaviour based on the current length of muscle. The contractile element simulates the active muscle behaviour based on muscle length, velocity, and activation state. The series of elastic elements are used for tendon-compliance but not always included.



(Adapted from: Hallquist, 2003)

Figure 4-20: Hill Muscle Model Schematic

Muscles do not carry compressive force and therefore at negative strains the force caused by the passive elements was set to zero. The data that had been used for passive muscle behavior were taken

from research by Winters (1995). The strain value of 0.6 at which the muscles generate a force of equal magnitude to the isometric force was also taken from the same study. The behavior of the passive muscles is shown in Figure 4-21 (Panzer, 2006).

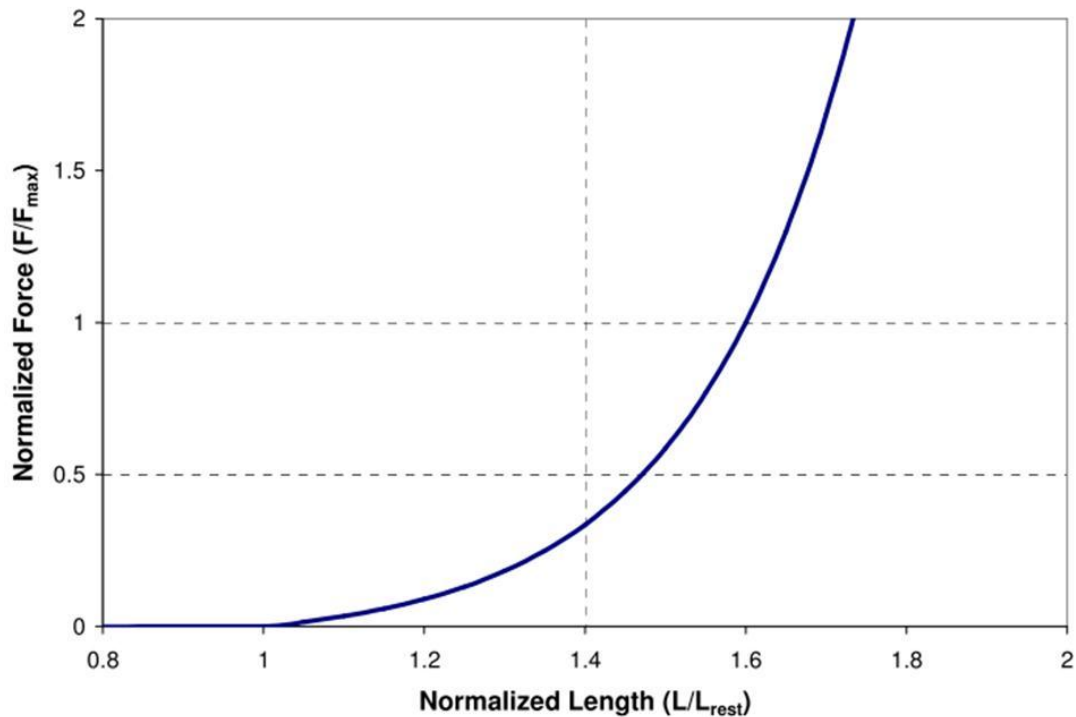


Figure 4-21: Parallel element response for Hill muscle model (Panzer, 2006)

The contractile force is the product of the peak isometric force and scale factors from muscle length, contractile velocity, and activation state ($a(t)$). The peak isometric force was calculated by the multiplication of the physiological cross-sectional area of the muscle by the isometric stress. The data used to calculate the peak isometric forces was taken from studies by Knaub & Myers (1998) and Winters & Stark (1988). Figure 4-22 shows the relationship of active force to the length of the muscle. The force is at its highest when the length is 1.05 times its original length and it is decreased if the ratio is altered (Winter & Woo, 1990). The muscle is capable of producing a force which is higher than the isometric force when it is contracting and it produces a lower force during extension (Winter & Woo, 1990; Fung 1993; Figure 4-23). There is a delay between the time that the brain triggers the muscles and the time that the muscle contraction occurs (Winters & Stark, 1988).

Figure 4-24 shows an example of muscle activation. In this Figure, $a(t)$ represents the scaling factor which enables muscle activation and the red line represents an ideal neural input.

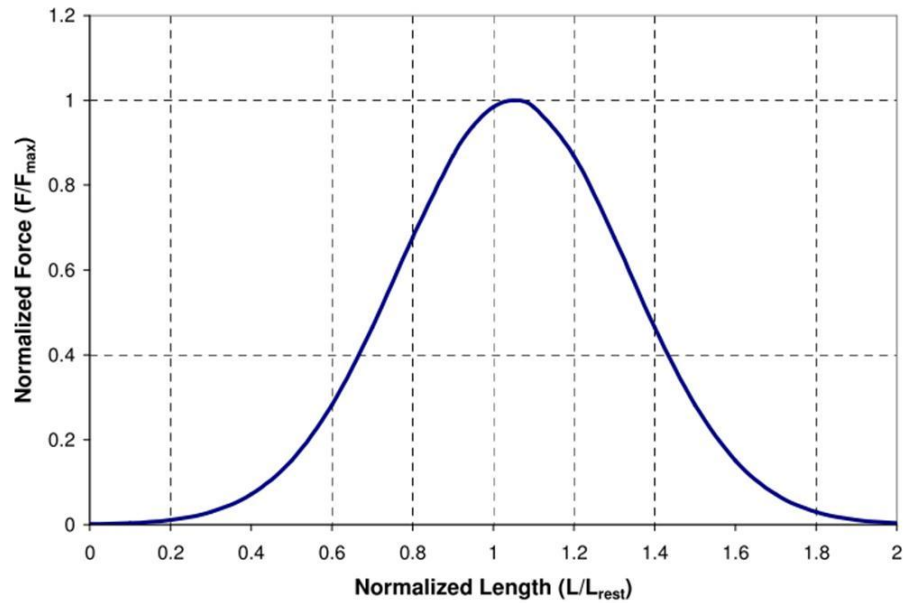


Figure 4-22: Active muscle force dependency on the length of muscle (Panzer, 2006)

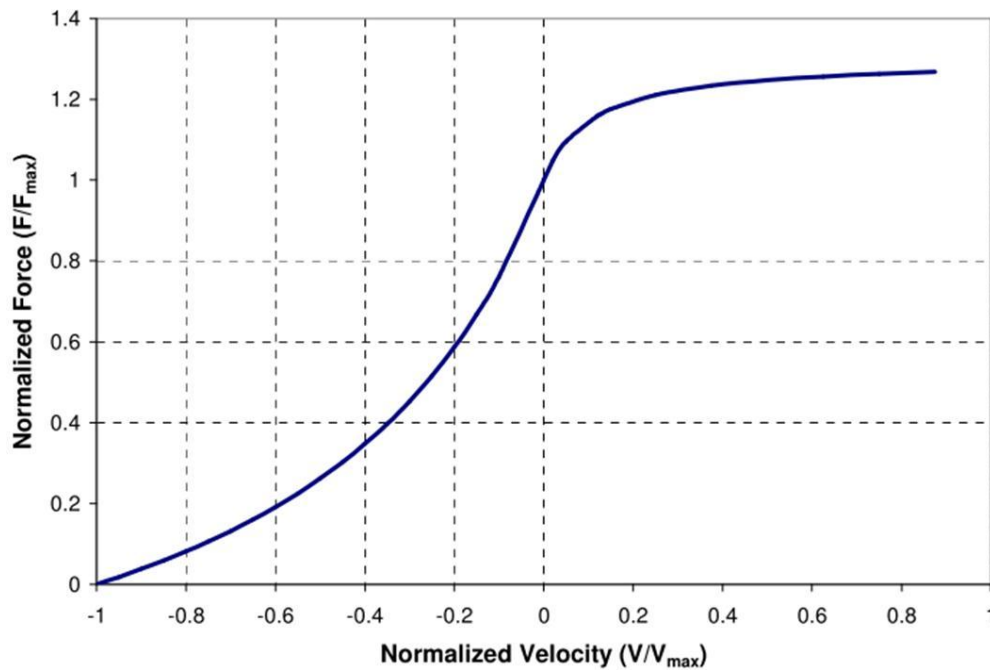
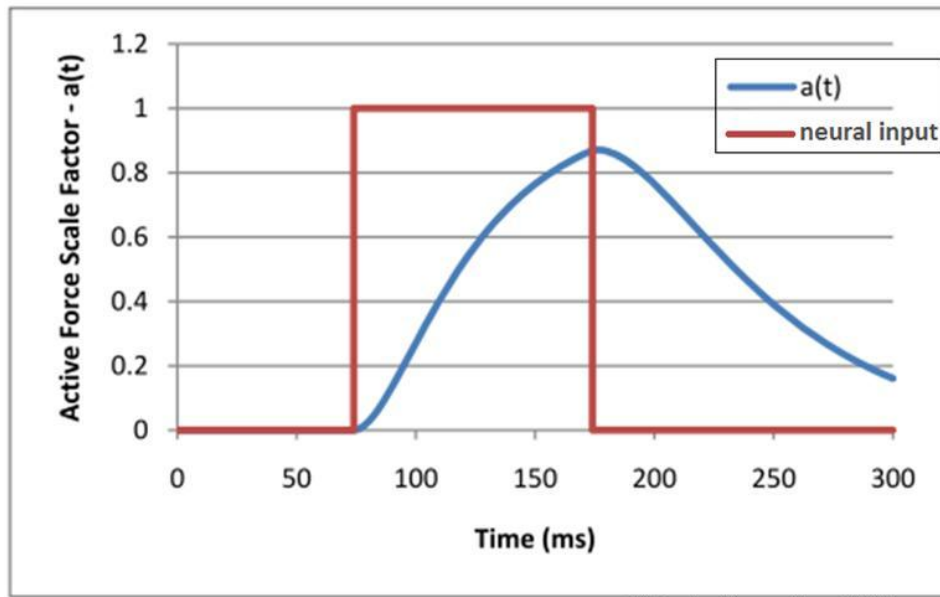


Figure 4-23: Active muscle force dependency on contraction velocity (Panzer, 2006)



(Adapted from: Fice, 2010)

Figure 4-24: Muscle activation function for a 100 ms step Neutral input starting at 74 ms (Fice, 2010)

The role of muscle activation has been studied in Finite Element models. Van der Horst (2002) studied the effect of muscle activation on frontal, rear, and lateral impacts. The findings of the simulations showed that muscle activation reduces head accelerations and rotations due to impacts. Brodin et al. (2005) investigated the role of muscle activation on frontal and side impacts and found that the rotation of the head is reduced as the result of muscle activity. Brodin et al. (2005) had also shown that the timing at which the different muscles are activated and the force production plays a role in the tissue strains resulted from the impact. The study on the UW model by Panzer (2006) also showed that the head acceleration during frontal impact was reduced as a result of muscle activity. Fice (2010) used the UW model to study the influence of muscle activation on rear impacts and found reduction on both head accelerations and ligament strains, especially in ALL and CL. The results of all these studies have shown that muscle activity improves the safety of the occupants during automotive collisions. The effect of muscle activation during out-of-position rear impacts will further be investigated in this thesis.

4.6 Model Enhancements

There have been a few improvements that were implemented to the model since the last full cervical spine validation by Fice (2010). These changes were made as a group decision by the members of the UW group who had worked on the cervical spine model. The changes includes ligament property changes in the upper cervical spine, addition of laxities to the upper cervical spine ligaments, implementation of step by step failure in the ligaments, and new muscle attachments.

The ligament properties of the upper cervical spine have been changed to those from the thesis by Mattucci (2011). The new material properties that were used for the upper cervical spine are illustrated in Table 4-5. Properties for the AAAM, PAAM, and the capsular ligaments in the upper cervical spine were also based on the work from Mattucci (2011), but were not published at the time. These materials properties, along with laxities, were added to improve the response of the upper cervical spine in all types of loading conditions.

Table 4-5: Upper cervical spine ligament properties

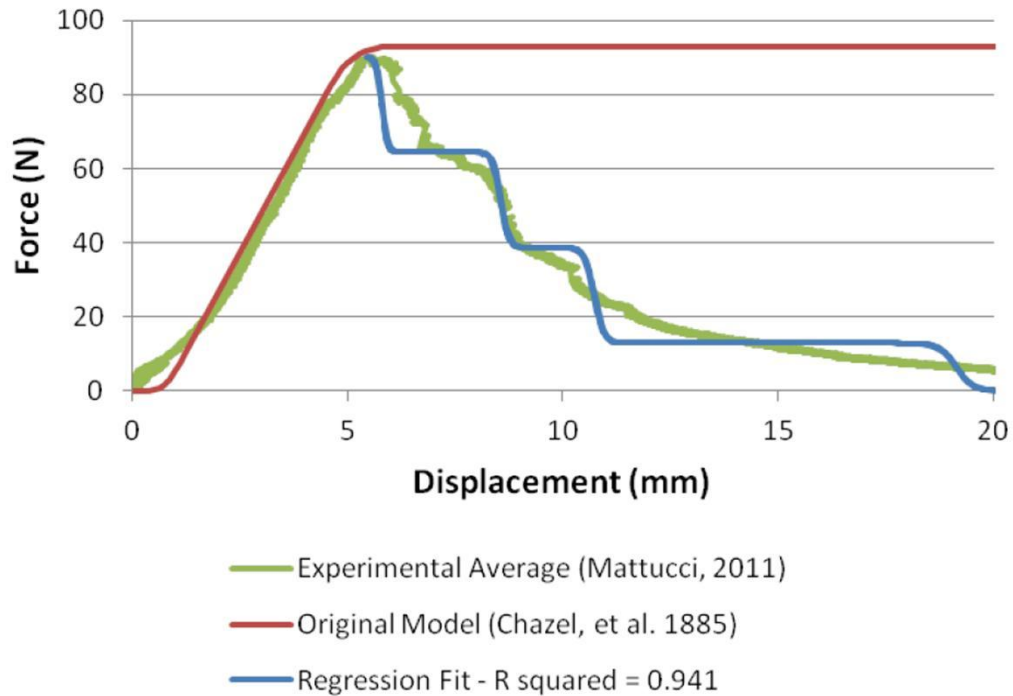
Ligament	Rate (s ⁻¹)	Failure Force (N)	Failure Elongation (mm)	Stiffness (N/mm)	Toe Region (mm)	Failure Stress (MPa)	Failure Strain	Modulus (MPa)	Toe Region Strain
TM Complex	Quasi (0.5)	1006 (416)	8.55 (2.57)	219 (51)	1.31 (0.38)	33 (12)	0.43 (0.10)	7.4 (2.5)	0.07 (0.01)
	High (150)	1582 (370)	4.71 (0.96)	590 (183)	1.25 (0.39)	46 (11)	0.27 (0.04)	17.3 (5.5)	0.07 (0.02)
	<i>p</i>	<i>0.08</i>	<0.01	0.03	<i>0.62</i>	<i>0.12</i>	0.04	<i>0.07</i>	<i>0.62</i>
TL	Quasi (0.5)	534 (116)	4.75 (0.31)	196 (22)	1.97 (0.07)	29 (11)	0.25 (0.08)	10.6 (3.0)	0.10 (0.03)
	High (150)	423 (76)	5.45 (0.36)	178 (42)	2.45 (0.30)	23 (6)	0.26 (0.03)	9.7 (2.9)	0.12 (0.02)
	<i>p</i>	<i>0.07</i>	<i>0.19</i>	<i>0.36</i>	<i>0.14</i>	<i>0.19</i>	<i>0.82</i>	<i>0.63</i>	<i>0.55</i>
AAOM	Quasi (0.5)	516 (192)	6.00 (1.32)	140 (45)	1.81 (1.06)	6.6 (2.3)	0.46 (0.10)	1.8 (0.4)	0.14 (0.07)
	High (150)	509 (108)	6.02 (1.74)	167 (64)	1.38 (0.73)	5.9 (1.4)	0.49 (0.16)	1.9 (0.7)	0.11 (0.04)
	<i>p</i>	<i>0.98</i>	<i>0.59</i>	<i>0.67</i>	<i>0.87</i>	<i>0.54</i>	<i>0.59</i>	<i>0.98</i>	<i>0.72</i>
PAOM	Quasi (0.5)	198 (80)	6.99 (2.30)	59 (28)	1.21 (0.71)	3.8 (1.0)	0.52 (0.13)	1.1 (0.4)	0.11 (0.06)
	High (150)	162 (52)	6.02 (4.18)	74 (54)	1.22 (0.54)	3.6 (1.3)	0.39 (0.20)	1.6 (1.0)	0.08 (0.05)
	<i>p</i>	<i>0.72</i>	<i>0.56</i>	<i>0.41</i>	<i>0.57</i>	<i>0.93</i>	<i>0.16</i>	0.05	<i>0.26</i>

Average (SD) values, with **significant** effects and *p-values* shown

(Adapted from Mattucci, 2011)

Compared to previous models that had been validated, the new model used progressive failure in the ligaments. In tensile loading, ligaments fail in groups and that causes step by step failure in the ligaments. Figure 4-25 shows how the model used the progressive failure in the ALL. The ALL was divided in 4 groups of beam elements and each group failed at a different displacement. The experimental observation by Mattucci (2011) had shown that most ligament failures initiated on the

outer edges and tore inward. This idea was used to model the order at which each group of ligaments failed in the model.



(Adapted from DeWit, 2011)

Figure 4-25: Post-failure regression fit for the ALL

The final change in the model was the muscle implementations. Previously, the model used rigid muscle connections to the vertebrae and the skull. In this implementation, the muscles were attached to a node that was rigidly attached to the vertebra. In the new implementation, the muscles are attached to the vertebra through spring elements and the attachment node is free to move. The new muscle attachments were meant to act as tendons to improve the biofidelity of the full cervical spine model.

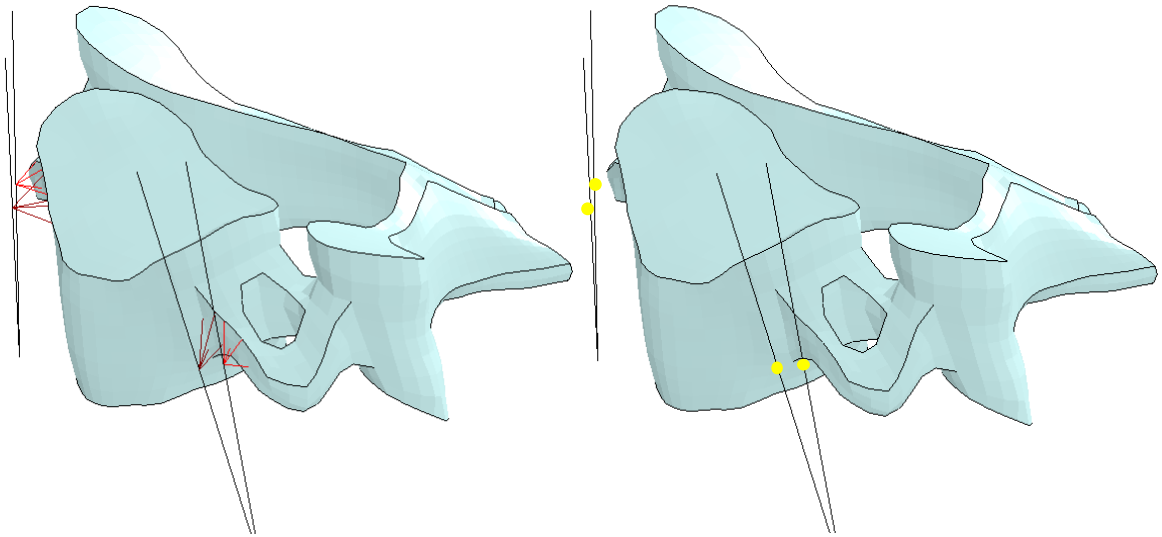


Figure 4-26: New muscle implementation (Left) & old muscle implementation (Right)

Chapter 5

Model Validation

This chapter explains the studies that have been used for the model validation. The model is validated against the upper and lower bound or one standard deviation away from the average values reported from in the literature. It must be noted that the error in the boundaries of the values reported, varies with the experimental procedures and the number of subjects tested; therefore limiting the quality of the validations process. The first section is a description of the validation at segment levels in axial rotation, extension, flexion, and tension. The second section focuses on the physiological range of motion of the whole cervical spine using the study by Ivancic et al. (2006b). The third section validates the model in neutral position rear impacts. The final section compares the out-of-position rear impact response of the model with the literature. In the reported figures of this chapter, x acceleration is positive in the anterior direction, z acceleration is positive in the superior direction, and the angle is positive in extension.

5.1 Segment Level Validation

In the latest segment level studies of the UW model by DeWit (2012), the middle and lower segments of the UW model were validated and it was suggested that some of the ligaments of the upper cervical spine require laxities in order to improve the biofidelity of the model. This section is the continuation of the work by DeWit (2012). The studies that were used to validate the upper cervical spine are summarized and the effect of laxities that were added to the simulations by DeWit (2012) to attain better behaviour for out-of-position impact scenarios are discussed. The loading conditions that were studied included axial rotation, extension, flexion, and tension. The major parameters that were studied were the force and the displacement/angle at which the failure occurred for each loading case, as reported in the literature. Because of the focus on out-position neck injury, the force at range of motion and the relative rotation of the C0-C1 and C1-C2 vertebrae were also validated.

The large number of different ligaments in the upper cervical spine and the amount of time required to run each simulation for each loading case posed challenges in the validation procedure. A few steps were taken to expedite the validation process. Laxities of -2, 2, and 4 mm were applied to all the ligaments of the upper cervical spine in order to find the primary and secondary effect of each ligament on the four loading cases. This procedure helped to identify the sensitivity of the parameters

that were studied from each loading case to all the upper cervical spine ligaments. Furthermore, during the validation process, the laxities were studied in increments of 0.5mm, unless the ligament was very sensitive to a loading type. This had a great impact in reducing the total time required for the validation.

The parameter study on individual ligaments allowed for a better understanding of how each ligament affects each loading case. The limitation to this study was the lack of information it provided on how the interaction of ligaments effect each loading case. Due to time constraints, the effect of the combined ligament laxities could not be covered in this thesis. The results of the study are summarized in Table 5-1. Pretension in some of the ligaments caused non-physical behaviour of the segments during the loading cases. Therefore the table only summarizes the effect of laxities between zero and four. If the laxity of 2mm in a ligament changed the average difference of force or displacement/rotation at failure between 5-10%, it was labelled as a secondary effect and if the change was higher than 10%, it was labelled as a primary effect.

Table 5-1: Study of laxity on individual ligaments

Extension		Flexion		Tension		Axial			
Failure		Failure		Failure		Failure		Range of Motion	
Primary	Secondary	Primary	Secondary	Primary	Secondary	Primary	Secondary	Primary	Secondary
AA-OM	PA-OM	CL01	Alar-OC	AA-AM	AA-OM	AA-AM	CL01	AA-OM	
AA-AM	PA-AM	CL12		CL01	Alar-OC	Alar-OC		Alar-OC	
Alar-OC	TM	PA-OM			CL12			Apical	
CL01								CL01	
CL12								CL12	
								ISL	
								PA-OM	
								PA-AM	
								TM	

In extension, AA-OM, AA-AM, Alar-OC, and capsular ligaments had primary effects and PA-OM, PA-AM, and TM had secondary effects on the deformation or force to failure. In flexion, the capsular ligaments and the PA-OM had primary effect and Alar-OC had secondary effect on the failure load. In tension, the anterior ligaments, Alar-OC, and capsular ligaments affected the failure response. It was later discovered that the combination of laxities in all the upper cervical spine ligaments could change the failure response behaviour in tension. In axial rotation, the ligaments that had affected the failure torque were the AA-AM, Alar-OC, and the CL01 ligaments. The force at range of motion was controlled by the CL01 and the posterior ligaments and the C01-C12 rotation ratios were controlled

by many of the ligaments. The capsular ligament of the atlas and the axis had the largest contribution to the rotation ratio of the upper cervical spine. The axial rotation case had to be validated in both failure mode and range of motion. Because of the significance of the CL12 ligament in axial rotation and the primary role it played in all other loading cases, the axial rotation loading case was selected to be the first case of validation, followed by extension, flexion, and tension cases.

The procedure that was used for the validation process is outlined in the flow chart in Figure 5-1. The effect of individual ligaments on the loading cases were studied and used as a guideline to validate the upper cervical spine model. The validation started with testing the model in axial rotation and validating it against the experiment by Goel et al. (1990) in failure and range of motion. Once the model was validated in the axial case, it was tested against the experimental data by Nightingale et al. (2007) in extension and flexion and against Dibb et al. (2009) in tension. The validation procedure ensured that the ligament laxities were tested against all loading cases. After the model was fully validated against all the cases, the new ligament laxities were implemented within the full cervical spine model in order to be further validated against the physiological range of motion experiment by Ivancic et al. (2006b).

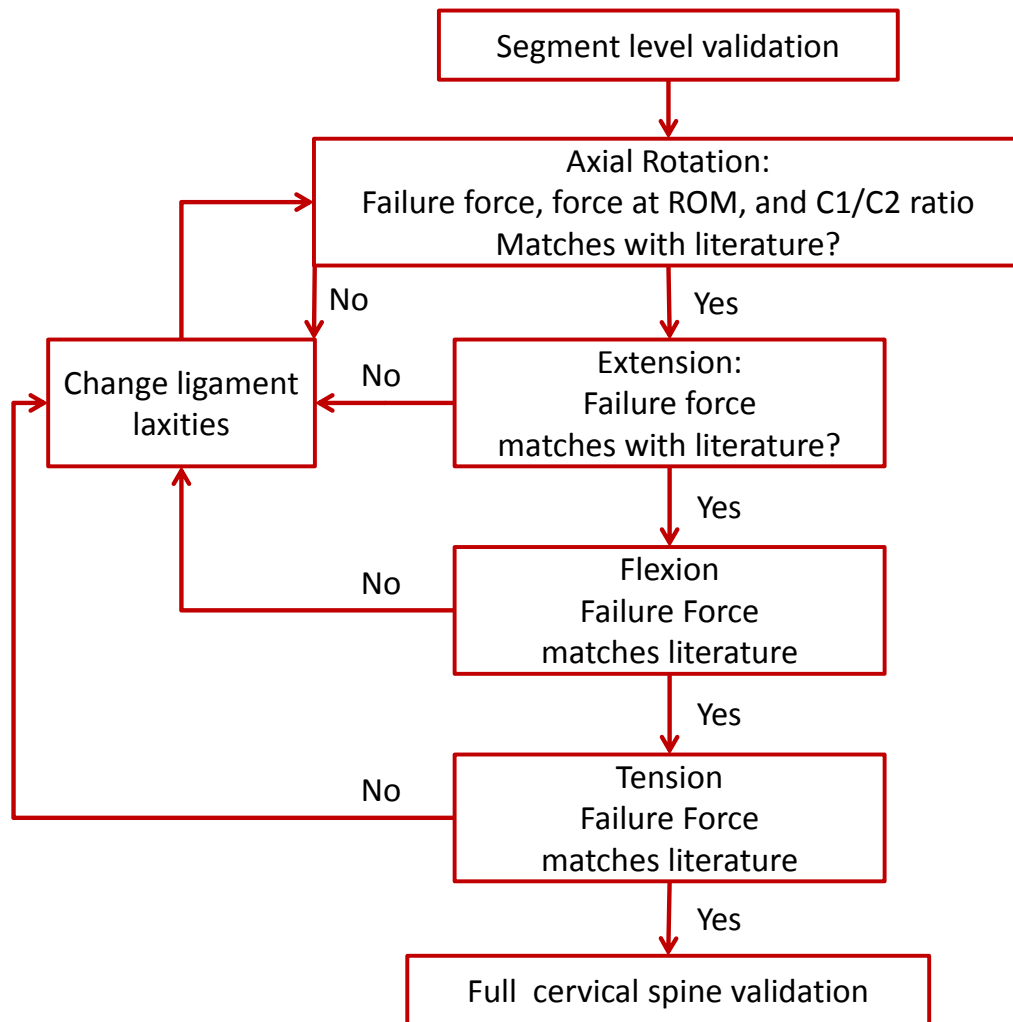
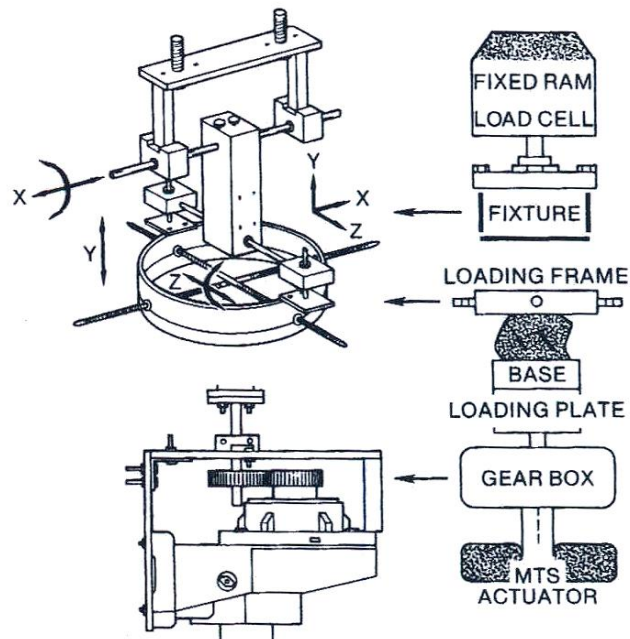


Figure 5-1: Validation process of segment level studies

The study that was used to validate the axial rotation of the upper cervical spine was done by Goel et al. (1990). In this study the C012 segments were dissected from the cervical spine. The musculature was also removed in order to test the behaviour of ligaments in the upper cervical spine. The test was performed using an apparatus shown in Figure 5-2. The C2 segment was fixed in all directions except the axial rotation. The skull was attached to the apparatus so that it was constrained in the axial rotation motion but free to move and rotate in all other directions. Twelve cadaver segments were tested and the average torque versus axial rotation for the experiment in range of motion was 2 Nm (at 29.9°) and 13.6 Nm (at 68.1°). The simulation setup was previously done by DeWit (2012). In the simulation setup, the boney plate of the C2 vertebra and the skull were used to apply the appropriate boundary conditions outlined in the experiment.



(Adapted from: Goel et al., 1990)

Figure 5-2: Segment level axial rotation test apparatus

Numerous iterations were performed to validate the segment model. The result from the final validation of the axial rotation is shown in Figure 5-3. The failure force and the angle at which the failure occurred were not changed significantly with the addition of laxities. The failure force and angle of the model with the new laxities are within the standard deviation of the study from Goel et al. (2009). The high values of the failure force could be explained through the younger ligament properties that were implemented in the model. The addition of laxities brought the torque in axial rotation much closer to the experimental value (Figure 5-4). The ratio of C01 and C12 rotation was calculated based on the literature and was found to be between 5-20% (Dugailly et al., 2010; Dvorak et al., 1987; Ishii et al., 2004; Ivancic et al., 2006b; Panjabi et al., 1988). Figure 5-5 shows the improvement that was obtained through the implementation of the laxities to the model.

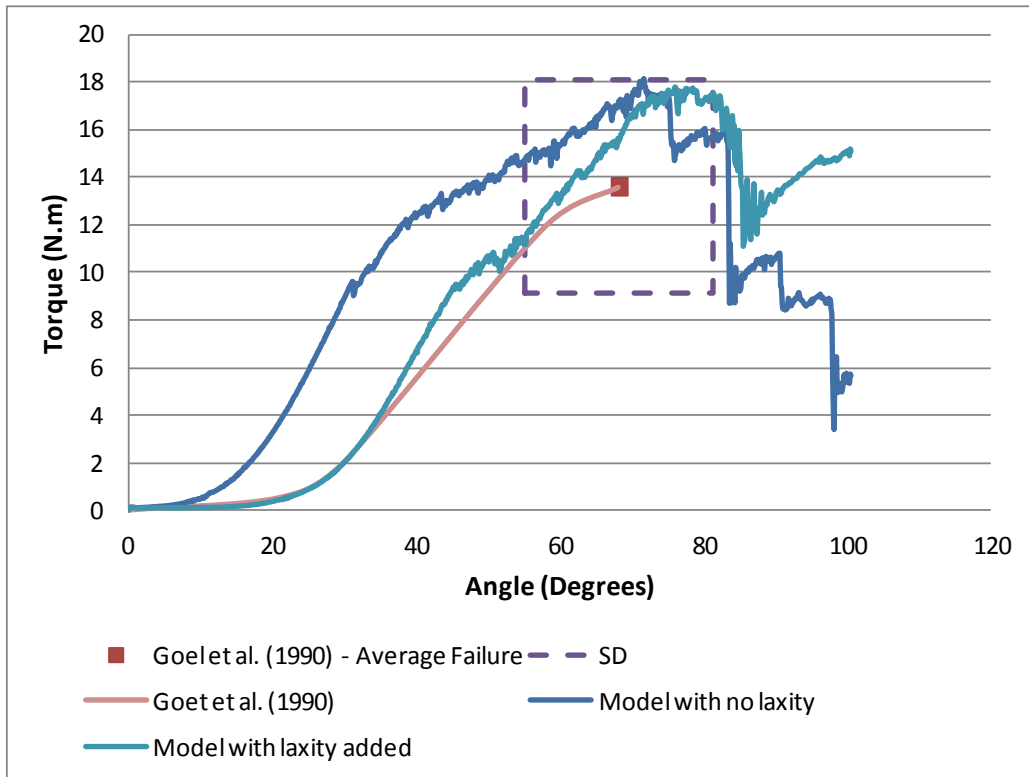


Figure 5-3: Segment level axial rotation simulation results

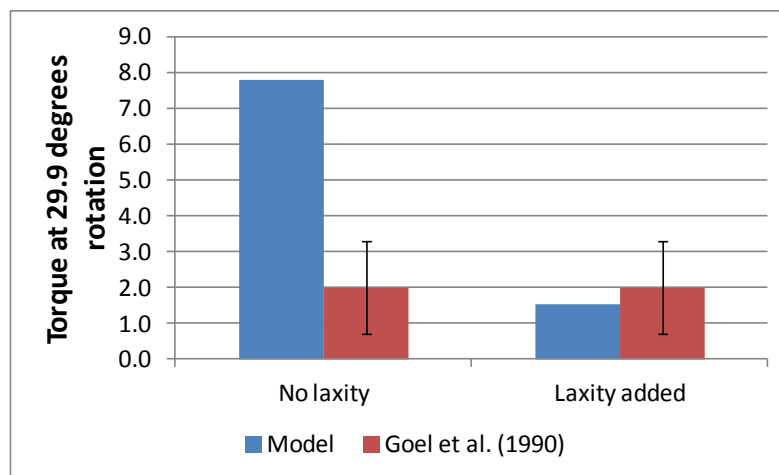


Figure 5-4: Torque comparison of axial rotation in range of motion

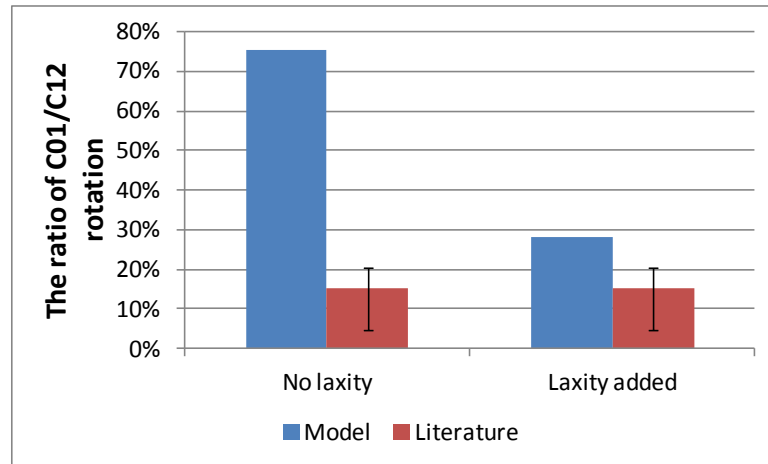
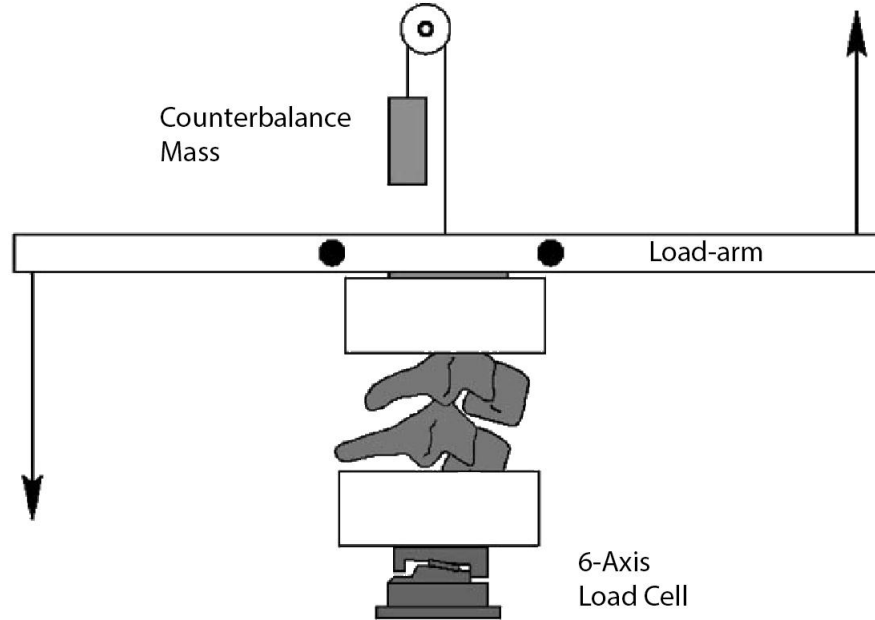


Figure 5-5: Comparison of C01/C12 ratio between the model and the average and range from the literature

The experiments that were used for the validation of flexion and extension loading cases were performed by Nightingale et al. (2002, 2007). The 2002 study was focused on a female population whereas 2007 study had a male sample group. The experiment that was used in this thesis was the 2007 study because the model represents a 50th percentile male. The apparatus that was used for the study is shown in Figure 5-6. The cephalad end of the segments were secured using a halo fixation and the extension and flexion moments were applied to the casting of the C2 vertebra. The purpose of the counterbalance was to ensure identical initial starting position between each run. To minimize the effect of shear and tensile loads, the counterbalance was removed during the failure test. The average failure moment and mean angle of 39 Nm at 58.7° and 49.5 Nm at 42.4° were reported for flexion and extension, respectively.



(Adapted from: Nightingale et al., 2002)

Figure 5-6: Segment level extension and flexion test apparatus

The simulation procedure by DeWit (2012) was used for both extension and flexion validations. In the simulations, the head was fully constrained in all directions and a pure rotation was applied to the bony plate of the C2 vertebra. Figure 5-7 shows the results from the original ligament properties and the results from the addition of laxities. The curves show that the addition of laxity has lowered the force during range of motion. The failure load and the rotational angle at which the failure occurs have increased but remain in the standard deviation of the results from Nightingale et al. (2007). Similarly in flexion, the load during range of motion is reduced and the failure load and rotation angle at which failure occurs were increased (Figure 5-8). In the case of flexion, the addition of laxity moved the failure load and the rotational displacement within the standard deviation of the experimental data. The increase in the angle and force in failure could be explained through the implementation of younger ligament properties in the upper cervical spine.

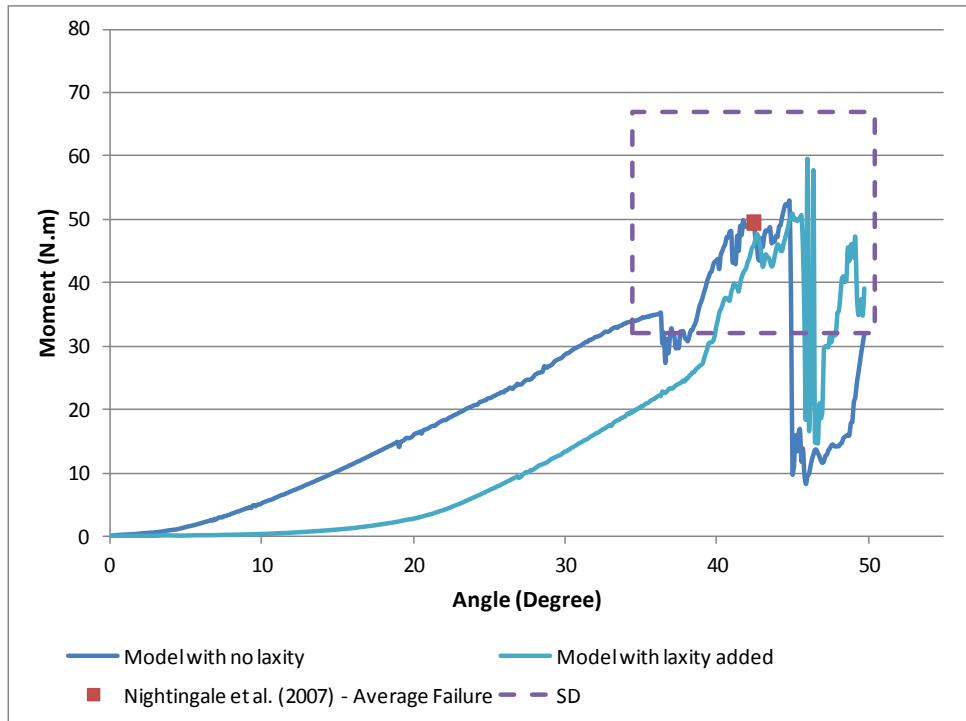


Figure 5-7: Segment level extension simulation results

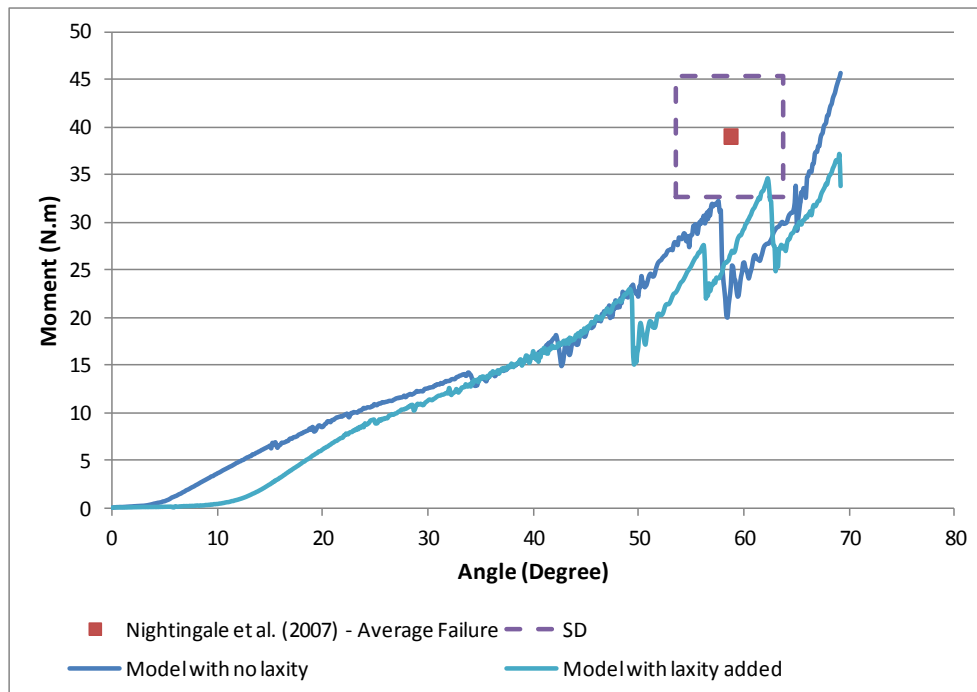
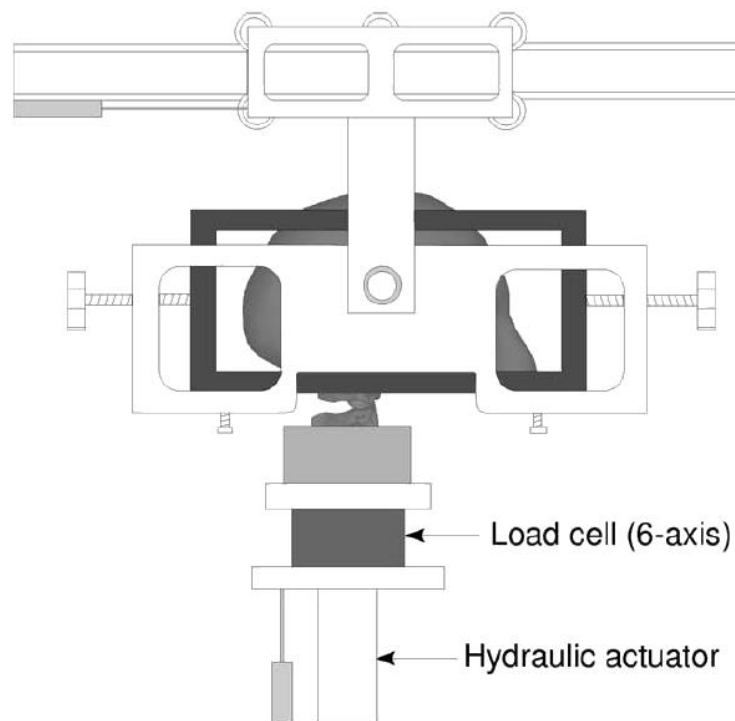


Figure 5-8: Segment level flexion simulation results

The validation of the upper cervical spine in tension was made possible by an experiment from Dibb et al. (2009). The apparatus used for the experiment is shown in Figure 5-9. The design of the apparatus allowed the segment to be pulled in tension from the inferior side. The segments were mounted so that the inferior end was fully constrained and the superior end could translate in anterior and posterior directions and rotate in sagittal plane. Seven cadaver segments were tested and the mean failure force for the upper cervical spine was found to be 2417 N at an average failure displacement of 10.8 mm.



(Adapted from: Dibb et al., 2009)

Figure 5-9: Segment level tension test apparatus

The simulation used for validation of the upper cervical spine in tension was developed by DeWit (2012). The skull and the base plate of the C2 vertebra were modeled with similar constraints as those of the experimental setup and a displacement boundary condition was applied to the C2 to replicate tension in the segment. The results from the original ligament properties and those with laxities are shown in Figure 5-10. The failure force is slightly higher than the average but within the standard deviation of the experimental data. The higher force is believed to be due to the younger age ligament

properties that were used in the cervical spine model. The displacement at which the failure occurs was shifted beyond the standard deviation which was noted in the study by Dibb et al. (2009). All ligaments have a contribution to the displacement at failure in the tension test. The shift in the value is due to the addition of laxities that were introduced to the model. The difference in the values could be explained through preloads that might have been applied to the samples that were not reported in the manuscript.

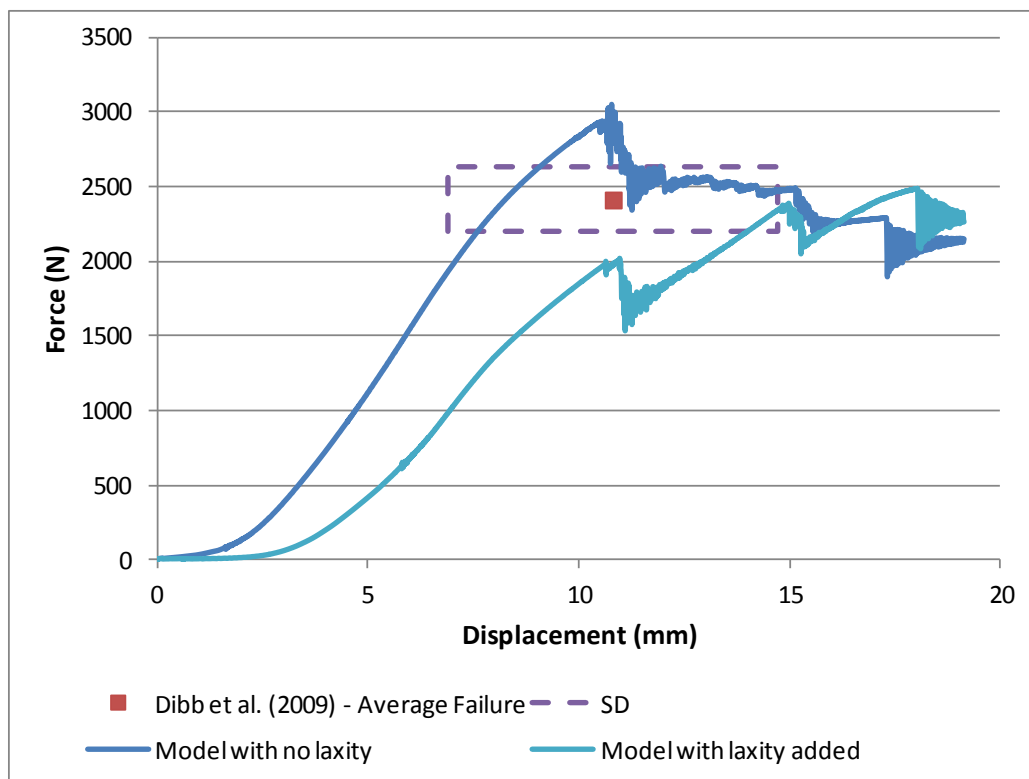


Figure 5-10: Segment level tension simulation results

The addition of laxities to the upper cervical spine improved the behaviour of the model in axial rotation, extension, flexion, and tension cases. The laxities that were used in the segment level validations were then used in the full cervical spine model in order to test the physiological range of motion behaviour.

5.2 Full Cervical Spine - Physiological Range of Motion Validation

The study of Ivancic et al. (2006b) was used to validate the model in a physiological range of motion. In the study, six whole human cervical spine specimens with an average age of 80.2 years (range:

79-93) were used without musculature. A custom surrogate head with a mass of 3.3 kg and sagittal, horizontal, and frontal plane moments of inertia of 0.019, 0.014 and 0.015 kg.m² was rigidly attached to the occipital mount. A three plane flexibility testing was used for measuring the physiological range of motion of the cervical spine. Unconstrained pure moments of 3, 1.5, and 1.5 Nm in axial, flexion-extension, and lateral directions were applied to the head in four equal steps while the T1 vertebra was fixed. Except the T1, all other parts of the cervical spine were allowed to move naturally during the head rotation. The resultant data were used to calculate the range of motion and the neutral zone at each intervertebral level.

The musculature in the cervical spine model was removed in order to mimic the experimental procedures of Ivancic et al. (2006b) (Figure 5-11). The mass and inertial properties of the skull were altered to those that were used in the experiment. The T1 vertebra had fixed constraints while all other parts of the cervical spine were free to move. The total head rotations of 66.8° in axial rotation, 54.6° in extension, 51.7° in flexion, and 30.4° in lateral bending that were calculated from data by Ivancic et al. (2006b) were applied to the skull and the intervertebral rotations were output. These values were compared to those reported from the experimental data in order to validate the full cervical spine model against physiological range of motion.

In each loading case, the difference between the rotation values of the model and the average±SD of the experimental data was calculated for all intervertebral bodies. The sum of these differences was divided by the total head rotation of the loading case. If the resultant value was less than 5%, the laxities were accepted; otherwise the validation process had to start again at the segment level.



Figure 5-11: Physiological range of motion simulation constrains

Approximately five iterations were performed between the segment level and the full cervical spine model to find the appropriate laxity values. The final laxity values that were determined for the model are shown in Table 5-2. As discussed previously, the laxities provided good results with respect to the literature in the segment level. The results from the full cervical spine simulations are summarized in Figure 5-12 to Figure 5-15. The results are expressed in ratio of intervertebral rotation to the total head rotation. This type of illustration allowed for the comparison of axial rotation and lateral bending between the model and the experimental data in both left and right directions. The results show that the addition of laxities improves the physiological range of motion of the model in the four loading conditions which were considered. In axial rotation, the results are within the standard deviations for all levels except C3-C4. The total difference with respect to the total rotation was calculated to be 2.1% as compared to 26.9% with the original ligament properties. In extension, segment levels of C3-C4 and C6-C7 are slightly outside of the standard deviation but all other segments are within the boundaries which were reported by the literature. The total difference with

respect to the total rotation was calculated as 1.9% as compared to 39.4% in the model without laxities. In flexion, the model was within the standard deviation of the literature in all intervertebral levels. In lateral bending, the C0-C1 and C7-T1 rotations are slightly away from the standard deviation of the reported values. The difference with respect to the total rotation is calculated as 2.5% as compared to 6% in the model without laxities. Overall, the addition of laxities improved the behaviour of the full cervical spine in the physiological range of motion.

Table 5-2: Final laxity values

Laxity (mm)	Ligament					
	AA-OM	AA-AM	CL12	ISL-12	PA-OM	PA-AM
	5	1.4	6	3	2	2

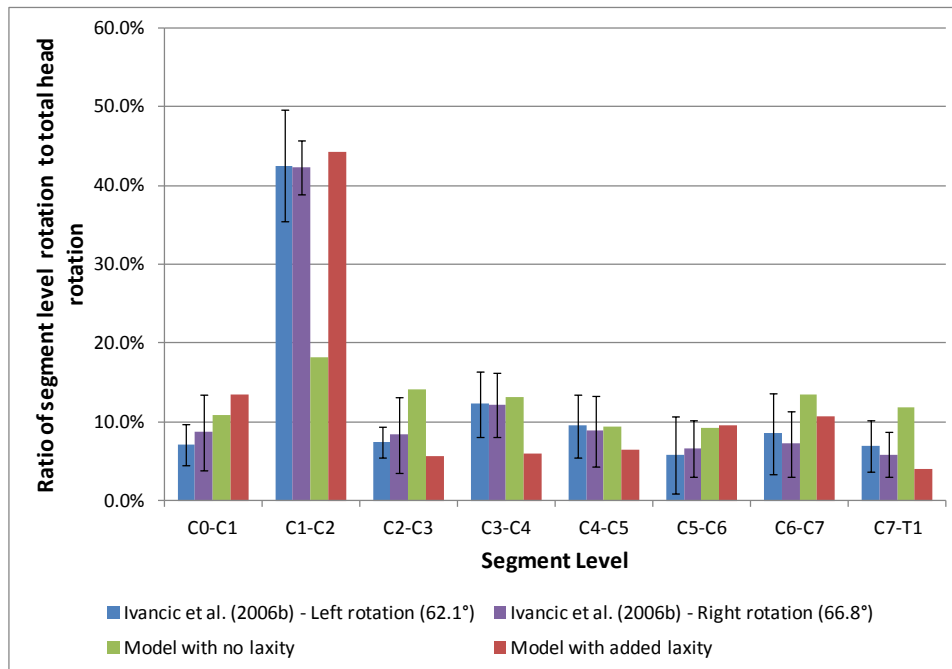


Figure 5-12: Axial physiological range of motion results from full cervical spine simulation

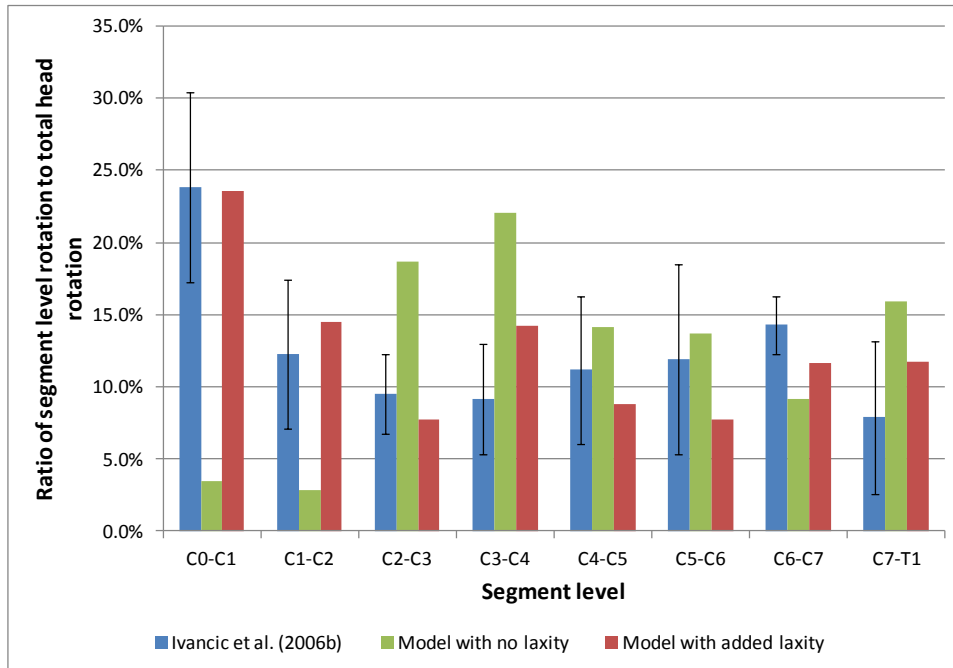


Figure 5-13: Extension physiological range of motion results from full cervical spine simulation

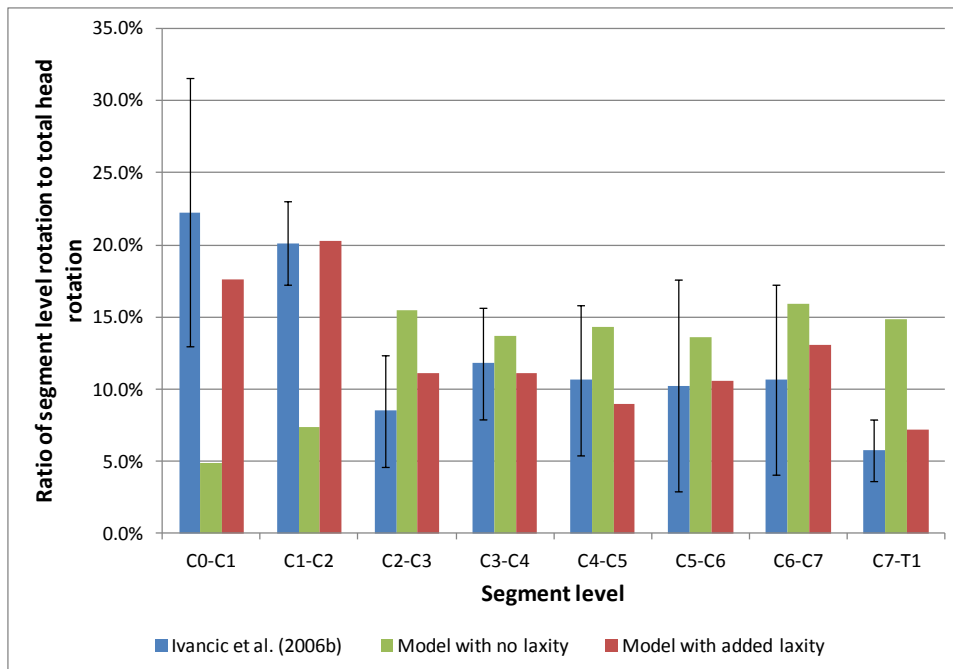


Figure 5-14: Flexion physiological range of motion results from full cervical spine simulation

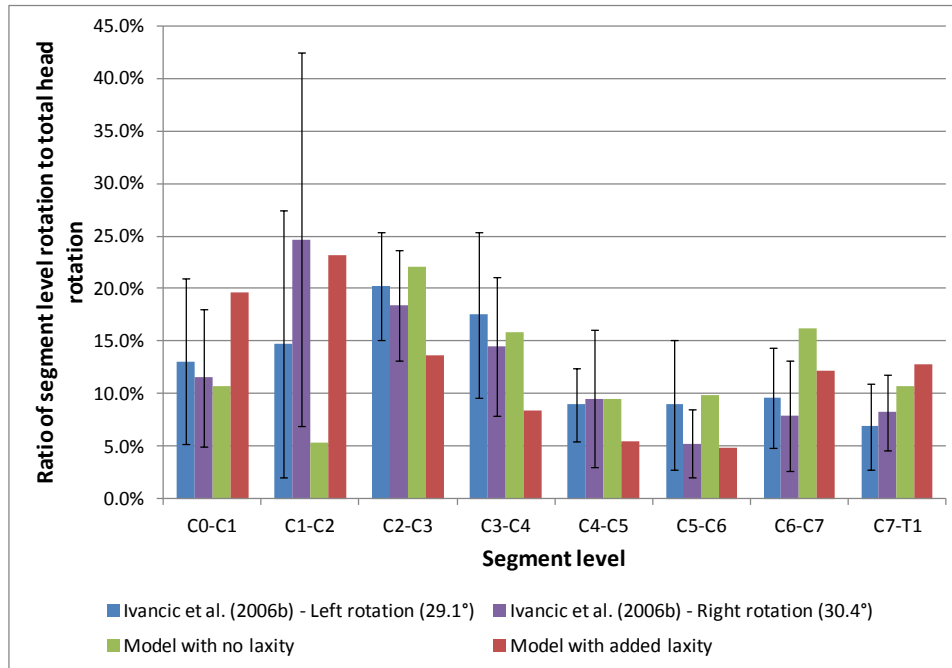


Figure 5-15: Lateral physiological range of motion results from full cervical spine simulation

5.3 Rear impact validations

Rear impact validation was necessary to ensure that the addition of laxities in the upper cervical spine had not negatively altered the model behaviour in impact. The model was simulated in rear impact to validate both kinematics and tissue strains behaviours.

A study by Deng (1999) was selected for the kinematics validations. The detailed procedure of the methodology used to evaluate the kinematics of the model is outlined in the thesis by Fice (2010). In the study by Deng (1999), 26 rear impacts were undertaken on 6 whole body cadavers at accelerations ranging from 5.0 to 9.9 g. The average (SD) age of the cadavers was 72.3 (20.5). The acceleration and rotation of the head and the T1 vertebra were measured by accelerometers and rotational rate sensors. A 7 g sled peak acceleration was simulated and the results were compared to the kinematics of sled peak accelerations of 6-8 g. The accelerations and rotations that were applied to the T1 are outlined in Figure 5-16. The T1 vertebra was constrained in all other directions and all other bodies in the cervical spine were free to move in all directions. Muscle activation was turned off in the simulation to mimic the cadavers' behaviour.

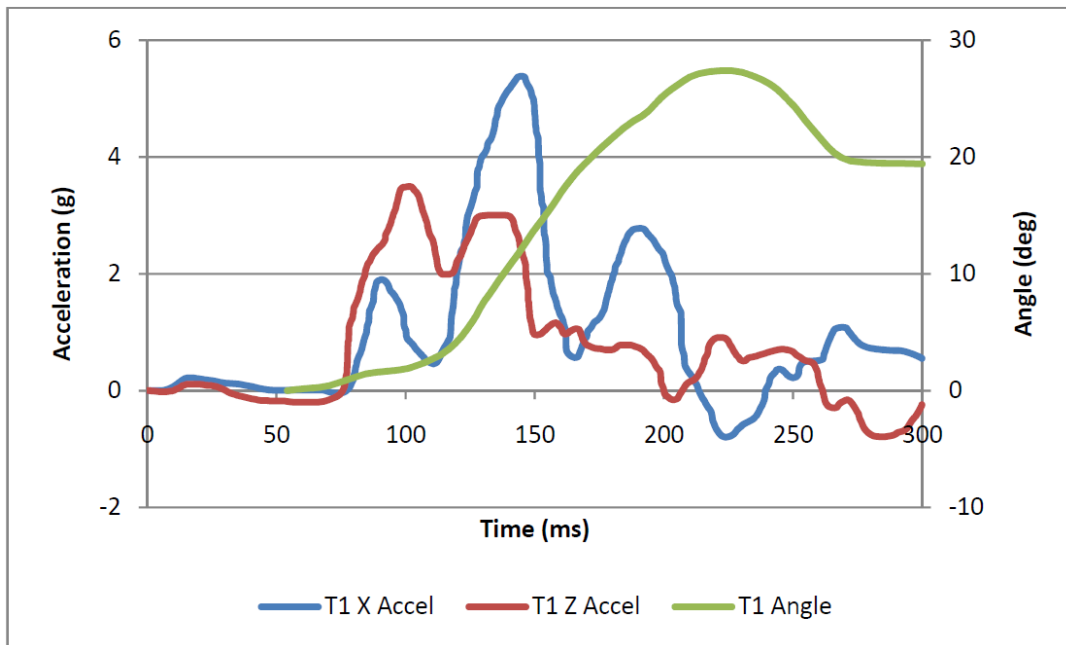


Figure 5-16: T1 prescribe motion for the 7 g rear impact (Deng, 1999)

The head rotation of the model during the 7 g rear impact is illustrated in Figure 5-17. The shape of the rotation curve was representative of the experimental data. The peak head extension was slightly higher than the experimentally measured value. The addition of the laxities resulted in a slight delay in the start of the rotation. The new muscle implementation did not have a significant effect on the angle of the head during rotation.

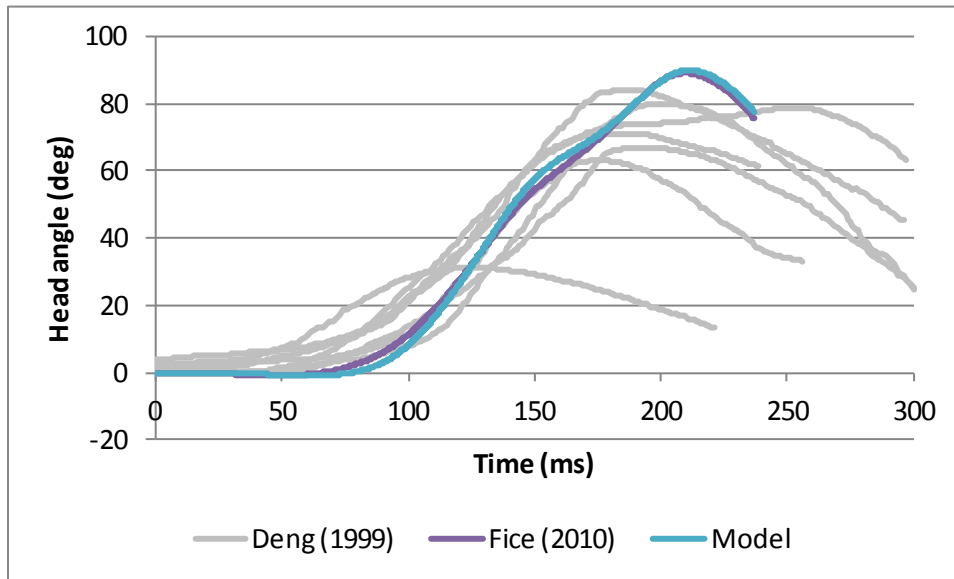


Figure 5-17: Head rotation of the model during 7 g rear impact

The head centre of gravity (CG) acceleration in the anterior direction was very similar to the validation results from Fice (2010) (Figure 5-18). The changes in the model did not have a significant effect in the acceleration of the head in the anterior direction. The model keeps a good representative shape compared to the experimental data from Deng (1999). In the superior direction, the peak acceleration of the model had increased from Fice (2010) due the changes of the model (Figure 5-19). The rate of velocity could increase in the regions of the ligaments where laxities are introduced. The new implementation of the muscles also allows more flexibility in the movement of the skull. This is because the attachment nodes are no longer restricted with the movement of the vertebrae. The resistance that limits the head movement is reduced due to the additional laxities in the ligaments of the upper cervical spine and the new muscle implementation; thus explaining the higher head acceleration during simulation. Even with the increase in the peak acceleration, the model was in good agreement with the shape and peak accelerations which were observed in the experiments and followed closely with the validation model by Fice (2010).

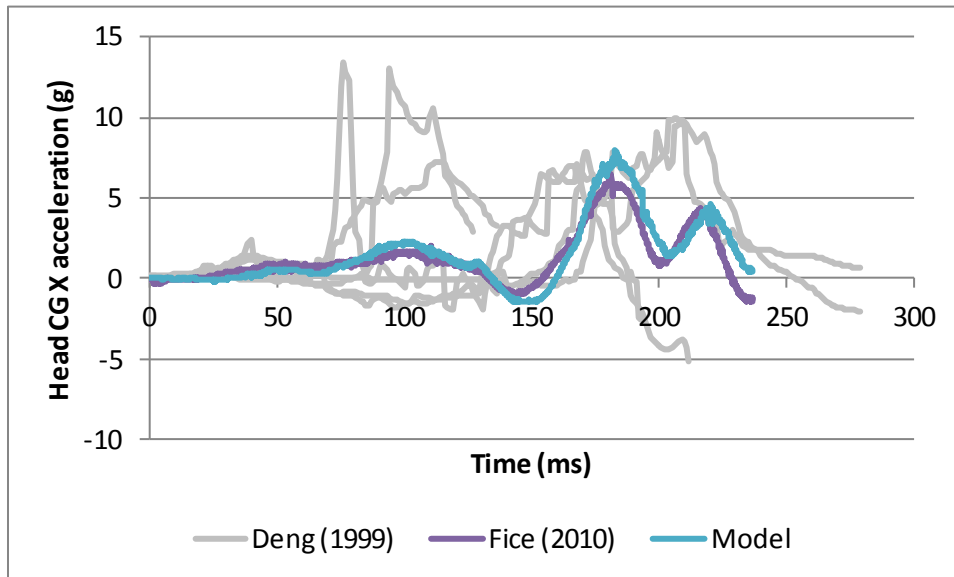


Figure 5-18: Head CG X acceleration of the model during 7 g rear impact

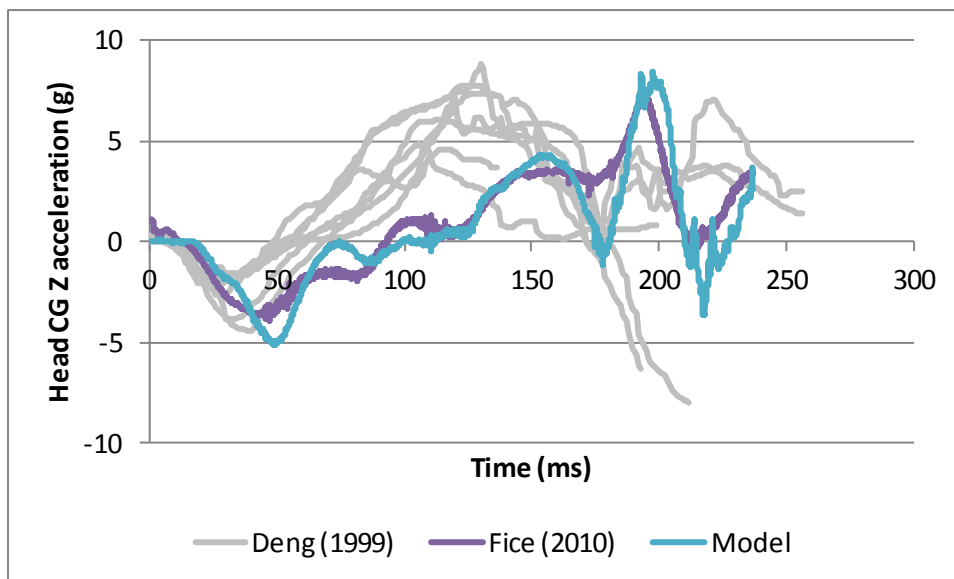


Figure 5-19: Head CG Z acceleration of the model during 7 g rear impact

The relative rotations of the intervertebral bodies were also validated against the literature. The shape and the peak values of the model were in good agreement with the experimental data that was provided by Deng (1999). The relative rotation of the C1-C2 vertebrae was higher than the model which was simulated by Fice (2010), however, within the boundary of experimental data. The

difference could be explained through the laxities that were introduced in the upper cervical spine ligaments, especially, the PA-AM and the capsular ligaments.

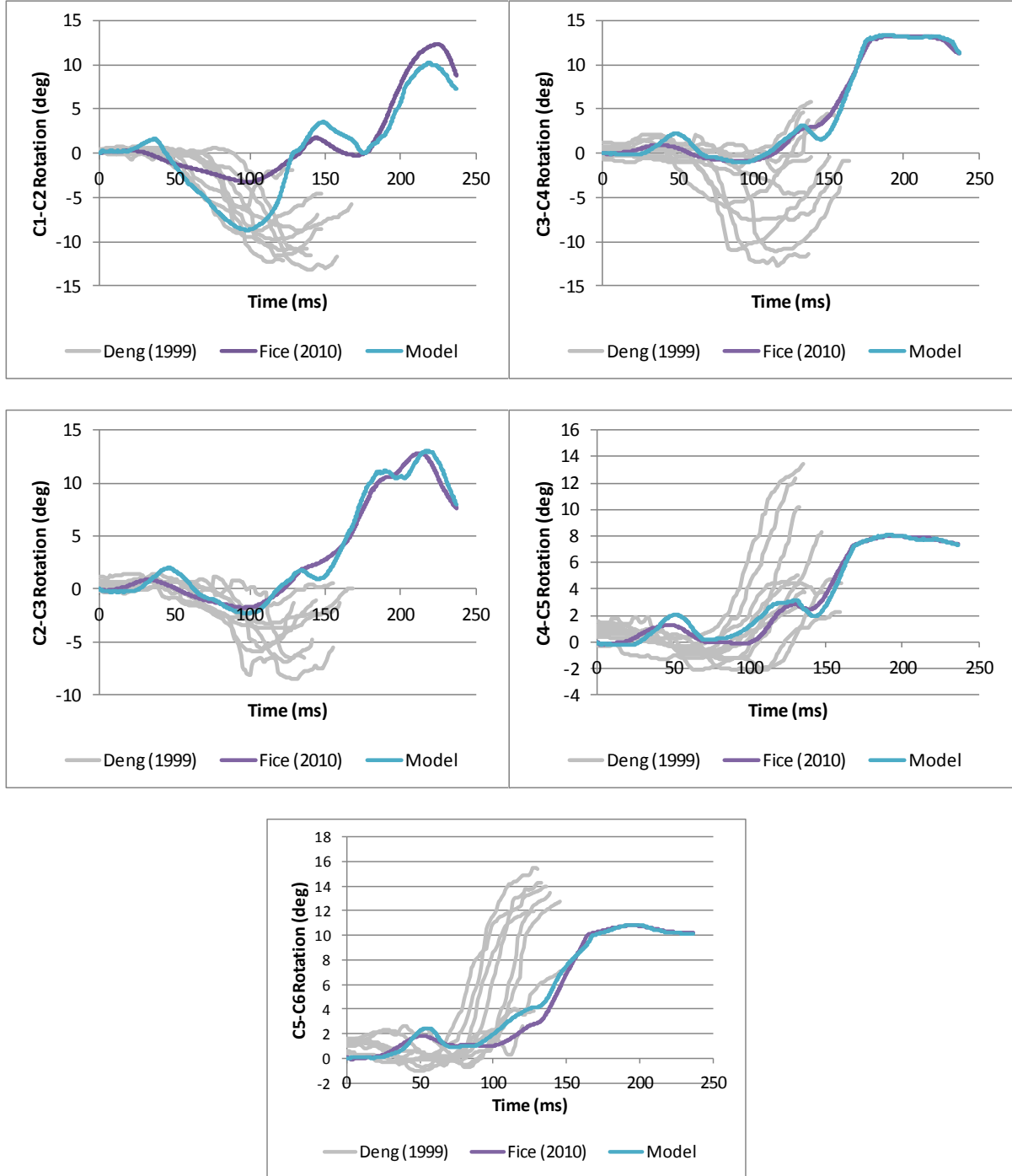
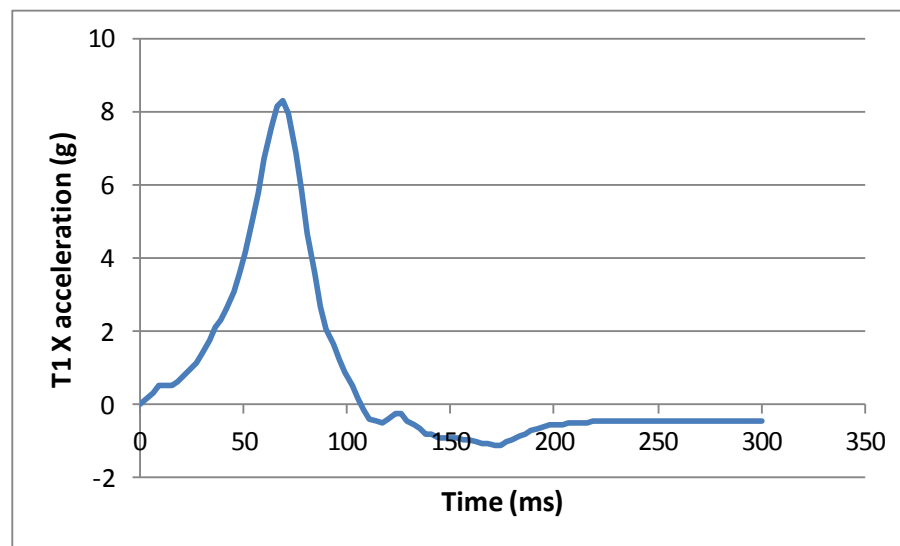


Figure 5-20: Intervertebral rotation validation during 7 g rear impact

The detail information on the studies and simulation methodologies used to validate the model in tissue levels in rear impact are outlined in thesis by Fice (2010). The studies by Ivancic et al. (2004), Ivancic et al. (2005b), Pearson et al. (2004), and Panjabi et al. (2004b) were used for soft tissue validations. In these experiments, cadaver models of a whole cervical spine from T1 to C1 were studied. A surrogate head with mass of 3.3 kg was attached to the C1. A cable and spring system was used to simulate the passive musculature in the cervical spine. Cables ran through wire loops which were attached to each vertebra and connected to springs with stiffness of 4 N/mm. The cadaver was subjected to rear impact and the strains in ligaments were measured. Active musculature was turned off since the experiment did not include muscle activity. The T1 vertebra was constrained in all directions except for the anterior and posterior direction. All other cervical spine properties were free to move in all directions. T1 acceleration of 8 g which is shown in Figure 5-21 was applied to the T1.



Adapted from Fice (2010)

Figure 5-21: T1 acceleration profile to simulate the 8-g rear impact

The ALL strains were compared to results from a study by Ivancic et al. (2004) and the CL strains were compared to results by Pearson et al. (2004). All the results between the C2-C7 vertebrae except the ALL strains at C4-C5 level were within the standard deviation of the experimental data (Figure 5-22). There was very small difference between the new model and the one simulated by Fice

(2010). The higher CL strain in C6-C7 could be due to the higher relative rotation of the vertebrae which is caused with the new muscle implementation.

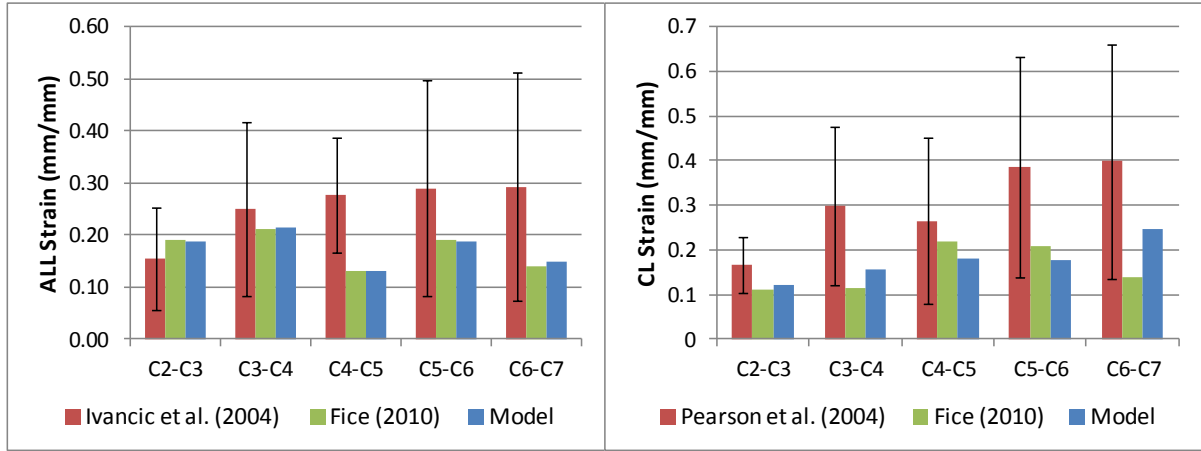


Figure 5-22: Peak ligament strain comparison during 8 g rear impact

The disc shear strains were compared to the study by Panjabi et al. (2004b). The shear strains were measured in the anterior, middle, and posterior side of each disc (Figure 5-23). The shear strains in the model for C2-C7 vertebrae were all within the standard deviation of the experimental data except for the C5-C6 vertebra. There were no significant differences in the disc strains between the results of the current model and the simulations performed by Fice (2010).

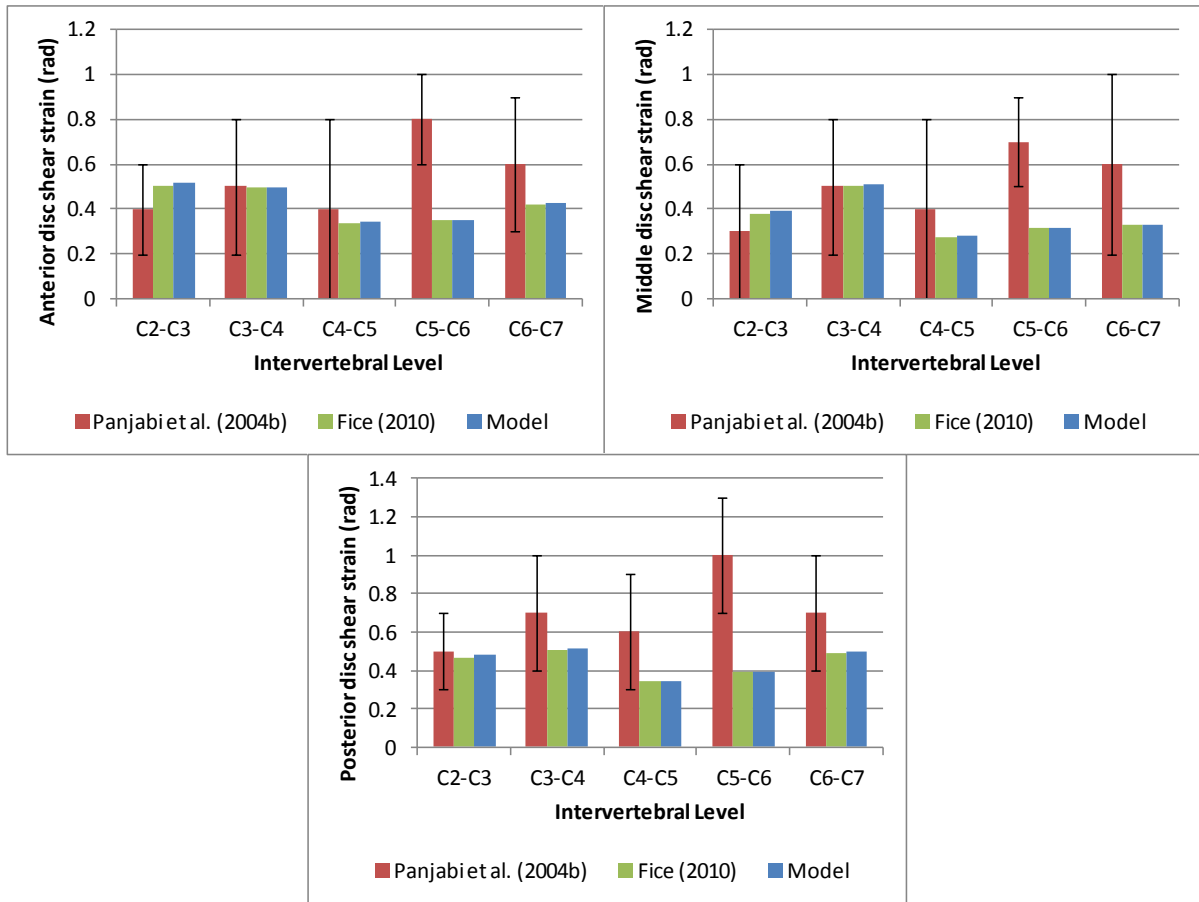


Figure 5-23: The peak disc shear strain during 8 g rear impact

The model was also validated against low impact levels which were conducted on human volunteers. The detail background for the methodology to simulate the experiment by Davidsson et al. (1998) is outlined in the thesis work by Fice (2010). Davidsson et al. (1998) experimented on 28 human volunteers at an average peak sled acceleration of 3.6 g. The volunteers sat on a laboratory seat which included a head rest and were impacted by applying a pulse to the seat. The acceleration and rotation data were captured by sensors that were attached over the skin of the volunteers.

A non-linear viscoelastic model of head rest was used for the simulation. The average sled anterior acceleration and the T1 accelerations from the study were used to simulate the rear impact (Figure 5-24). The skull and all other vertebrae were free to move in all directions. The size of the model relative to the headrest is shown in Figure 5-25.

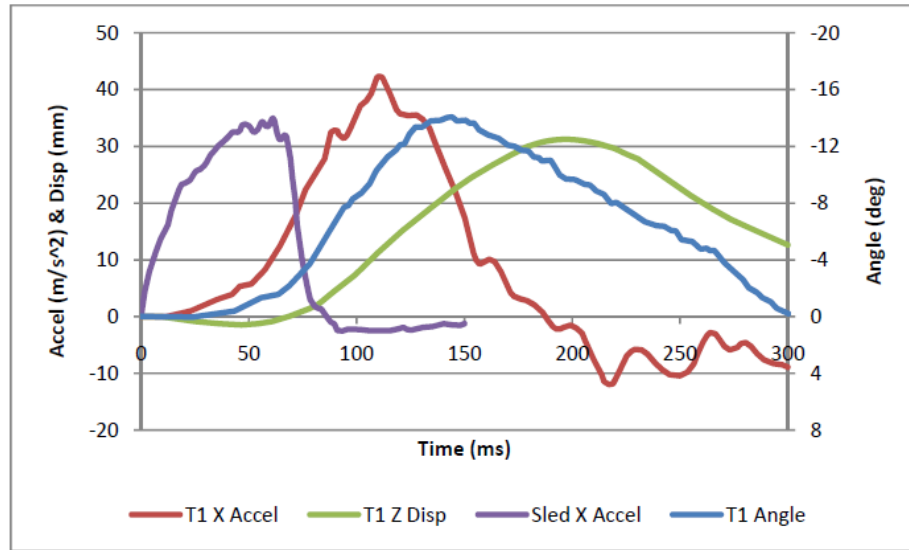


Figure 5-24: T1 inputs for the 4 g simulation (Davidsson et al., 1998).

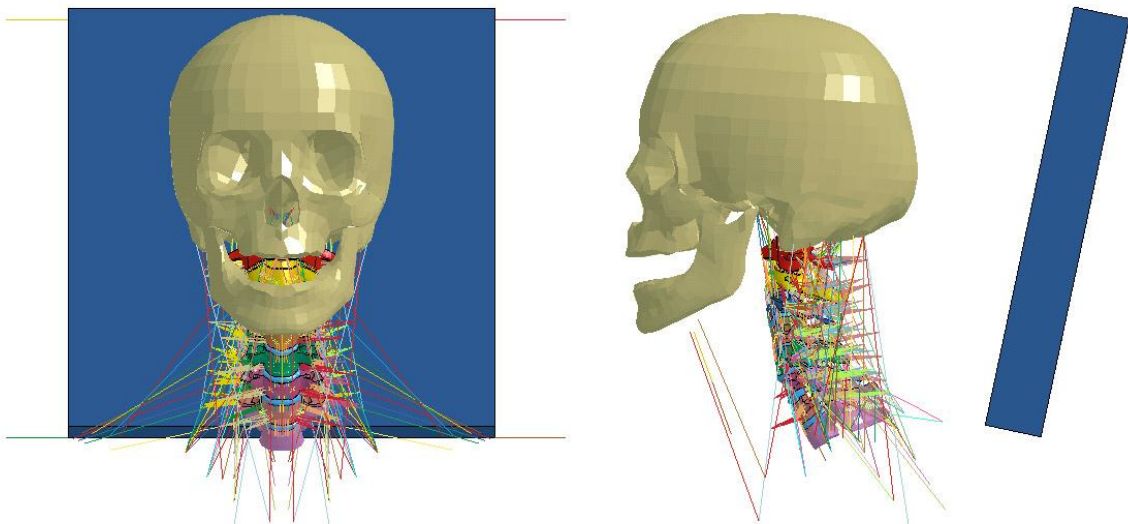


Figure 5-25: Model during the 4 g simulation

The upper and lower corridors of the volunteer response from the experimental data by Davidsson et al. (1998) were presented in the work by Hynd et al. (2007). The model generally followed the response very closely and was within the corridors (Figure 5-26 & Figure 5-27). In the anterior direction, the response of the skull relative to the T1 started to diverge at about 230 ms. In the

superior-inferior direction, the head displacement slightly diverged from the experimental data between 140-240 ms. The results which were obtained were similar to those from Fice (2010).

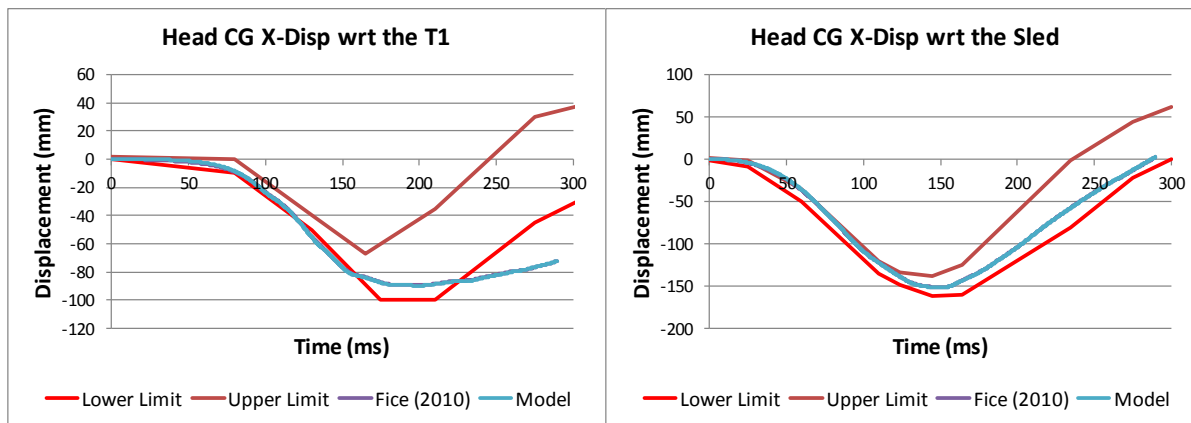


Figure 5-26: Head CG X displacement response to a 4 g rear impact

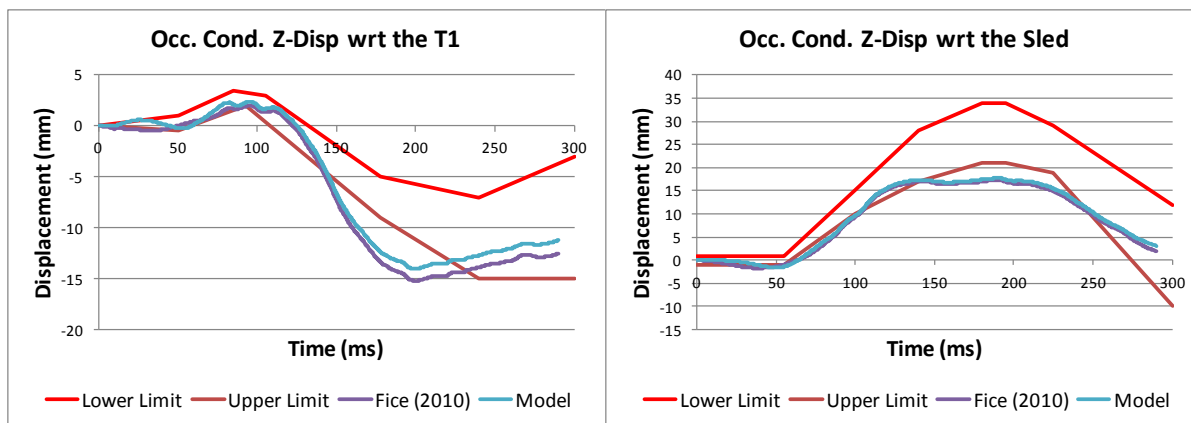


Figure 5-27: Head occipital condyle displacement response to a 4 g rear impact

5.4 Model Comparison to Simplified Out of Position Experiment

The behaviour of the cervical spine in out-of-position rear impacts was studied by Ivancic et al. (2006a). This study used six human whole cervical spine specimens (occiput-T1) with an average (Range) age of 80.2 (79-93) years. The setup for the study is shown in Figure 5-28. A surrogate head with the mass of 3.3 kg and horizontal, sagittal, and frontal plane moments of inertia of 0.014, 0.019, and 0.015 kg m² was rigidly attached to the occipital mount. A muscle force replication system with cables and springs was used to mimic the behaviour of passive muscle in the cervical spine. Two anterior, two posterior, and eight lateral cables were passed through wire loops

and pulleys, then connected to springs. The stiffness of springs was 4 N/mm in anterior and lateral directions, and 8 N/mm in the posterior direction. Spring preloads were also applied to the springs. The surrogate head was rotated with an average (SD) head-T1 rotation of 28.4° (5.3°) of left axial rotation, 3.5° (3.7°) of flexion, and 17.9° (4.7°) of left lateral bending. The head positioning device was disengaged with the initiation of the impact. Various impact levels were used in the study and sample acceleration data for an 8 g impact and the resulting head kinematics from one of the specimens were reported.

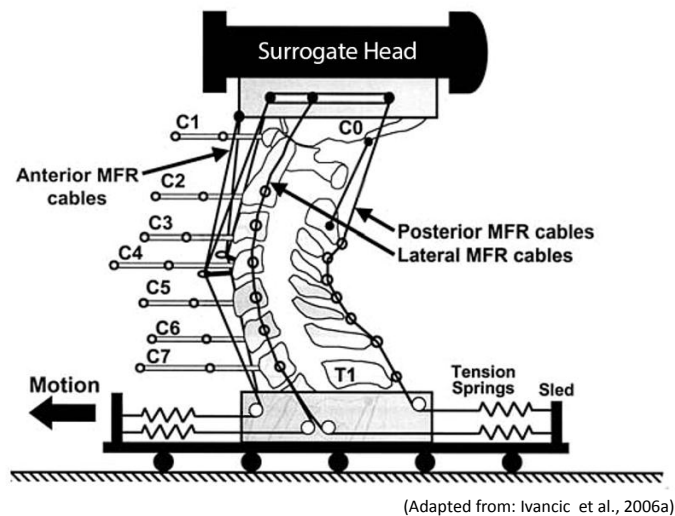


Figure 5-28: Ivancic et al. (2006a) experiment setup

The 8 g impact was simulated using the model in order to compare the results with the experimental data. The active muscle behaviour was turned off since the experiment did not use muscle activation. The mass and inertial properties of the head were changed to those of the study. Head rotation of 28.4° in left axial rotation, 3.5° in flexion, and 17.9° in left lateral bending were applied to the model. After the head was in the proper out-of-position posture, the reported acceleration for the specimen, shown in Figure 5-29, was applied to the T1. During the impact, the T1 was only allowed to move in the anterior direction and the rest of the components were free to move in any directions.

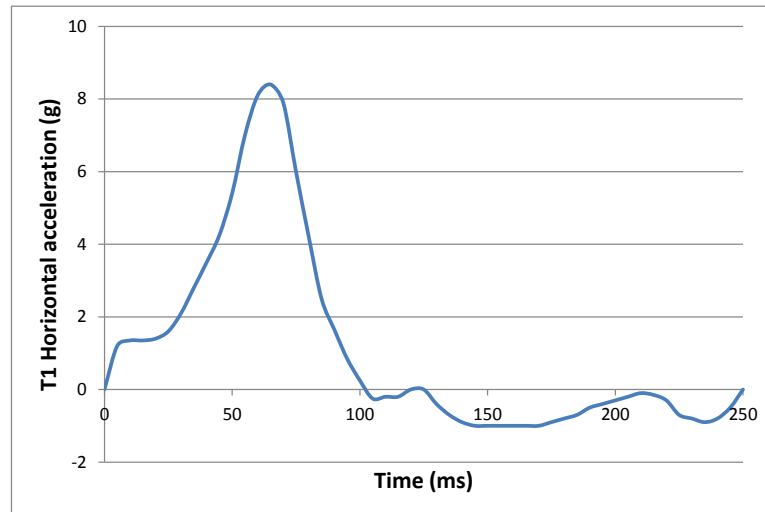


Figure 5-29: Applied acceleration to the T1 for 8 g rear impact

The head kinematics of the model was compared to data by Ivancic et al (2006a) (Figure 5-30). The positive values in this figure represent extension and right lateral and axial rotations. It must be noted that since the average and the range of the head kinematics were not reported, the comparison was made to one specific specimen; therefore there is a difference in initial rotation between the model and the reported experimental result in Figure 5-30. Overall, the model has very similar behaviour as compared to the experimental data in extension-flexion and axial rotation. In extension-flexion, the model moves into flexion earlier than the experimental data. In axial rotation, the experimental model reached the neutral position faster than the FE model. The peak axial rotations are very similar between the model and the experiment. In lateral rotation, the model moves in the direction of right axial rotation at the initiation of the impact. The experimental model goes through left lateral bending before the muscles in the right lateral side of the cervical spine start to pull the head back to the neutral position. The difference could be explained through the different muscle implementation of the Finite Element and the experimental model. In the numerical model, the origins of the lateral muscles could put higher bending moments on the head, therefore moving the head in the direction of neutral position at the initiation of the impact. There is limited information on the location of the cable attachments in the experiment and the head kinematics of other specimens and therefore it could only be concluded that the numerical model behaves similarly to the experimental data in extension and axial rotation.

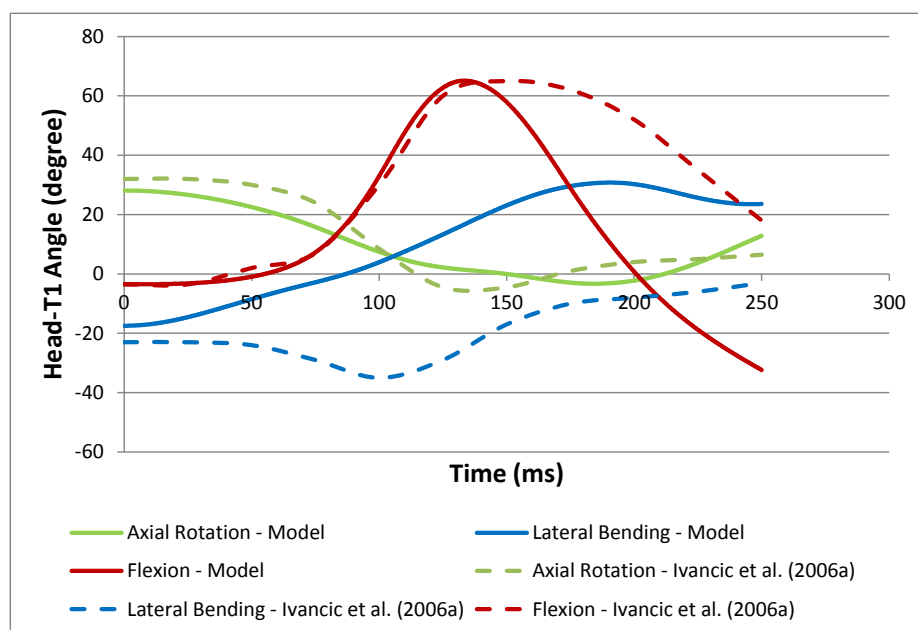


Figure 5-30: Head angles comparison with Ivancic et al. (2006a) experiment

Ivancic et al. (2006b) used the same setup and experimental methods to measure the intervertebral rotation of the neck during rear impact. The intervertebral rotation of the model was compared to the experimental results. Since the acceleration profile for the experiment was not provided in the literature, the same acceleration profile which was used for one of the specimens in Ivancic et al. (2006a) was used for the simulation (Figure 5-29). In extension and axial rotation, most of the results are within one standard deviation of the experimental data. In axial rotation, the shape of the rotation is in good agreement with the experimental data as the largest rotation is between the atlas and the axis. In flexion, C0-C1 and intervertebral rotation below the C5 vertebra are all above one standard deviation of the experimental data. In left lateral bending, the results for mid and lower cervical spine are outside one standard deviation of the experimental values. The validation of the model in physiological range of motion with respect to the whole cervical spine cadavers of Ivancic et al. (2006b) experiment suggests that the ligament properties are well defined for extension, flexion, lateral bending, and axial rotation. The main difference between the simulation and the experimental setup by Ivancic et al. (2006b) is the different implementation of the muscles. In the flexion mode of rotation, the head spent less time in extension in the simulated model. The movement of the head in flexion causes higher intervertebral rotations than the experimental data. In left lateral bending of the simulated model, the head quickly moves toward neutral posture and then moves into

right lateral bending; whereas, in the experiment, the head movement consists of further left lateral rotation. This explains the lower values of the left lateral bending between the vertebral bodies of the model as compared to the experiment. The difference in the implementation of the muscles makes the comparison challenging. The comparison does show that the two models behave very similarly in extension and axial rotation. There is limited information on the lateral bending of the neck during driving. Checking for pedestrians and traffic usually requires neck movements in the axial direction. Therefore axial rotation will be the focus of study in this thesis.

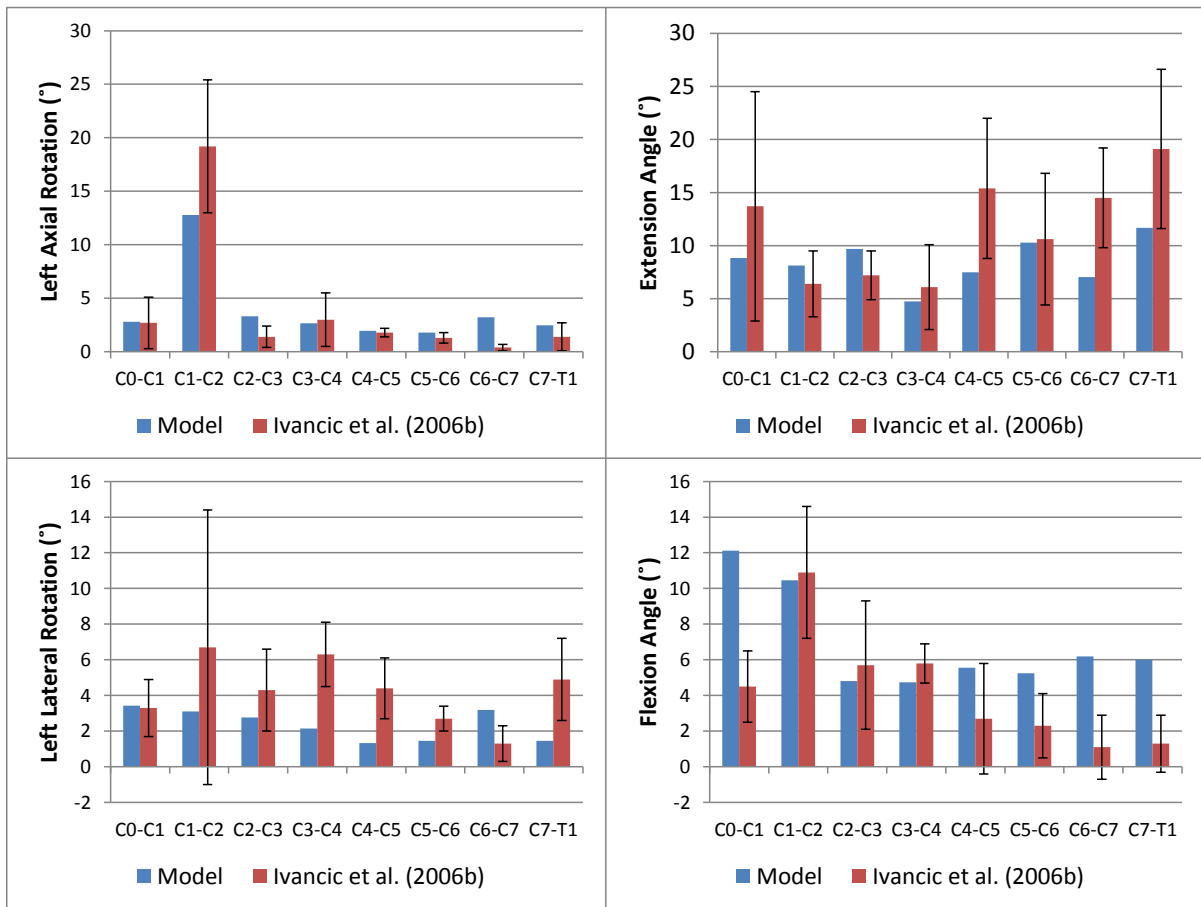


Figure 5-31: Intervertebral rotation comparison with Ivancic et al. (2006b) experiment

Chapter 6

Soft Tissue Strain Evaluation for Out-of-Position Impact

This chapter is focused on comparing the difference between neutral and out-of-position neck response and the potential for neck injury. The effect of severity of impact and rotation were two of the factors that were studied. Since active musculature has been shown to reduce impact severity in neutral position, its effect on out-of-position was also investigated. In the figures of this chapter, x acceleration is positive in the anterior direction, z acceleration is positive in the superior direction, angle is positive in extension, A represents active musculature, and P represents passive musculature.

6.1 Modeling Out-of-Position Impact

In this section, the out-of-position model will be described. Due to the availability of its impact data, the study by Deng (1999), mentioned earlier in Chapter 5, was selected. The simulation procedures followed were based on the study by Fice (2010). Three different impact levels of 7 g, 12 g, and 16 g were studied. Deng (1999) tested impact severities of up to 10 g and to simulate the 12 g and 16 g rear impacts, the x-acceleration of the 10 g experimental impact was scaled following Fice et al. (2011). The acceleration curves for the 7 g and 10 g impacts are outlined in Figure 5-16 and Figure 6-1.

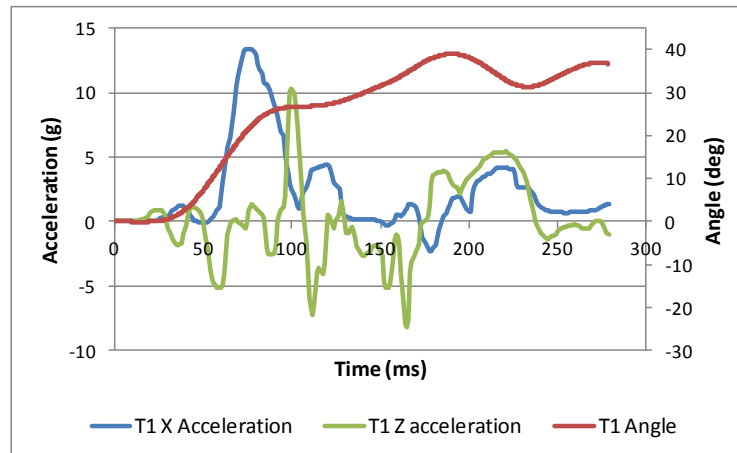


Figure 6-1: T1 prescribed motion for 10 g rear impact (Deng, 1999)

As discussed previously in Chapter 4, muscle activity could potentially reduce the chance of neck injury. Muscle activation in the model allows for better understanding of the cervical spine behaviour

in vehicle occupants. Both passive and active muscles were simulated and the resulting strains were used for the comparison process.

Axial rotation of the head is one of the factors that may affect the severity of neck injury. In theory, rotation causes initial strain on the ligaments and the addition of strain caused by impact could result in ligament injury in the cervical spine. In this section, three different head rotations outside the neutral position were studied. The selected rotations were based on an experiment discussed earlier by Shugg et al. (2011). The rotations of 24.5°, 42.5°, and 60.5° were selected for the study. These values are based on the average plus one standard deviation of rotation of the head to the right during average driving conditions. The values also cover a good range of head rotations and 60.5° is a good representation of the physiological range of motion of the cervical spine.

To simulate the out-of-position rear impact scenarios, the head was first rotated around the occipital condyle as shown in Figure 6-2. The head rotation was applied in the axial directions while the head was free to move in all other directions. After the rotation, an acceleration of 7 g, 12 g, or 16 g was applied to the T1 vertebra. A total of 24 simulations were performed to analyse the effect of rotation, impact, and muscle activations on the cervical spine (Table 6-2). A sample motion of the head through 7 g impact at a rotation of 60.5° is shown in Figure 6-3.

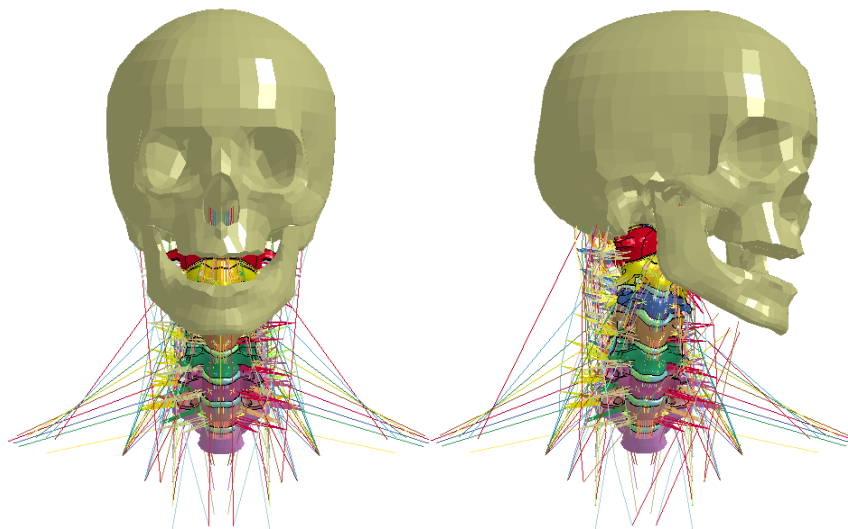


Figure 6-2: Application of 60.5° axial rotation on the cervical spine

Table 6-1: Study simulations

Axial Rotation	Simulations		
Neutral	7g-A	12g-A	16g-A
	7g-P	12g-P	16g-P
24.5°	7g-A	12g-A	16g-A
	7g-P	12g-P	16g-P
42.5°	7g-A	12g-A	16g-A
	7g-P	12g-P	16g-P
60.5°	7g-A	12g-A	16g-A
	7g-P	12g-P	16g-P

A - Active musculature
P - Passive musculature

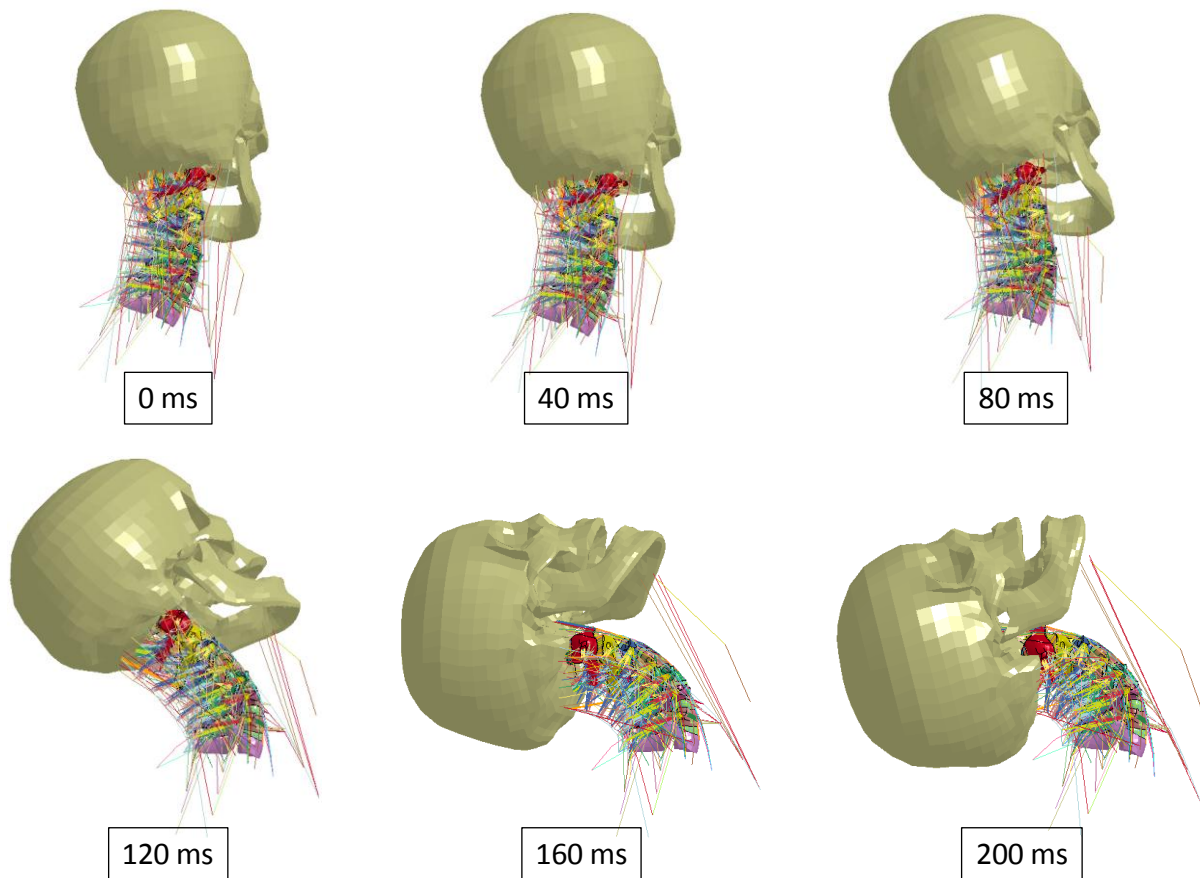


Figure 6-3: Cervical spine motion during 60.5° out-of-position impact (7 g)

6.2 Impact Severity

The severity of impact can increase the chance of injury on the cervical spine. Higher impact severity can contribute to more motion in the cervical spine leading to hyperextension and increases the likelihood of ligament injuries. The effect on ligament strains (Figure 6-4, Appendix A) shows how the risk of injury to the capsular ligaments increases as a result of impact severity in both neutral and out-of-position postures. The increase in ligament strains with respect to impact severity is seen in most of the ligaments of the cervical spine. There are some exceptions in the posterior ligaments of the lower cervical spine. The low strains in these regions are because the ligaments do not go through much extension during the impact. The anterior ligaments of the lower cervical spine go through higher extension with the increased motion of the cervical spine and therefore are affected by the severity of the impact. The increase in severity of impact puts many of the ligaments of the cervical spine at risk.

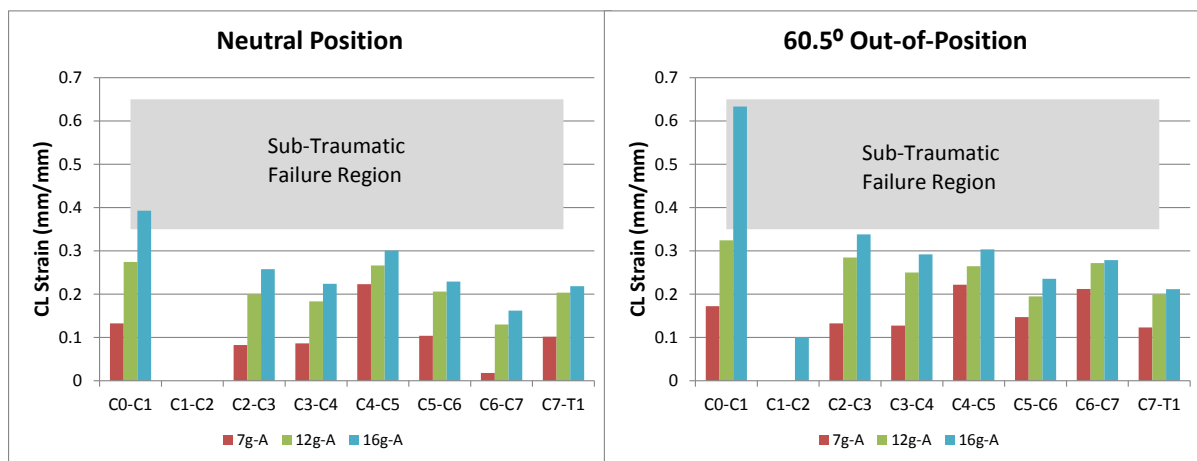


Figure 6-4: Effect of impact severity on capsular ligaments

Table 6-2 illustrates the impact levels that may cause excessive ligament strains in the cervical spine during simulations with active musculature. The simulations showed that there is no risk of excessive ligament strains in cervical spine during a 7 g neutral posture rear impacts. In out-of-position, there is a slight chance of injury in the C2-C3 Alar ligament during the 7 g rear impact; however, all other ligaments are below the failure corridors reported by the literature. During 12 g rear impact, the alar ligaments between C7-T1 and C2-C4, ligamentum flavum between C1-C2, and the apical ligament become in risk of injury. Chance of ligament injury is higher at 16 g rear impact. The ligaments, their location, and the impact levels which may cause injury with muscle activation in the model are

outlined in Table 6-2. Based on the results of the study, the C0-C4 region of the cervical spine are the most vulnerable to rear impact during neutral and out-of-position rear impacts.

Table 6-2: Impact levels that may cause injury to the cervical spine ligaments

Ligament	Cervical Spine Region							
	C0-C1	C1-C2	C2-C3	C3-C4	C4-C5	C5-C6	C6-C7	C7-T1
ALL		16g	7g*-16g	12g-16g			16g*	12g-16g
CL	16g							
ISL		16g						
LF		12g-16g	16g*					
PLL			16g*					
TL			x	x	x	x	x	x
TM			x	x	x	x	x	x
SC			x	x	x	x	x	x
IC			x	x	x	x	x	x
Apical	12g*-16g		x	x	x	x	x	x
Alar	16g*		x	x	x	x	x	x

Note: The results are based on the highest ligament strains within all simulations with active musculature

* Applies to Out-of-Position cases only

6.3 Active and Passive Musculature

Muscle activity is an important factor in impact analysis as it represents the biomechanic behaviour of live humans instead of cadaver models. The effect of active musculature was analyzed by running the model in both passive and active musculature modes, as discussed previously in Chapter 4. The results of the simulations are shown in Appendix A. Figure 6-5 illustrates the effect of muscle activation on the capsular ligaments. By providing more force to resist the motion of neck during whiplash scenarios, the active musculature helps to reduce the ligament strains in the neck. Based on the result of the simulations, the muscle activation has a large effect on protecting the ligaments in the upper cervical spine. Brodin et al. (2005) came to a similar conclusion when she used a cervical spine model to simulate frontal impacts.

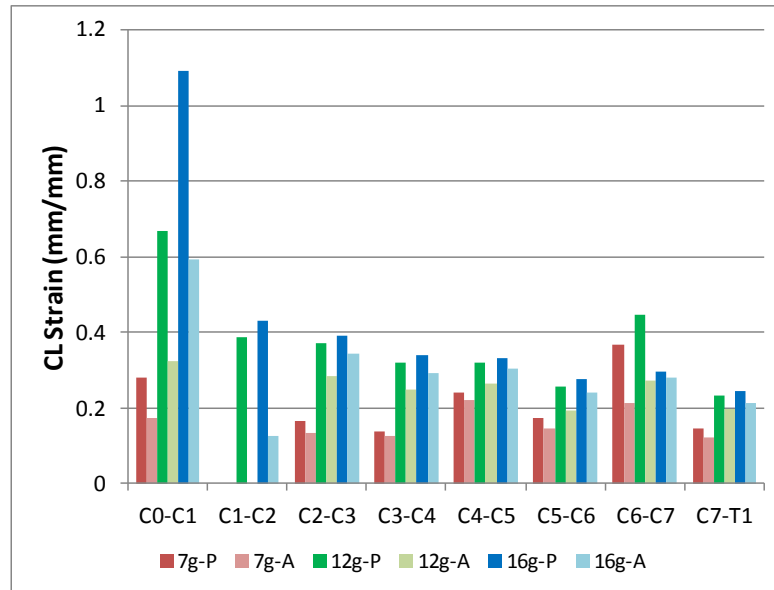


Figure 6-5: Effect of muscle activity on ligament strains of capsular ligament

The results obtained from the study of muscle activation are in good agreement with previous studies by Van der Horst (2002) and Fice (2010) who had observed reduction in ligament strains due to muscle activation in rear impact neutral position models. The simulation showed reduction of ligament strains in both neutral and out-of-position postures. Based on the results, muscle activity provides good support for the cervical spine and helps to reduce ligament injuries on the neck. Further development of procedures to test cadaver models with muscle activity could improve knowledge on the biomechanics of the cervical spine.

The result of the simulations may suggest that the chance of injury in the cervical spine is higher for people who have muscle activation dysfunctions. Inability for the muscles to activate on time can result in higher strains in the cervical spine ligaments. The model shows that the increase is often larger in the upper cervical spine, especially the capsular ligaments. Since facet joints are the most suggested source of pain in the neck, any dysfunctions in muscle activation patterns may result in chronic pain in the cervical spine.

6.4 Rotation Severity

The ligament strains in the neck may increase due to the initial strain resulting from the rotation of the neck. After the initial strain, the impact may cause additional strain in the ligaments, thus causing injury. In axial rotation, most of the relative rotation of the neck is between the atlas and the axis

vertebrae. Therefore the increase in potential of injury should be seen in the upper cervical spine ligaments. In rear impact the posterior ligaments may not see an increase in strains. This is because the primary motion of the head in rear impact is extension and the posterior ligaments do not experience much strain during this mode of motion.

Three different rotations were simulated to investigate the effect of low, medium, and high axial rotation on the cervical spine. Complete results from the simulations are available in Appendix A. Figure 6-6 illustrates the effect of rotation on the ligaments during 16 g rear impact with active musculature. In general, axial rotation can result in higher strains in some of the ligaments of the cervical spine, therefore, increasing the potential for injury.

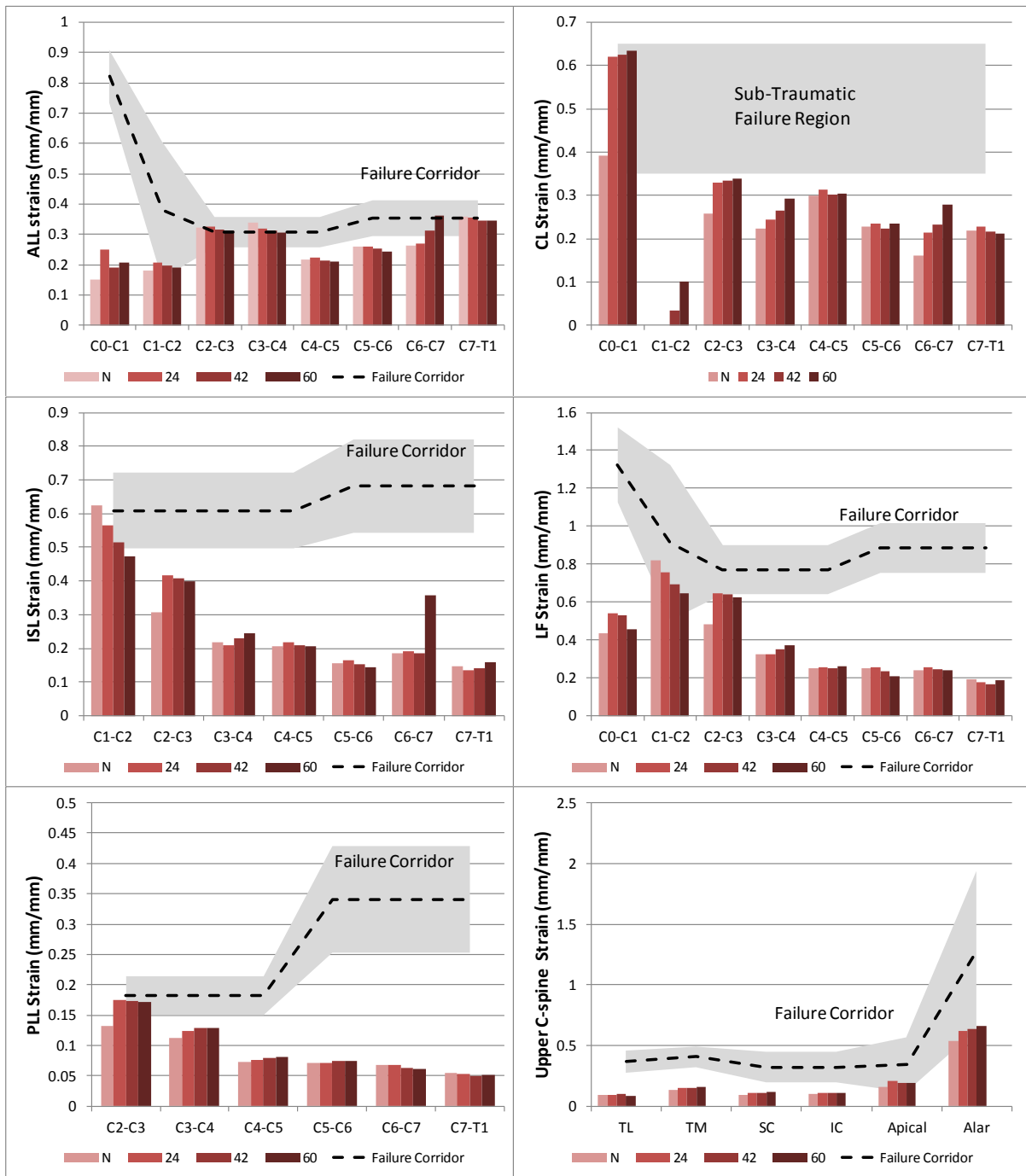


Figure 6-6: Effect of rotation on ligament strains during 16 g rear impact with active musculature

The anterior longitudinal ligaments are responsible for controlling the extension of the neck and therefore are important in rear impact collisions. The ALL between C0-C2 and C6-C7 experience

higher strains in out-of-position and push the strains to the failure regions between C1-C2 and C6-C7. The severity of rotation is not as apparent in the upper cervical spine region but can be seen between the ALL of C6-C7. In this region, the strain increases almost linearly with the increase in rotation.

As discussed previously, the distraction of facet joints is the most supported mechanism for whiplash pain in impact scenarios. Therefore the capsular ligament strains compared to literature can provide a good understanding of which facet joint is most likely to be affected by the rotation of the head during rear impact collisions. The results of this study showed that axial rotation of the neck results in higher strains in the capsular ligaments between C0-C4 and C6-C7 regions. The C0-C1 capsular ligament experienced a very high increase in ligament strain during out-of-position scenarios. The C1-C2 capsular ligament did not have any strains during neutral and low axial rotation positions due to its initial laxity. Lord et al. (1996) who had analyzed the middle and lower cervical spine of whiplash patients had found the lowest joint damage in the C4-C5 facet joints; this could be because this region is not highly affected by the rotation in the cervical spine. Capsular ligaments in the regions of C1-C2, C3-C4, and C6-C7 experience an increase of strain with the addition of rotation relative to the rest of the cervical spine. Overall, axial rotation of the neck increases the chance of injury to the upper cervical spine and very high rotations may cause injury in the C3-C4 or C6-C7 regions of the neck.

Interspinous ligaments are located on the posterior side of the cervical spine and contribute to limiting the flexion motion of the neck. This ligament is within the failure corridors in the region of C1-C2 at 16 g rear impact. As rotation of the neck increases, the strain in this region is reduced. The C2-C3 region of the neck experiences higher strains in out-of-position; however, the strains at high rear impacts are not high enough to reach the failure corridors. The C6-C7 experiences very high ISL strain in 60.5° relative to all other rotation angles. This can be explained through the relative motion of C6-C7 vertebrae during the extension motion (Figure 6-7). The spinous process of C6 vertebra (purple) does not keep contact with the C7 vertebra and experiences very high extension as it slides beyond a point where its motion could be limited by the C7 vertebra during the impact. The strain in this rotation for the C6-C7 region does not reach the failure corridors. The axial rotation of the neck does not affect the ISL strains in the C3-C6 and C7-T1 regions of the spine.

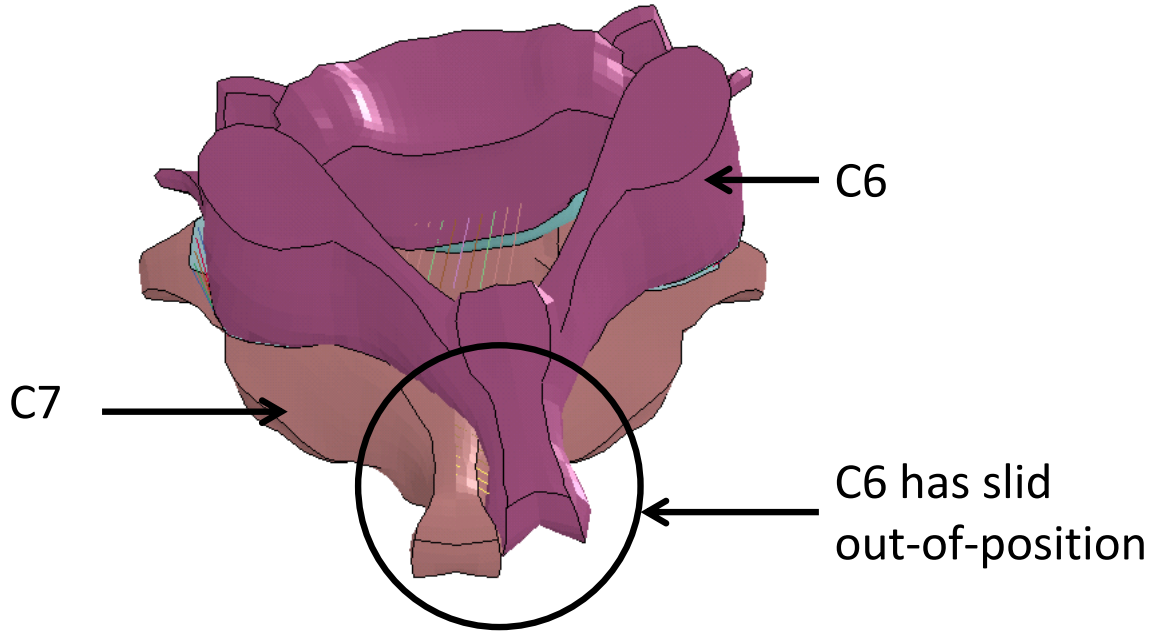


Figure 6-7: C6-C7 relative rotation during extension in 16 g rear impact

The severity of rotation on strains of the ligamentum flavum and posterior longitudinal ligament was very similar. In both ligaments, in the region of C2-C3, the out-of-position impact increased the strains and in the case of the 16 g rear impact, the strains reached the failure corridors. In the C3-C4 region an increase in axial rotation slightly increases the strains in the ligaments. In all other regions of the middle and lower cervical spine, the severity of rotation has minimum or no effect on peak strain values of the LF and PLL. In the upper cervical spine region of the ligamentum flavum, out-of-position strains are higher between C0-C1 and lower between C1-C2. The strains of LF during 12-16 g rear impact in the C1-C2 region can lead to injury in the ligament.

The superior crus, tectorial membrane, apical, and alar ligaments of the upper cervical spine were affected by the severity of rotation. The tectorial membrane and the superior crus experienced a slight increase in the peak strains, but not enough to raise them into the failure corridors. Axial rotation may cause the apical ligament to be injured in severities of 12-16 g rear impact. In the alar ligaments, the increase of axial rotation pushed the strains during 16 g rear impact to the failure corridors. The increase in the strains for the alar ligaments is in good agreement with the MRI study by Kaale et al. (2005). In this study it was reported that there is a higher chance of injury of the alar and transverse ligaments in out-of-position collisions. The results from this study did not show any evidence of increase in transverse ligament strains with respect to axial rotation.

Initial axial rotation of the cervical spine can increase the chance of injury in the neck. The increase in axial rotation moves many of the ligament strains into the injury corridors following impact. The highest increase in strain was seen in the capsular ligaments of the upper cervical spine. The higher strain in the alar ligament due to rotation was in good agreement with the literature (Kaale et al., 2005). There was also evidence to suggest that transverse ligaments may have higher potential of injury in a non-neutral position. The results of this study did not show an increase in the transverse ligament due to axial rotations. Increase in potential for injury in the transverse ligaments may be due to initial lateral, flexion, extension, or combination of different rotation modes on the neck.

In the experiment that was conducted by Panjabi et al. (2006b), it was shown that in 8 g out-of-position rear impact which was conducted on a cadaver with replicated passive muscles, the C0-C1, C3-C6, and C7-T1 regions may experience increase in flexibility parameters of neutral zone and range of motion in axial rotation with respect to the 2 g baseline impact. In the 7-12 g numerical model out-of-position rear impact simulations performed with passive musculature, it was seen that there is a possibility of injury in the ALL ligaments of C2-C4 and C7-T1 regions. The capsular ligament strains from non-neutral posture rear impact simulations in most regions of the neck were also very close or inside the sub-traumatic failure region. Injury in these ligaments may cause the increase flexibility of neutral zone or range of motion that was observed in the study by Panjabi et al. (2006b).

In summary, out-of-position posture could lead to a higher chance of injury in rear impact collisions. The severity of impact increased the potential for injury by increasing ligament strains in the cervical spine. Active musculature protected ligaments against high strains and therefore reduced the chance of injury. The severity of axial rotation on the neck affected some of the ligaments in the cervical spine but did not follow the same pattern in all cases. The highest increases in strains from out-of-position rear impact collision were seen in the C0-C1 capsular ligaments. The increase in strain caused by out-of-position posture could cause injury to the ligaments of the neck. The results were in good agreement with the experiments by Sturzenegger et al. (2005) who concluded that head posture had a significant effect on the persistence of neck injury symptoms.

Chapter 7

Summary and Recommendations

7.1 Summary

Whiplash injuries are common automotive injuries that may cause both acute and chronic neck pain. Whiplash-Associated Disorders are high in cost and may cause long term quality of life issues, with symptoms persisting from one to two years following exposure. There is some evidence that out-of-position head posture significantly affects the persistence of symptoms on victims. The purpose of this thesis was to study the effect of out-of-position neck posture in rear impact automotive collisions.

There are various methodologies that are used to evaluate cervical spine injuries. Crash tests using crash dummies along with the N_{ij} criterion are currently the most common method used in industry. The N_{ij} criterion considers the combination of tensile forces and extension-flexion moments of the neck to evaluate injury. Unfortunately, N_{ij} criterion and most other neck injury criteria were not derived for out-of-position whiplash scenarios. Detailed Finite Element Analysis was selected for injury evaluation because it provides a biofidelic model that can be validated by the literature and be used to find the location of injury based on various loading cases.

The cervical spine model that was used for this study was previously developed at the University of Waterloo. This model was previously validated in segment level in flexion, extension, axial rotation, and tension. The full cervical spine was also validated in frontal and rear impacts in head kinematics and tissue levels. Due to the findings from previous work on the model, changes to ligament properties of the upper cervical spine and muscle implementation were applied to the model, including laxities to the upper cervical spine to improve the biofidelity of the model. Because of the changes, the model had to be reevaluated in different modes of loading in order to be used in out-of-position rear impact scenarios.

A detailed study which included validation of the upper cervical spine in segment level and the full cervical spine in physiological range of motion was used to improve the laxities of the upper cervical spine. To start evaluating the laxities, a study was conducted to find out which ligaments have an effect on the forces and moments of the upper cervical spine in tension, flexion, extension, and axial rotation. Laxities were added to individual ligaments and the change in resulting forces and

deformation to failure were studied. This study provided a starting point to how laxities may be used to help achieve a more biofidelic model.

To find the appropriate laxity for the upper cervical spine ligaments, the model was validated at both the segment and full cervical spine levels. At the segment level, the failure force and displacement were considered in extension, flexion, axial rotation, and tension. In full cervical spine model, the model was validated against physiological range of motion in extension, flexion, axial rotation, and lateral bending. An iterative process was performed and laxities were altered until the parameters studied in the model were in agreement with the literature at both segment and full cervical spine levels.

Axial rotation was the most important motion of the neck for out-of-position impact evaluation since it is the most common mode of rotation during driving. Therefore the axial rotation at the segment level was first to be studied. Laxities were added to the model to assess forces and displacements at failure and physiological range of motion. To optimize the biofidelity in the model in axial rotation, the relative motion of the C1-C2 was also considered since the atlas and the axis carry most of the head motion during axial rotation. After satisfying the axial results reported by the literature, the laxities were tested in extension, flexion, and tension. The failure forces and the displacements to failure were used to evaluate the biofidelity in these loading cases as they are important parameters when evaluating the failure data during rear impact. The laxities were altered until the simulation results satisfied the reported values from the literature. Once the laxities were selected, the model had to be reevaluated in rear impact before further studies.

Similar studies that were performed previously to validate the model in rear impact were selected because of the availability of data. The model was first validated for global kinematics against 7 g rear impact. The head motion, relative vertebrae rotations, and the head CG acceleration in the anterior and superior direction were in good agreement with the literature. The new muscle implementation and the laxities that were introduced to the ligaments had increased the acceleration levels of the head in the superior direction and the relative flexion motion of the C1-C2 vertebra. These changes were due to the additional motion that is allowed in the upper cervical spine due to the new muscle implementation and the laxities in the posterior longitudinal ligaments.

The next step was to validate the model at the tissue level. The model was validated against 8 g rear impact on a cadaver model with muscle force replication. The mass of the head was altered to the

experimental data and the loading was applied to the T1 to simulate the experiment. Most of the strains in the anterior longitudinal ligament and capsular ligaments were within one standard deviation from the literature. The disk shear strains were also compared to those of the literature and the results were generally within one standard deviation of the literature with the exception of the C5-C6 region. The variation between the model and the experiment could be explained through the difference between the muscle implementation of the two models.

A volunteer based experiment was used to validate the active musculature of the model. A non-linear viscoelastic model of headrest was used for the simulation. Active musculature was used to mimic the neck behaviour of volunteer human subjects during impact. An acceleration of the T1 and headrest were applied to the model and the head's kinematics was observed. Generally the shape of the kinematics was in good agreement with the literature. Overall, the model showed good agreement with the literature in neutral position impact scenarios.

The model was also compared to an experiment which had tested the behaviour of the cervical spine in out-of-position impact. The mass and inertia of the head were changed to those of the experiment and passive muscles were used for the comparison. The head was rotated in flexion, lateral, and axial rotations and an 8 g rear impact pulse was applied to the T1. The head kinematics and relative vertebral rotations were in good agreement with the literature in axial rotation and extension. The differences in flexion and lateral bending may be caused by the different muscle implementation between the two models. Based on the comparison, the model in out-of-position behaves well in extension and axial rotation which are the primary modes of motion during out-of-position rear impact scenarios.

The model was used to study whether out-of-position impacts change the potential of injury. Three parameters of impact severity, muscle activation, and rotation severity were studied. Three different impact severities of 7, 12, and 16 g were used to find the effect of impact severity on the neck. Active and passive muscles were both simulated to see whether or not active musculature can protect the neck against injury. Finally three axial rotations of 24.5°, 42.5°, and 60.5° were studied along with neutral head posture in order to study the effect of non-neutral head postures during impact.

The increase in impact severity proved to be very important in all the rear impact simulations. The strains in most of the ligaments increased with the increase of impact severity. Some of the posterior ligaments in the lower cervical spine did not experience increase in strains since they did not

contribute much in the extension motion of the neck. With the presence of muscle activation, none of the ligaments of the cervical spine were in the injurious zone during the 7 g rear impact. At 12 g, the anterior longitudinal ligaments and the ligamentum flavum had a possibility of injury. During 16 g rear impact all the ligaments were at higher risk of injury.

Muscle activation reduced the risk of ligament injury of the cervical spine. Active muscles provide additional force to resist motion of the neck, therefore reducing the strains on the ligaments. The reduction in strain was visible during impact in both neutral and out-of-position postures. The results of this study suggest that people that may have muscle activation dysfunctions may suffer from ligament injuries at lower impacts. The strains were generally higher with passive musculature and therefore active musculature decreases the potential for ligament injuries during rear impact collisions.

It was concluded that out-of-position postures can lead to a higher potential of ligament injuries during rear impact collisions. In general, capsular ligaments experienced the highest increase in strains with axial rotation. The increase in strains which is caused by non-neutral postures during impact was enough in some ligaments to push them into the failure corridor. In higher impact levels, the alar, apical, and capsular ligaments of the upper cervical region are at higher risk of injury. In the middle cervical spine region, the ligamentum flavum and the posterior longitudinal ligament have potential for failure due to the axial rotation of the head. In lower cervical, the anterior longitudinal ligaments may be injured due to a non-neutral posture. The results suggest that initial rotation of the head before the impact may cause more severe injuries in the neck.

7.2 Recommendations

The validation of the musculature in out-of-position may improve the biofidelity of the model. The validation of the model in physiological range of motion was completed without the presence of passive muscles. In the future, with the availability of more data, it will be useful to validate the musculature of the cervical spine in physiological range of motion and impact cases. Neck rotation in human requires the activation of certain muscles. The initial activation may cause different relative rotations between the vertebrae or different muscle behaviour during impacts. At the present time, it is very difficult to model head rotation via muscle activation. Instead the rotation was achieved by enforcing the head to rotate about a certain axis. Alteration of this methodology and activating the muscles in order to rotate the head may improve the impact results.

Implementing the ligament properties reported by Mattucci (2011) to the middle and lower cervical spine may improve the quality of the model. The ligament properties of the upper, middle, and lower cervical spine model are from different experiments and various authors in the literature. Changing the properties to match the study by Mattucci will make sure the experiment setup and the age of specimens were consistent for all the cervical regions. Implementation of these properties will require validation of these two regions against the literature.

The model has not currently been validated with experimental data in out-of-position rear impacts. The difficulty to model the muscle force system from the literature was due to the lack of information provided and also the instabilities that were caused from attempting to model cables that slide inside wire loops. With the advancement of software technology it will be helpful to model the muscle replication system of the experiment. Validating the muscle replication system that was used in the literature cannot provide much certainty on the proper muscle behaviour in out-of-position. However, it does help validate the interaction of ligaments, discs, and vertebral bodies in out-of-position rear impact.

The laxities of the upper cervical spine were evaluated based on an optimization process that only considered laxity increments of 0.5 mm. This decision was adapted to reduce the time required for the validation process. An optimization process which is performed by LS-OPT software can improve the ligament properties of the upper cervical spine. LS-OPT software enables the user to use the data from the literature to optimize the laxities in much finer detail. With improvements on processing speeds, this will become easier and the optimization may improve the biofidelity of the model.

The out-of-position rear impact simulations were performed on external servers with a running simulation limitation time of seven days. To simulate the head rotation properly, it would be useful to rotate the head for a long period of time and hold it in place for all the forces on the neck to settle before the application of the impact. Due to the complexity of the model and the time limitation on the servers, the head was only rotated for about one second before the application of the impact. With the advancement in processing speeds in the future, the model can be simulated with a higher rotation period to see whether there is any sensitivity on the final results.

Vertebral arteries are one of the regions that can be damaged due to out-of-position posture. There is little evidence to support this idea; however, modeling the vertebral arteries can help to understand

whether or not the hypothesis is true. Once the vertebral arteries are modeled, the strain in the arteries could be used to find out whether there is any potential of injury in this region of the neck.

The effect of lateral bending, flexion, extension, and their combination with axial rotation could also be investigated. This thesis only focused on the effect of axial rotation. The main reason was the availability of data on the average and peak rotations of the head during driving. During axial rotation, the head was allowed to rotate freely and therefore there was a small amount of lateral bending and extension-flexion on the neck. However, studying more severe rotation cases in lateral, flexion, and extension situations can help future engineers to develop better seat designs that can protect vehicle occupants.

The study of the effect of out-of-position injuries in frontal and side impacts can also be very helpful. The posterior ligaments are more vulnerable in frontal impact. The initial strains due to the rotation of the head, especially in the upper cervical spine region, may amplify the resulting impact strains and cause failure in these ligaments. The capsular ligaments are also vulnerable in side impacts because of their location. Therefore the initial strain can cause injury to the facet joints and cause chronic pain. The combination of these impact cases may also be useful for designing more effective headrests.

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Appendix A

Rear Impact Results

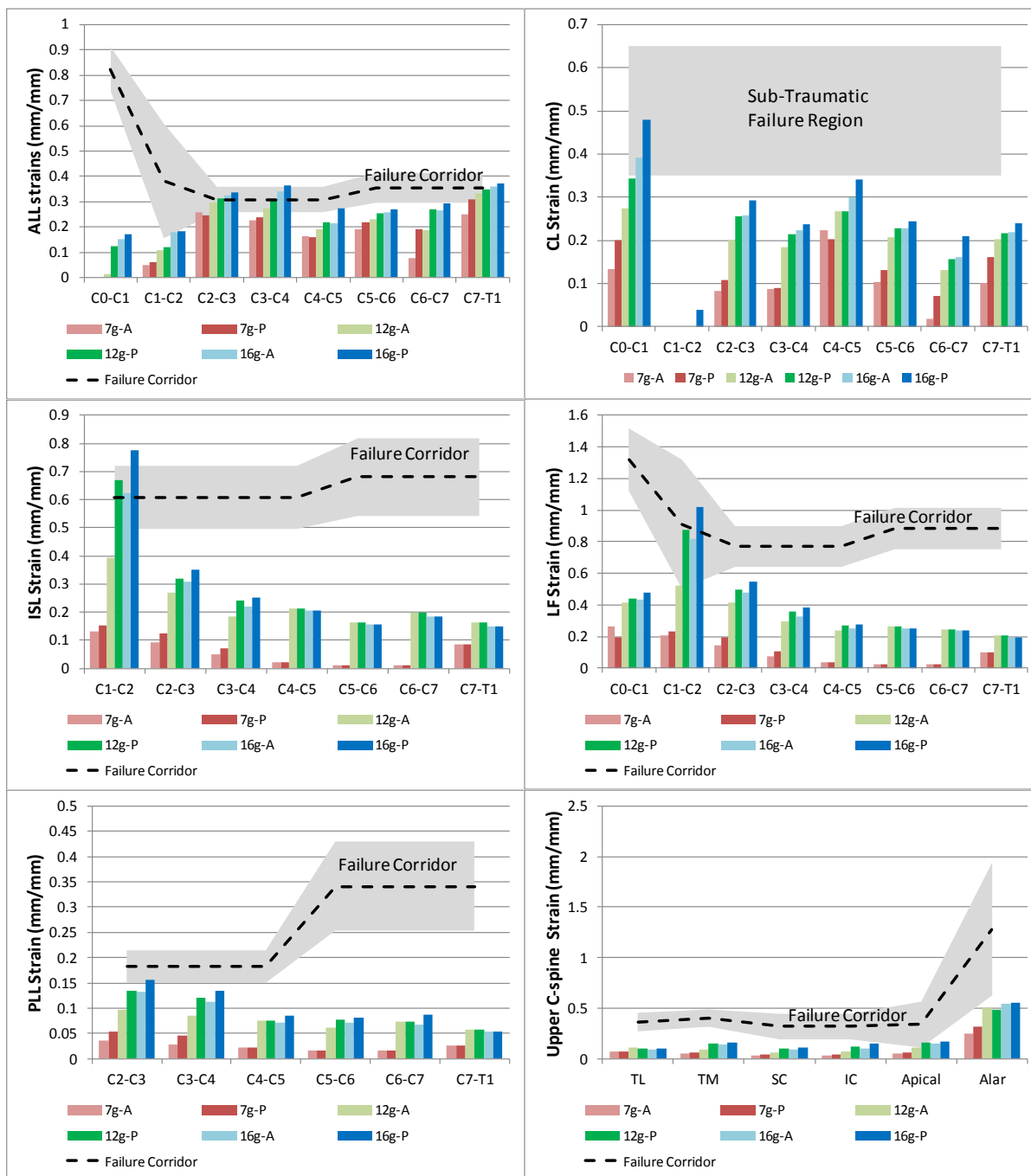


Figure A1: Results for neutral position rear impacts (A: Active and P: Passive musculature)

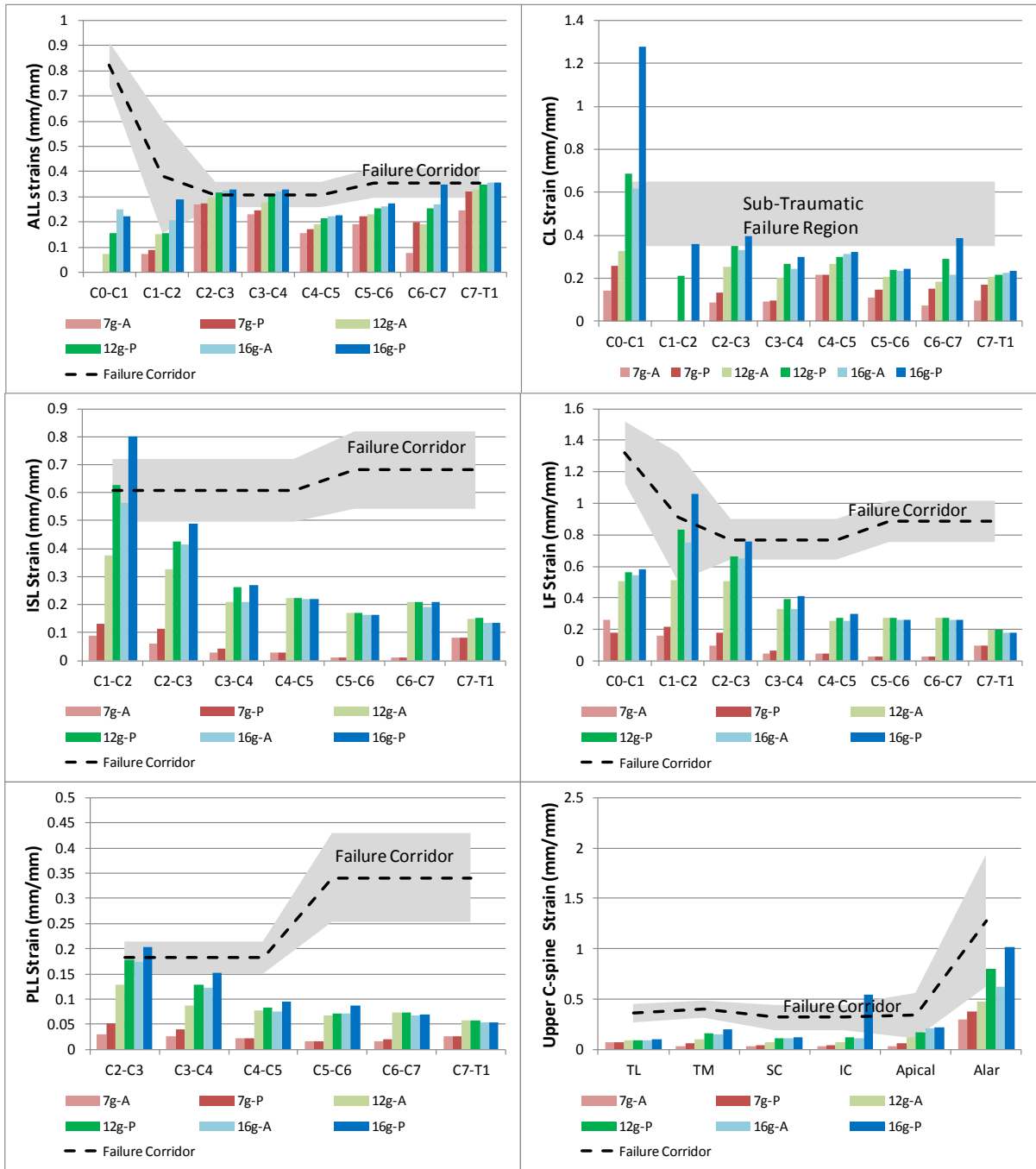


Figure A2: Results for 24.5° out-of-position rear impacts (A: Active and P: Passive musculature)

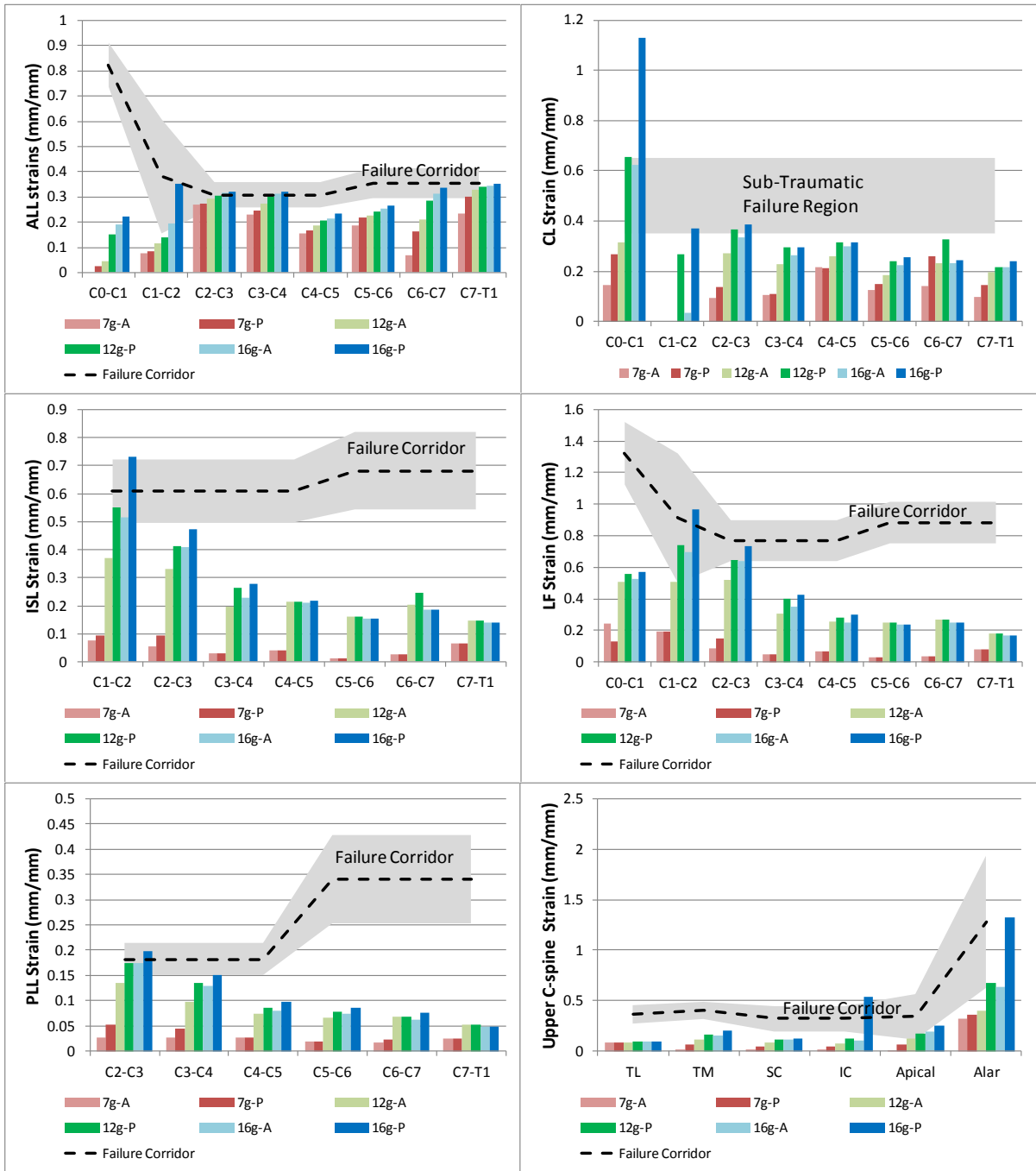


Figure A3: Results for 42.5° out-of-position rear impacts (A: Active and P: Passive musculature)

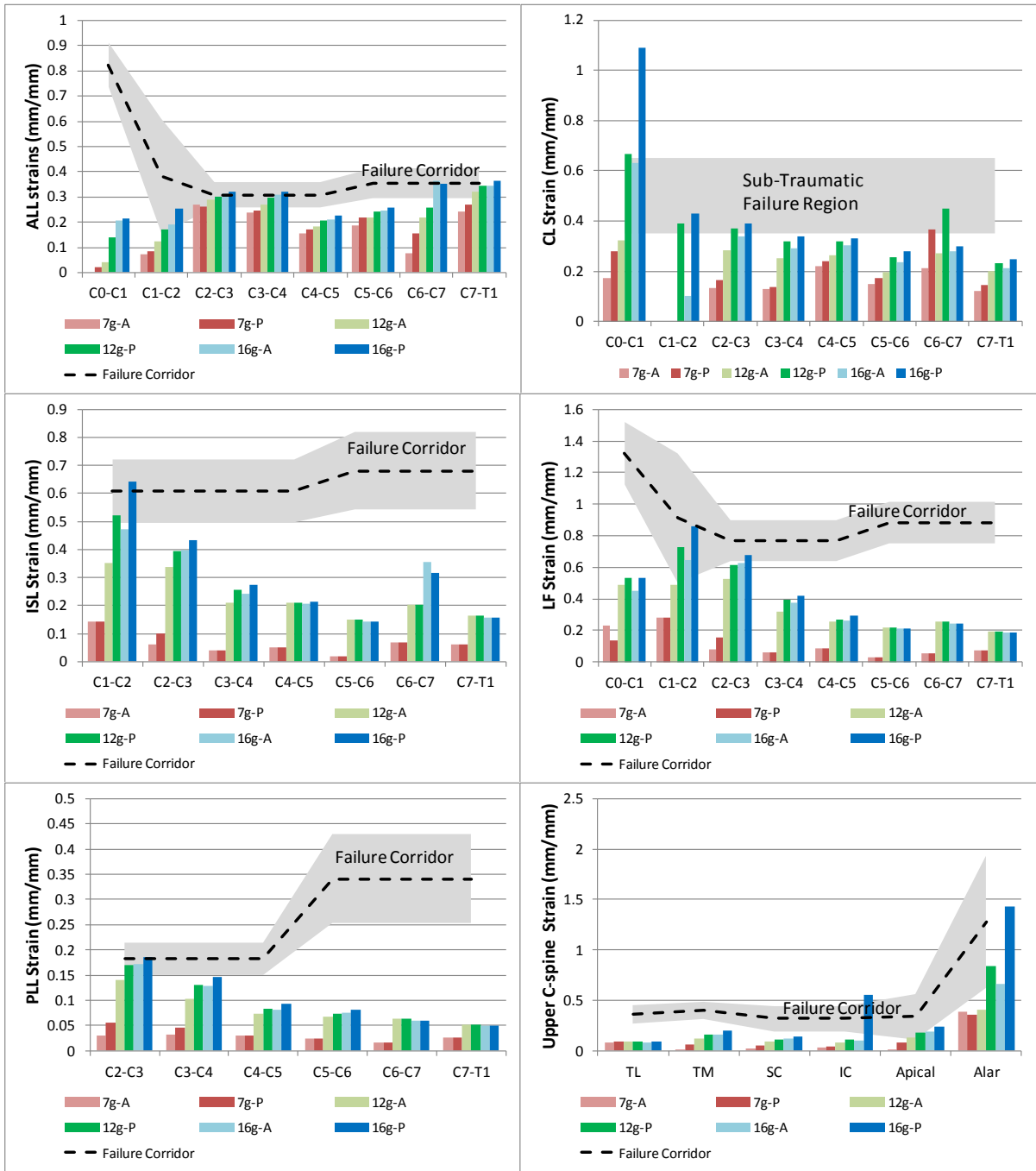


Figure A4: Results for 60.5° out-of-position rear impacts (A: Active and P: Passive musculature)