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entitled

Biomechanical Effects of Initial Occupant Seated Posture During Rear End Impact Injury

by

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Submitted to the Graduate Faculty as partial fulfillment of the requirements for the

Doctor of Philosophy Degree in Biomedical Engineering

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An Abstract of

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Whiplash injuries remain the most common consequence of motor vehicle
accidents and account for a significant annual cost to society. In United States alone,
whiplash injuries occur in more than one million people every year. It has been reported
that the annual cost for whiplash injuries in the USA is $2.7 billion and £2.5 billion in
the UK. Whiplash injury can reduce victim’s quality of life for a significant amount of
time, ranging from weeks to few months and sometimes, symptoms may persist for years.
Whiplash is a soft tissue injury and the spectrum of injury may include damage to
intervertebral ligaments, discs, facet joints and surrounding soft tissues. Whiplash is the
most common injury sustained in the rear-end motor vehicle collisions accounting for
more than one-thirds to half of the total reported whiplash cases.

As an automobile occupant is hit from behind, the forces from the seat back
compress the kyphosis of the thoracic spine, which provides an axial load on spine
together with shear force which makes it easier for soft tissue injuries to occur. An out of
occupant seating posture can change this complex loading compared to the normal
posture. Based on the clinical and epidemiological evidence, the axial rotation of the head
or inclined head and the body forward postures prior to impact had an increasing
whiplash injury risk following automotive rear impacts. There is very less understanding of the injury patterns when the vehicle occupant seating posture is out of position (non-typical posture) and is subjected to a rear end collision.

A detailed 3D osseoligamentous human head-pelvis FE model has been developed. The model consists of a skull, vertebrae, viscoelastic discs, frictionless facet joints, nonlinear ligaments, ribcage and pelvis. All of the material properties were derived from literature. The bony structures were modeled as linear elastic material models and the discs and ligaments were modeled as non-linear models. The main aim of this thesis was to develop a head-pelvis FE model to study the influence of different out of occupant seated postures on the injury pattern due to rear end impact loading. The initial goal was to validate the model under static and dynamic loading conditions. Static validation was done in flexion/extension, lateral bending, and axial rotation by comparing the results with cadaveric experimental data. Dynamic validation was performed by providing an input sled acceleration to the FE model and comparing the global as well as intervertebral motion responses with the experimental corridors.

FE model was first tested in the normal seating posture in the standard seat with headrest due to the rear end impact of 4g. The headrest and the seat were modeled as a hyperfoam material model in ABAQUS. The normal seating posture was then modified to create different out of occupant seating postures: body forward (BF-body leaned forward by certain angle measured in degrees with respect to the vertical); head turned (HT-head rotated to the right by certain angle measured in degrees), combination of the body forward and head turned posture (BF-HT) and head flexed (HF-head flexed by 20 degrees) that were derived from the literature. The first part of the experiment was to
study the effect of progressively increasing body forward seating postures (BF-15, BF-20 & BF-25) on the injury pattern due to the same rear end impact. The next test was to investigate the effect of progressively increasing head turned (HT) seating postures (HT-15, HT-30 & HT-45) on the whiplash injury outcome. Furthermore, head turned, body forward, head flexed and BF-HT postures were compared with the normal seating posture to identify high risk out of occupant postures. The final test investigated the effect of increased impact severity of 10g on the identified high risk out of occupant postures. The risk of whiplash injury was quantified by measuring intra-discal pressures, facet stresses and ligament strains.

Results showed a non-linear increase in the range of motion, intervertebral disc pressures, facet stresses and ligament strains with the progressive increase in body forward and the head turned postures. Intervertebral discs had increased pressures with all the out of occupant seating postures when compared to the physiological values indicating high risk of disc injury. Results indicated that body forward posture increased the risk of capsular ligament injury and head turned posture increased the risk of capsular and alar ligament injury which agrees with the clinical literature. Body forward head turned posture (BF-HT) had increased risk of both alar as well as capsular ligament injuries.
I would like to dedicate this work to my parents Palepu Venkata Subrahmanyeswara Rao and Palepu Padmavathy.
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List of Abbreviations

ALL..........................Anterior Longitudinal Ligament
CL .............................Capsular Ligament
FE ..............................Finite Element
FEA .........................Finite Element Analysis
HR ..............................Head Restraint
IDP .............................Intradiscal Pressure
PMHS .........................Postmortem Human Subjects
ROM .............................Range of Motion
List of Symbols

Km......Kilometer
h.........Hour
msec ...Millisecond
mm .....Milimeter
N.........Newton
MPa....Mega Pascal
Chapter 1

Introduction

1.1 Rationale for Research

A total of 1.8 million people in the United States suffer from chronic pain and disability following motor vehicle accidents (MVAs) each year [1]. Whiplash injuries remain the most common consequence of motor vehicle accidents and account for a significant annual cost to society. In United States alone, whiplash injuries occur in more than one million people every year [2]. Mordaka et al reported that the annual cost for whiplash injuries in the USA is $2.7 billion and £2.5 billion in the UK [3]. Whiplash injury can adversely affect the victim’s quality of life for a significant amount of time, ranging from weeks to few months and sometimes, symptoms may persist for years [4].

Whiplash is a soft tissue injury and the spectrum of injury may include damage to intervertebral ligaments, discs, facet joints and surrounding soft tissues [5]. Whiplash is the most common injury sustained in the rear-end motor vehicle collisions. Out of total reported whiplash cases, rear end impacts account between 38% - 52% [6]. Also, the risk of neck injury is the highest in rear end crashes as compared to other crash configurations, including frontal, side, rollover, and multiple event collisions [7]. Spine
injuries (thoracic and lumbar spine) are the second largest group in rear end impacts that are at high risk after the neck injury [7].

Behavior of the human body under these traumatic impacts and other physiological situations can be understood by performing anatomical, biomechanical, clinical, descriptive, epidemiological, and field investigations. Out of these, biomechanical studies have the unique advantage of directly examining the effect of applied external loads (simulating described scenarios) on the ensuing response of the load bearing structures of the human body.

Based on the literature, as an automobile occupant is hit from behind, the forces from the seat back compress the kyphosis of the thoracic spine, which provides an axial load on the lumbar spine and cervical spine. An out of occupant seating posture can change all the spinal curvatures compared to the normal posture. It has been clinically identified that there is an increased whiplash injury risk when the occupants head is rotated or inclined prior to the impact [53, 122]. There is very less understanding of the injury patterns when the vehicle occupant posture is in out of position (non-typical posture) and is subjected to a rear end collision.

Most of the biomechanical studies in the literature have addressed the whiplash injury during normal posture using intact cadaveric specimens, computational studies and anthropomorphic dummies. Previous researchers often use finite element (FE) models of crash test dummies to assess the injury potential of occupants during motor vehicle collisions. However, there is paucity in the biofidelity required to predict the injury of intrinsic anatomical structures of the spine (ligaments, discs and facet joints) in these
dummies and their FE representations. They correlate the occupant injury based on global kinematic indicators.

The goal of this dissertation is to develop and validate a detailed three-dimensional computational model of osseo-ligamentous human head-pelvis that describes the dynamic behavior of the human head and spine in rear end impact situations. The model has to provide insight into the motion of the head relative to the torso (global behavior) and into deformations and loads that occur within the spine (local behavior). Such a model will contribute to the understanding of injury mechanisms. In addition, this study aims to provide insight in the role of out of occupant seating posture on the global and local behavior of the head and spine system.

This approach would simulate the whiplash phenomena on novice human osseoligamentous head to pelvis finite element model and will perform a parametric analysis to analyze the biomechanical risk factors by varying the initial seated postures.

Hypothesis is that the rear end impact injury pattern of an out of pre-collision occupant posture differs from the injury patterns of a normal vehicle occupant posture and may worsen the injury. We plan to test the hypothesis using the following aims:

**Specific Aim1: To develop and validate an accurate anatomical three-dimensional finite element (FE) model of human head-pelvis.**

An accurate 3D finite element human head-pelvis model is developed to address the effect of compressive loads developed in the spinal column by straightening of thoracic spinal curvature during the impact, thus resulting in altered kinematics and kinetics. The developed model will be validated against the available set of experimental data in the literature.
Specific Aim2: To conduct a parametric analysis for identifying the potential injurious out of position occupant seating postures.

This aim focuses on identifying different initial seating postures based on the current literature and to simulate the whiplash scenario of the respective out of occupant postures using FE analysis. The resulting injury patterns will be analyzed on the basis of kinematics, stresses and strains of the spinal structures. The objective is to identify the potential high-risk injurious out of occupant seating postures.

Hypothesis 1: With the progressive increase in the out of occupant body forward posture, there will be a non-linear increase in the pattern of responses.

Hypothesis 2: The body forward out of occupant posture increases the risk of facet joint capsular ligament injury compared to other postures.

Hypothesis 3: The head turned out of occupant posture will increase the risk of upper cervical ligament injury compared to other postures.

Specific Aim3: To analyze the whiplash injury risk in out of occupant seating postures with increased impact severity.

Hypothesis: The risk of whiplash injury decreases with a lesser sled acceleration magnitude.

1.2 Outline

The second chapter of this dissertation provides a brief description of the anatomy and biomechanics of the cervical spine. This information provides a knowledge basis required to continue with the finite element model description, validation and injury prediction throughout the dissertation.
Chapter three presents a literature review of whiplash symptoms and possible causes. To begin the chapter, the reader is given an introduction to the current knowledge of symptoms associated with the whiplash injury pain. Then, review of sources of whiplash pain focuses on the organic causes of whiplash.

The fourth chapter continues with the full human head-pelvis model development and validation. This includes the description of materials and methods used for the finite element model development and the types of validation carried out. These include the ligamentous head-pelvis model under flexion, extension, lateral bending and axial rotation pure moment loadings compared to cadaveric tests, and the global as well as segmental responses of the FE model compared to postmortem human subjects (PMHS) and volunteer sled tests. Then, varying out of occupant seating postures are studied by subjecting them to different rear end impacts.

The validation results for both pure moment physiological loading and dynamic rear end impact are presented in the fifth chapter. Results of the out of occupant postures subjected to rear end impacts are also presented in this chapter. In addition to the range of motion (ROM), other parameters like ligament strains, intervertebral disc pressures and facet joint stresses for rear impact loadings are included.

These parameters are compared to published injury criteria in order to determine how different tissue injuries are affected by the out of occupant postures in chapter six. This chapter also includes a discussion of the soft tissue stress and strain results. It also provides conclusions and recommendations that follow from the results and discussion from the preceding chapters.
Chapter 2

Biomechanical Anatomy of Cervical Spine

2.1 Overview

The spine can be considered in five regions: cervical, thoracic, lumbar, sacral and coccygeal. There are 7 cervical vertebrae (C1-C7) in the neck, 12 thoracic vertebrae (T1-T12) in the chest, and 5 lumbar vertebrae (L1-L5) in the abdomen, 5 sacral (fused S1-S5) vertebrae in the pelvis (that form the sacrum) and 4 coccygeal vertebrae in the “tail bone” (that form the coccyx). The S-shape curvature of the spine is formed by lordosis (posterior curvature) in the cervical and lumbar regions, and kyphosis (anterior curvature) in the thoracic region (Figure 2-1). The sacrum also demonstrates a kyphotic curvature. The S-shaped curvature of the human spine allows us to function with an erect posture under a broad range of activities associated with daily life. Benefits include improved flexibility and energy absorption, which are needed to support the body and distribute loads during common activities such as lifting or gait.
2.2 Cervical Spine

The base of the skull (C0) and the two superior most cervical vertebrae, the atlas (C1) and the axis (C2), form the upper cervical spine. This segment is commonly referred to as the occipitoatlantoaxial complex. In the occipito-atlanto joint (C0-C1), the concave superior articular processes of the atlas articulate with the condyles of the occipital bone. Both articulating surfaces are covered with a thin layer of cartilage, the thicker of which is found on the atlas. The joint is united by a loose fibrous capsule attached to the articular margins, and is strengthened by lateral occipito-atlanto ligaments posteriorly and laterally. The anterior occipito-atlanto ligament is continuous with the capsule and provides further stabilization. These joints are surrounded by a synovial membrane. A
representative occipito-atlanto joint and the important biomechanical components are illustrated in Figure 2-2. The atlas pivots about the axis around the odontoid process (also called the dens). The dens remains positioned about the anterior aspect of the atlas under the constraint of an osteoligamentous ring formed on the ventral side by the atlas and on the dorsal side by the transverse ligament. There are two articulations between atlas about the axis. The first articulation is between the anterior arch of the atlas and the anterior aspect of the dens called the anterior atlanto-odontoid joint. The second articulation is between the transverse ligament and the posterior aspect of the dens, called syndesmo-odontoid joint. The dens and the articulating surface of the atlas are both covered by cartilage. Further stabilization is provided by the alar ligaments which connect the dens to the occipital condyles and the lateral masses as illustrated in Figure 2-3.

Figure 2-2: The three-dimensional anatomy of the occipito-atlanto joints. [8] {Adapted from White and Panjabi (pp. 197, 1978)}.  

8
In the upper cervical spine the majority of axial rotation occurs between the atlas and the axis. Flexion, extension and lateral bending are primarily a function of the occipito-atlanto joint. Together these joints account for a substantial amount of head motion.

The vertebrae of the lower cervical spine (C3-C7) all have similar bony architecture and ligamentous insertion points. The bony structures of the lower cervical spine consist of the vertebral body, transverse processes, facets, lamina and pedicles (forming the neural arch), uncinate processes and the spinous process.

The vertebral body is roughly an elliptical cylinder with a concave superior surface and convex inferior surface. The vertebral body consists of low density trabecular bone surrounded by a thin cortical shell. The inferior and superior surfaces of the vertebral body form the end-plates at which the adjacent intervertebral discs are attached. Three zones can be identified in the end-plates according to Coventry et al [10]:

\textbf{Figure 2-3: Ligaments of the occipito-atlanto-axial joint.} [9] \{Adapted from DePalma and Rothman (pp. 17, 1970)\}. 

\textbf{Figure 2-3: Ligaments of the occipito-atlanto-axial joint.} [9] \{Adapted from DePalma and Rothman (pp. 17, 1970)\}. 

...
Central zone with numerous small vascular perforations; (2) Peripheral zone with fewer large vascular perforations; and (3) Apophyseal ring which surrounds the outside of the end-plate. The superior end-plate has bony protuberances at the lateral margins termed the uncinate processes. These processes are located more anteriorly in the cephalad vertebrae and progress towards a more posterior position in the caudal vertebrae. Tulsi and Perrett observed that the tallest uncinate processes are on C5 and C6 (as much as 9 mm in height), while the smallest are observed in C3 and C7 with a complete absence in some C7 vertebrae [11]. Uncinate processes of a given vertebra may extend beyond the inferior margins endplate of next the cephalad vertebra, thus overlapping. This overlap is possible due to the geometry of the inferior endplate, which curves in the caudal direction at its lateral aspects. Goel et al reported that the uncinate processes effectively reduce motion coupling and primary cervical motion (motion in the same direction as load application), especially in response to axial rotation and lateral bending [12].

The intervertebral foramina are located at the lateral aspects of the cervical vertebrae. Generally, their shape is elliptical or round allowing for the protection of vascular and nervous structures. The facet joints (also called zygapophyseal joints) are diarthrodial joints formed between adjacent articular processes. They contain joint capsules, a synovium and are relatively free to move.

The spine can be considered as three column structure, whereby the two articular facets and the vertebral bodies with interposed discs form the three columns across which load is transferred. Facet orientation and morphology radically differ between the upper and lower cervical spine. The articular surfaces of the occipito-axial facets demonstrate a high degree of curvature, and their neutral position lies nearly in the transverse plane.
The articular surfaces of the atlanto-axial facets are typically more flat, and are oriented almost in line with the transverse plane. Panjabi et al described the three-dimensional anatomy of the human cervical facets [13].

Table 2.1: Summary of cervical facet angles based on the quantitative observation of 12 human spines. [13]

<table>
<thead>
<tr>
<th>Vertebra</th>
<th>Facet Angle (degrees from transverse plane)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior</td>
<td></td>
</tr>
<tr>
<td>C3</td>
<td>52°</td>
</tr>
<tr>
<td>Inferior</td>
<td>43°</td>
</tr>
<tr>
<td>Superior</td>
<td></td>
</tr>
<tr>
<td>C4</td>
<td>48°</td>
</tr>
<tr>
<td>Inferior</td>
<td>46°</td>
</tr>
<tr>
<td>Superior</td>
<td></td>
</tr>
<tr>
<td>C5</td>
<td>46°</td>
</tr>
<tr>
<td>Inferior</td>
<td>52°</td>
</tr>
<tr>
<td>Superior</td>
<td></td>
</tr>
<tr>
<td>C6</td>
<td>48°</td>
</tr>
<tr>
<td>Inferior</td>
<td>55°</td>
</tr>
<tr>
<td>Superior</td>
<td></td>
</tr>
<tr>
<td>C7</td>
<td>56°</td>
</tr>
<tr>
<td>Inferior</td>
<td>50°</td>
</tr>
</tbody>
</table>

The pedicles, laminae and spinous processes are all formed of relatively dense bone. Together these structures form the neural arch, which constitutes the posterior aspect of the vertebral body. The neural arch provides mechanical protection to the spinal cord. Additionally, ligaments use the lamina and the spinous process as muscle insertion sites that serve to resist large flexion moments across the cervical spine. This may be the last line of defense the cervical spine has in preventing hyperflexion injuries that may result in neurological deficits.
2.2.1 The Intervertebral Disc

The intervertebral disc is the largest avascular structure in the human body. Its composite structure consists of a gelatinous nucleus pulposus located in the approximate geometric center surrounded by fibrous annular rings (Figure 2-4). The material properties of the disc change substantially with increased age. Fissures in the intervertebral disc that run approximately parallel to the uncinate process were identified by Luschka. These joints formed in this area between the uncinate process and the superior adjacent endplate are referred to as Luschka's joints or uncovertebral joints. This feature is unique to the cervical spine, and is exceptional in that the fissures are present in the adolescent disc and are not simply a feature of degeneration [14].

![Figure 2-4: Anatomy of the intervertebral disc showing annulus fibrosis and nucleus pulposus. (Source: http://ittcs.files.wordpress.com/2010/05/img_0179.jpg)](http://ittcs.files.wordpress.com/2010/05/img_0179.jpg)
2.2.2 Annulus Fibrosus

The outer portion of the disc surrounding the nucleus is termed the annulus fibrosus. The main distinctions between it and the central nucleus pulposus is the presence of lamellae and a change in the proportions of molecular constituents between the two regions. The annulus fibrosus on the whole has about 12 annular lamellae, each approximately 1 mm thick. The posterior lamellae are generally thinner and less distinct. Within each layer of the annulus fibrosus are parallel collagen fibers oriented at 65° (±2.5°) with respect to the vertical axis, compared to the lumbar spine which is 70° (±1.8°) [15]. These fibers alternate in direction between adjacent lamellae.

2.2.3 Nucleus Pulposus

Analysis of the adult nucleus pulposus reveals a heterogeneous structure composed of proteoglycans with some collagen fibrils. In the young spine, about 80 to 88% of the nucleus pulposus is composed of water. The nucleus pulposus occupies 30 to 50% of the intervertebral disc volume [16]. While the young adult nucleus reveals a gel-like loose translucent fibrous structure, the aging adult nucleus rapidly loses its fluid properties. Thus, aging of the nucleus results in structural and functional changes including the appearance of tears and gaps [17, 18].

2.2.4 Luschka's Joints

A unique feature of the cervical intervertebral disc is the presence of fissures or clefts that run along the uncinate processes medially. These fissures, called Luschka’s
joints or uncovertebral joints, appear in the annular lamellae. Collagenous fibers adjacent to these joints have been demonstrated to run parallel to the fissure [14, 19-20]. Luschka's joints appear in the latter part of the first decade of life and increase in size throughout life [21]. In the elderly, the fissure is seen to have developed additional smaller side fissures that are irregular and very extensive. Luschka's joint has been shown to exist in young and elderly populations and is an important aspect of biomechanical modeling of the cervical spine.

2.3 Thoracic Spine

The thoracic vertebrae increase in size from T1 through T12. They are characterized by small pedicles and long spinous processes that extend inferiorly at approximately 60° as they project from the posterior of the spine. This orientation is in contrast to the cervical spinous processes that are roughly horizontal. A relatively large intervertebral foramen in the thoracic vertebrae is thought to reduce the incidence of neural compression.

The rib cage is supported by the thoracic vertebrae. The vertebral body of a typical thoracic vertebra is heart shaped with the lateral and antero-posterior dimensions being roughly equal. Two costal facets are located on either side of the vertebra. The superior costal facets are larger and placed on the upper border of the body near the pedicle. They articulate with the head of the rib at the corresponding level. The inferior costal facets are smaller and placed on the lower border in front of the inferior vertebral notch. They articulate with the rib at the level inferior to the vertebra. At T11 and T12, the ribs do not attach and are called "floating ribs". The facets are oriented almost
 vertically. The thoracic spine's range of motion (ROM) is limited by the large number of rib-vertebra articulations and the long spinous processes.

### 2.4 Lumbar Spine

The lumbar vertebrae increase in size from L1 through L5, and are the largest vertebrae in the spinal column. The pedicles are longer and wider than those in the thoracic spine. The spinous processes are horizontal and more squared in shape. The intervertebral foramena (through which the nerve roots exit) are relatively large. Yet, nerve root compression is more common than in the thoracic spine. These vertebrae bear much of the body's weight and related biomechanical stress. Facet orientation is almost vertical.

### 2.5 Sacral and coccygeal Spine

The sacrum is formed from five fused vertebrae (abbreviated S1 through S5). The sacrum is roughly triangular in shape and joins the two iliac bones to form the posterior pelvis below the lumbar region. The last lumbar vertebra (L5) articulates with the sacrum. Immediately below the sacrum are five additional bones, fused together to form the Coccyx (i.e. tailbone).

### 2.6 Spinal Ligaments

Ligaments of the spine act to carry tensile forces that resist excessive motion, thus stabilizing the spine. The anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), capsular ligaments (right and left CL), ligamentum flavum (LF) and the
interspinous ligament (ISL) are of particular biomechanical interest. Anatomical locations of these ligaments reveal that any type of rotational motion will produce a tensile force in at least one of these ligaments.

The ALL is a continuous ligament which originates as a band attached to the inferior surface of the occiput and ends at the first segment of the sacrum. It is firmly attached to the anterior surface of the vertebral bodies and may be loosely attached to the intervertebral discs as well. The lateral edges are blended with the periosteum. Tension in the ligament develops when the spine is in extension.

The PLL is also a continuous ligament which originates at the occiput and runs down the posterior aspect of the vertebral column from within the neural canal before terminating at the coccyx. The PLL is similar to the ALL in that it is firmly attached to the vertebral bodies. However, the PLL is also firmly attached to the intervertebral disc.

The articular facets are surrounded by the right and left capsular ligaments (CLs) which enclose the joint cavities. The CLs serve to stabilize the articulation of the adjacent facets, thereby reducing excessive separation of the surfaces. While coupled motion which is seen in the cervical spine is due in part to the facet orientation, it is unlikely that these motion patterns would be maintained in the absence of the CLs.

The ligamentum flavum (LF) is an extremely elastic ligament. It has also been called the 'yellow ligament', which is due to its relatively high elastin fiber content when compared to other spinal ligaments. The LF is a flat band that spans the space between the laminae of adjacent vertebrae. Biomechanically, this ligament acts to resist flexion.

The interspinous ligament (ISL) bridges the gap between adjacent spinous processes. This ligament is slack when the cervical spine is in a neutral position. Tension
develops only when a significant amount of motion has already occurred and transmissions of tensile forces have been taken up by other ligaments such as the PLL, CLs and LF. The relative position of the ISL with respect to the intervertebral disc and its initial slackness, suggest that it plays a role in resisting only larger flexion motions. The ISL can effectively be considered an 'endstop'. In the cervical region, posterior to the spinous process, lies the ligamentum nuchae which extends from occiput to C7. This ligament is continuous with the supraspinous ligament in the lumbar and thoracic regions, which runs from C7 to the sacrum.

2.7 Intact Spine Biomechanics

The spine acts to support the trunk and transfer muscular loads to the lower body. Compressive loads of approximately 75 N on the cervical spine and 400 N on the lumbar spine are experienced while standing upright. The posterior column transmits approximately 20% of this load, while the anterior column bears the rest. The natural S-shape curvature of the spine is due to cervical lordosis (concave backwards), thoracic kyphosis (concave forwards) and lumbar lordosis with average values of 90°, 39° and 57°, respectively. In trauma, clinical intervention may be required to restore these lordotic and kyphotic angles, which are critical to normal alignment and kinetics.

The load-bearing axis of the spine is the longitudinal region of the spinal column that bears a substantial amount of the axial load and about which excessive spinal element distraction or compression does not occur under flexion or extension [22]. It is generally located about the mid-dorsal region of the vertebral body (Figure 2-5). Traumatic conditions affect the load sharing process of the spinal column ultimately
shifting the loading-bearing axis and destabilizing the spine. Hence, proper stabilization systems need to be implemented to preserve the load-bearing axis following trauma.

Figure 2-5: (A) The load-bearing axis (shaded region) is generally considered to be located in the region of the middle column of Denis, (B) the load-bearing axis is shifted dorsally in the cervical spine during extension, and (C) it is shifted ventrally during flexion. (Source: http://img.medscape.com/fullsize/migrated/448/307/nf448307.fig6.gif)

Spinal kinematics are often described based on the principle axes of motion: flexion-extension, lateral bending and axial rotation. Together, flexion-extension can be referred to as sagittal bending, which generally occurs with fore or aft motion. Lateral bending refers to bending to either side, and can be either left or right lateral bending. Axial rotation refers to motion about a superior-inferior axis, for example rotating the head to the left or right.
Chapter 3

Whiplash Injury Review

3.1 Introduction

In 1995, the Quebec Task Force on whiplash conducted an extensive literature review and defined whiplash as "an acceleration-deceleration" mechanism of energy transfer to the neck. It may result from rear end or other motor vehicle collisions, but can also occur during diving or other mishaps. The impact may result in bony or soft tissue (whiplash) injuries, which may lead to a variety of clinical manifestations (Whiplash-Associated Disorders, WADs) [23].

Annual costs for these whiplash injuries in USA alone were estimated to be $2.5 billion [3]. Even in other countries, rear end impacts were found to be responsible for most of the WAD injury patterns and similar associated costs have been reported [24-25].

Whiplash associated symptoms include neck pain, shoulder pain, headache, paresthesia, weakness, dizziness, visual problems, tinnitus, Cognitive impairment and back pain [26-33]. In 2006, Bogduk did a comprehensive review of literature and reported that neck pain, headache, and shoulder pain were the most common whiplash symptoms in both acute and chronic cases [33].
For modeling the whiplash scenario, it is absolutely critical to understand the whiplash injury associated symptoms. This chapter focuses on understanding the whiplash associated symptoms and identifying the underlying structures which form the sources of pain by reviewing the clinical evidences in the literature. Then, it proceed with the understanding of injury mechanisms by reviewing the myriad of biomechanical literature.

3.2 Clinical Observations

Whiplash injury can reduce victim’s quality of life for a significant amount of time, ranging from weeks to few months and sometimes, symptoms may persist for years [4]. Based on the amount of time symptoms persist, whiplash injuries can be grouped into "acute injuries" (last a short period of time) and "chronic injuries" (last for a long time). The European Foundation of IASP (International Association for the Study of Pain) Chapters (EFIC) defines chronic pain as pain lasting longer then 3-6 months [34].

Sturzenegger et al performed a cohort study with 137 patients and studied the influence of accident mechanisms on the symptoms and signs after whiplash injury [32]. The onset of the symptoms occurred in 45.9% of the patients within 1 hour after the accident, in 28.6% of patients within 24 hours after the accident, while for a smaller group (25.5%) the symptoms occurred after 24 hours. More severe symptoms were observed for three accident mechanisms: an unprepared occupant, rear-end collision, and rotated or inclined head position at the moment of impact [32].
Radanov et al evaluated a sample of 117 patients who had whiplash associated symptoms. Initial examination was performed 7.2 ± 4.2 days after trauma, and follow-up examinations 3, 6, 12, and 24 months later. At 2 years, 18% of patients still had injury-related symptoms. Also, a higher incidence of rotated or inclined head position at the time of impact was correlated with increased severity of symptoms [28].

Other studies reported that more than 50% of whiplash patients had symptoms one year following the whiplash [35-36]. Another study in Sweden evaluated the severity of injuries among the car occupants, especially drivers and reported that 1 out of 10 occupants remained disabled at least one year after a collision [37]. Quebec study showed that 3% of people with WADs were still absent from their usual activities or work, one year after the collision [23].

In a clinical review by Bogduk et al, around 40% of the patients with whiplash injuries had chronic pain and they also indicated that approximately 10% will have constant, severe pain indefinitely [38].

Researchers have proposed and investigated several structures in the neck as possible sources of whiplash associated symptoms or chronic pain. These structures include the capsular ligaments and facet joints, other ligaments, the vertebral discs, the vertebral artery, the dorsal root ganglion and dorsal root, and muscles. Out of these structures, the leading contenders for explaining chronic neck pain following WADs are injuries to the facet joints, the intervertebral discs and the upper cervical ligaments [38].
3.2.1 **Facet Joints**

Based on the clinical research, the facet joint has been reported as the most common source of neck pain, accounting for 25-62% of patients suffering from neck pain [39]. In 2004, Manchikanti et al evaluated 500 patients with spinal pain and identified painful cervical facets in 55% of patients with neck pain. This group investigated the facet joint pain in patients by placing diagnostic (short term and long term) anesthetic blocks in the facet joints. They reported a positive pain relief when both blocks reduced pain and observed that the duration of pain relief was increased with the long term block [40].

Cervical facet joints are particularly relevant to the whiplash associated symptoms. In 1991, Jonsson et al studied 22 cervical spines from traffic accident victims with fatal cranio-cerebral injuries (19 male, 3 female, mean age 26 years) and found 77 facet joint or ligament flavum lesions [41]. Clinical and animal experimental studies of whiplash have revealed tears of joint capsules, hemarthroses, and fractures of the articular cartilage and subchondral bone [42-44].

Detecting these injuries is tedious and they are not frequently detectable on conventional X-ray examinations. It has been demonstrated that only way to diagnose the painful cervical facet joints is through anesthetic blocks [45].

Fifty consecutive referred patients with chronic neck pain after whiplash injury were studied using double blind, controlled diagnostic blocks of the cervical facet joints. Similar to the earlier study, positive diagnosis was made only if both short term and long
term blocks relieved the patients' pain. Painful joints were identified in 54% of the patient's and concluded that facet joints were the most common source of chronic pain after whiplash [45].

Similarly, Lord et al used an anesthetic block and a placebo in a double blind study with 68 patients. They observed a 50% prevalence of C2-C3 level joint pain among patients with dominant headache and also supported that cervical facet joint pain is common among patients with chronic neck pain after whiplash (Figure 3-1) [46].

Figure 3-1: Segmental level distribution of facet joints in 31 patients with diagnosed facet joint pain [46].
3.2.2 **Intervertebral Discs**

In an investigational study, 22 cervical spines were extracted from traffic accident victims with fatal cranio-cerebral injuries (19 male, 3 female, mean age 26 years). They investigated those spines and found 22 disc lesions [41].

Petterson *et al* performed a blinded study to evaluate the relationship between magnetic resonance imaging findings and clinical findings after whiplash injury. Magnetic resonance imaging (MRI) and clinical examination were performed (blinded manner) at a mean of 11 days after trauma and after 2 years, for 39 patients (20 women and 19 men with a mean age of 32 years). They found that 33% of patients had moderate to severe disc herniations. They found the symptoms related to impingement of the medulla oblongata at 2 years follow-up in five of six patients with severe disc herniations. They suggested that disc pathology seems to be one contributing factor in the development of chronic symptoms after whiplash injury [47].

3.2.3 **Cervical Ligaments**

In postmortem studies, researchers have documented that even in the absence of dislocation during fatal head/neck injuries, cranio-vertebral ligaments are vulnerable to trauma. One study reported that out of 21 cases with head-neck injuries without dislocation, 99% of showed either laceration or sprain of the alar ligaments [48]. In
another study, alar ligament lesions were reported in 36% of total 30 cases investigated [49].

Magnetic resonance imaging (MRI) has been used by several investigators to examine alar ligaments in human volunteers after trauma. In a study performed by Patijn et al in 2001, alar ligaments were examined in 12 patients with chronic whiplash syndrome and 6 controls. In spite of ligament identification in all cases, they could not prove alar ligament damage as a causative factor of the impairment [50]. Few others tried to use MRI in finding the alar ligament lesions, but could not achieve clinical relevance [51-52].

In 2006, Krakenes and Kaale et al examined 92 whiplash injured individuals (33 males and 59 females with a mean age of 40 years; range, 14–61 years) sustaining a frontal or rear-end collision 2 to 9 years (mean, 6 years) [53]. All these patients were prospectively registered and who met the criteria for WAD 2 (neck pain, decreased range of motion, and palpation tenderness) after 3 months were included. Individuals (n=30) without any known neck injury (11 men and 19 women; mean age, 46 years; range, 28–66 years) were included as control group. They defined an imaging protocol for the craniovertebral junction that used high spatial resolution with good contrast and 2mm thick sections to successfully identify lesions to the alar ligament, tectorial membrane, posterior atlanto-occipital membrane (PAOM), and transverse ligament. Three radiologists interpreted the images in this blinded study. To monitor the status of the injury after every three months, they created a grading criteria, Table 3.1.
They obtained moderate to good agreement between observers (kappa values) for the ligaments examined. This study also provided questionnaires to the whiplash group to find out if their head was rotated at the moment of impact and if they were involved in a frontal or rear-end impact. High-grade changes in the transverse ligament were significantly more common in frontal than in rear-end collisions (61.1% vs. 10.5%, \( P < 0.001 \)) and atlanto-occipital membrane (37.1% vs. 5.3%, \( P < 0.001 \)). Also, they reported significantly more high grade alar ligament lesions among those with rotated head position (Table 3.2) than those with neutral head position (85.1% vs. 46.7%, \( P < 0.001 \)).
Table 3.2: MRI classification of ligaments in 92 whiplash patients by impact direction at the moment of impact and by head position [53].

<table>
<thead>
<tr>
<th>MRI Grade</th>
<th>Head Position</th>
<th>Impact Direction</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Neutral [no. (%)]</td>
<td>Rotated [no. (%)]</td>
</tr>
<tr>
<td>All ligaments*</td>
<td>24 (53.3)</td>
<td>7 (14.9)</td>
</tr>
<tr>
<td>0-1</td>
<td>21 (46.7)</td>
<td>40 (86.1)</td>
</tr>
<tr>
<td>Transverse ligament</td>
<td>28 (62.2)</td>
<td>27 (57.1)</td>
</tr>
<tr>
<td>0-1</td>
<td>17 (27.8)</td>
<td>20 (42.2)</td>
</tr>
<tr>
<td>Tectorial membrane</td>
<td>37 (82.2)</td>
<td>39 (83.0)</td>
</tr>
<tr>
<td>0-1</td>
<td>8 (17.8)</td>
<td>2 (17.0)</td>
</tr>
<tr>
<td>Posterior atlanto-occipital membrane</td>
<td>35 (77.8)</td>
<td>35 (74.5)</td>
</tr>
<tr>
<td>0-1</td>
<td>10 (22.2)</td>
<td>12 (25.5)</td>
</tr>
</tbody>
</table>

*Highest assigned grade if difference between right and left.

3.3 Biomechanical Approach

In the field of biomechanics, voluminous amount of literature has been published on the testing and understanding the whiplash mechanism, which is associated with rear end impacts. The trend so far in this biomechanical literature has been to study the injuries due to rear end impact and identify the sources of pain along with underlying mechanisms using the following approaches:

- **In vivo** studies
- Anthropomorphic test dummies
- **Ex vivo** studies
- computational modeling
3.3.1 **In vivo Studies**

Studies on neck injuries using human volunteers have been carried out in USA especially since 1960s [54]. Matsushita *et al* was the first to use cineradiography technique to analyze the cervical spinal motion of human volunteers during impact. They reported that the cervical spine first went into flexion and then extension. Also, they demonstrated that cervical spinal motion was influenced by the seating posture [55]. In the leaning-forward posture, the cervical spine was affected by compression load resulting from upward movement of the upper thoracic spine, and the cervical spine length was shortened [55].

Few researchers performed sled test simulations with volunteers and investigated the human neck response during impact with high-speed video camera, accelerometers, and electromyography. They found that the actual kinematic responses of the human head, neck and trunk that occur during low velocity rear end impacts were more complex than previously thought [56-57]. However, they did not analyze the motion of each cervical vertebra during impact.

Kaneoka and Ono *et al* analyzed motion of each cervical vertebra during simulated rear-end car collisions [58]. Ten healthy male volunteers were recruited for this study. Volunteers sat on a sled that glided backward on inclined rails and crashed into a damper at 4 km/hr. Rotational angle of each vertebrae and C5-C6 FSU instantaneous axes of rotations were quantified using cineradiographic technique (Figure 3-2).
Three distinct patterns of cervical spine motion were observed after impact namely extension only, flexion-extension, and no extension. Most of the volunteers demonstrated motions belonging to flexion-extension group. In this group, they reported that cervical spine was forced to move from the lower vertebrae during rear-end impacts. In the early phase, C6 rotated backward before the upper vertebrae. Thus, the cervical spine showed a flexion position (initial flexion). After C6 reached its maximum rotational angle, C5 was induced to extend. Due to pattern of the upper motion segments going into flexion, and the lower segments into extension, the cervical spine took an S-shaped
position. They concluded that this S-shaped pattern could be related to the injury mechanism.

In a recent study, Dehner et al simulated rear-end sled collisions using 8 human subjects to identify sequences of motion in which the risk of injury to the cervical spine may increase [59]. Motion data was captured using high speed camera at a rate of 500 frames/sec and acceleration data using accelerometers. The sequences of motion were identified based on acceleration and motion parameters (Table 3.3). They concluded that during the extension phase, acceleration and movement pattern occur that could lead to cervical spine injuries.

Table 3.3: Sequences of motion classified into different phases based on the kinematics during respective occurrences of time in a rear end impact in vivo experiment performed by Dehner et al [59].

<table>
<thead>
<tr>
<th>Phases</th>
<th>Time stamps</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Latency</td>
<td>0-67 ms</td>
<td>After the sled impact, the head initially remains in its baseline position.</td>
</tr>
<tr>
<td>Translation</td>
<td>67-85ms</td>
<td>Dorsal translation of the head alone without a rotational component.</td>
</tr>
<tr>
<td>Extension</td>
<td>85-139 ms</td>
<td>Combined dorsal extension and translational motion of the head. At the same time, the beginning of head–headrest contact takes place.</td>
</tr>
<tr>
<td>Rebound I</td>
<td>139-191 ms</td>
<td>Head returns to its initial position after the maximum head extension is reached.</td>
</tr>
</tbody>
</table>
Even though these tests capture actual head and cervical spine kinematics, they are limited by the fact that the forces acting on the cervical spine can only be determined approximately. Direct force measurements are not feasible and this requires reference to indirect measurement parameters in the form of acceleration data [55].

A study examining electromyography signals from neck musculature during human volunteer rear impact auto collisions reported that initial muscle activation did not occur until 100 msecs after event starts, and full muscle contraction did not occur until 150–170 msecs after event onset when the head–neck is already in extension [57]. Furthermore, animal experiments have demonstrated that the time to develop muscle force is approximately 200 msecs [60]. The reflex muscle contraction in the unaware occupant in whiplash does not occur in sufficient time to alter markedly spinal kinematics, resulting in soft tissue injury during the retraction phase [61]. Therefore, we are not considering the effect of muscles on whiplash injury in this study, considering the occupant to be unaware of the impact.

Overall, these human *in vivo* studies could not provide a better understanding of the intrinsic biomechanics of structural components potentially involved in injury and/or pain producing processes. Moreover, contributions made to understand the injuries used the biomechanical parameters such as the overall motions of the head–neck. Also, it is
impractical to measure the forces or strains on intervertebral discs and individual ligaments in a living subject and it would be unethical to expose the human volunteers to higher accelerations.

3.3.2 Anthropomorphic Test Dummies

Anthropomorphic models (also referred to as “test dummies”) are full body replicas that are used to simulate the effects of vehicle impact on humans. Population sampling has allowed for the creation of models specifically suited for the study of rear-impact collisions. The Anthropomorphic dummies (ATD) are made to represent the body mass and properties of the average human in order to best replicate the musculoskeletal response during whiplash. Due to these advantages, the ATDs are widely used in the automobile industry for various safety evaluations.

Svensson et al used a rear impact dummy (Hybrid III) fitted with a modified neck (RID-neck) to show that changing the head restraint backset greatly affected the forces and bending moments at the superior and inferior ends of the cervical spine as well as the head accelerations and angles [62].

While ATDs can replicate some behavior of whiplash, the construction of the neck joints within the model limits the accuracy of test results. Even though ATDs have trouble simulating the complex motions of the spine during vehicle collisions, their use has led to various safety improvements since their inception.

In summary, similar tests of human volunteers on crash dummies might provide useful kinematic data but, as there is no attempt to replicate the intervertebral disc and
ligament structure in the dummies, cannot throw light on the actual mechanism of injury [63-64].

3.3.3 Ex vivo Studies

In vitro studies were effective to understand the biomechanics from sub injury to injury producing load or acceleration levels in the past decade. All the isolated head-neck rear end impact tests have provided invaluable information in understanding the whiplash injury mechanisms.

Panjabi and group conducted several studies to understand the mechanism of whiplash injury [65-68]. They estimated the potential risk of injury based on the damage caused by the impacts to the spinal structures (such as facet joints, capsular ligaments and intervertebral discs) compared to the tissue damage by physiological range of motion. In all their studies, they used isolated cervical spines and attached surrogate head to simulate the whole cervical spine model.

Grauer and Panjabi et al imposed several rear end impacts with the maximum sled acceleration ranging from 2.5g to 8.5g on 6 cadaveric specimens [65]. Motion of individual vertebrae along with surrogate head were recorded using high-speed cinematography (500 frames/sec). Physiologic flexion and extension motions were also recorded with an Optottrak motion analysis system by loading the specimens up to 1.0 Nm before the impact testing. The experimental setup for the impact is shown in Figure 3-2. They reported that the peak intervertebral rotations of C6-C7 and C7-T1 significantly exceeded the maximum physiologic extension in all the impact cases. While
the lower levels were subjected to maximum extension, the upper cervical levels were consistently in flexion resulting an S-shaped curvature. This typical pattern was observed before the neck was in full extension. Figure 3-3 depicts the observed S- and C-curvatures and phase differences between the spine and head.

Figure 3-3: Schematic diagram of the bench-top whiplash sled apparatus used by Panjabi and group for testing the isolated head-neck cadaveric specimens [65].

Figure 3-4: Schematic diagram of the normal posture (NP), observed S- and C-curvatures and phase differences between the spine and head during the rear end impact experiment [65].
In 2004, Pearson et al used similar set up and demonstrated that the facet joint compression exceeded physiological limits [68]. Later, Ivancic et al performed a study with 12 cervical spines (6 rear end impact-exposed and 6 control) and prepared 66 capsular ligament specimens (C2/3 to C7/T1) to determine if rear end impact caused increased capsular ligament laxity. This was done by applying quasi-static loading to whiplash-exposed and control capsular ligaments. They reported a significant increase in capsular ligament laxity (0.9 mm at 5 N tensile load) compared to the control ligament [69].

Similarly, Panjabi et al showed that the annular tissue of the disc is potentially injured due to the rear end impact by comparing the whiplash exposed disc fiber strains to the physiological levels [70]. They also reported that the axial deformation at the anterior disc region exceeded physiologic levels at even 3.5 g peak sled acceleration.

Other ligament structures in the neck were studied by Tominaga et al, in which 98 bone-ligament-bone specimens (C2–C3 to C7-T1) were prepared from six cervical spines following 3.5, 5, 6.5, and 8 g rear impacts and pre- and post-impact flexibility testing. These bone-ligament-bone specimens were elongated to failure at a peak rate of 725 (STD 95) mm/s. They reported that the whiplash-exposed ligaments had significantly lower failure force, 149.4 vs. 186.0 N, compared to the control ligaments supporting the ligament injury hypothesis of whiplash syndrome [71]. These tests could not identify the individual ligaments being damaged and exactly at what spinal level those ligaments are at high risk.
Yoganandan and group also performed rear end loadings on intact human cadaver head–neck complexes. Their results supported the formation of the non-physiologic S-curve. They also reported greater segmental angles and facet joint kinematics of the cervical spine for females than males in the rear end impacts. In addition, they observed kinematics of lower facet joints to be fundamentally different between posteroanterior acceleration loading and normal physiological extension, which explains the increased vulnerability of facet joints to injury [72-74].

All these isolated head-neck tests have provided invaluable information into the complex. However, several problems remain: firstly, cervical spine ligaments have been associated with the whiplash injury, especially in the upper cervical region. The head of the specimen replaced by surrogate head destroys vital ligaments in the C0–C2 complex and secondly, the postero-anterior loading of the rear end impacts do not consider the effects of thoracic ramping on the isolated head-neck complex. These isolated head-neck specimens do not account for the effects of compression due to the straightening of thoracic spine [75-77].

Few investigators used full post mortem human subjects (PMHS) to accurately reproduce whiplash injuries [78-81]. Yang et al proposed capsular stretch as injury mechanism causing neck pain [78]. Later, the same group conducted tests of rear end impacts using PMHS and captured the cervical motion with high-speed x-ray images. They reported the absence of S-curvature and supported the hypothesis that significant stretch of the facet joint may be the source of neck pain [81]. However, these studies
could not elucidate the damage caused in the critical spinal structures (parameters like intervertebral disc stresses and upper cervical ligament strains).

3.3.4 Influence of seating postures

In rear impacts, the thoracic interaction with the seatback tends to straighten the thoracic kyphosis and create a ramping effect at the cervicothoracic junction, wherein the upper thorax translates superiorly while rotating posteriorly. This phenomenon has been reproduced experimentally using human volunteers and full body cadaver specimens subjected to rear impacts [79,82,83]. The magnitude of these motions depend on numerous factors [84]:

- Stiffness of the seatback
- Initial curvature of the thoracic spine
- Angle of the seatback
- Magnitude of rear impact
- Awareness of the occupant
- Occupant seating posture

A study on different factors (Figure 3-4) influencing whiplash associated disorders by Wilkerson et al indicated that "occupant seating position" was the most significant parameter and there was no effect of gender and body mass index (BMI) [85].
As indicated in the clinical observations section, it was also clinically evaluated that the occupant seating position just before the impact may increase the severity of the whiplash injury [53]. This section focuses on reviewing the features of past work done with respect to the out of occupant seating postures.

Most of the studies in the past concentrated on understanding the injury mechanisms of whiplash with normal posture of the vehicle occupant, however, few of them attempted to investigate certain aspects related to out of occupant seating position. Sundararajan et al performed PMHS tests with normal and out of occupant head neck positions (normal posture, zero clearance between head and head restraint, body forward position) [81]. Compared to rest of the two postures, body forward position had the largest facet joint stretch. These studies indicate that the abnormal postures may increase the severity of facet joint injury.
Few other human volunteer studies were conducted in the past, where, along with the normal posture, few out of occupant postures were simulated in the experiment. The details of these studies are presented in the Table 3.4 [55-57].

Table 3.4: Details of the human volunteer studies conducted with out of occupant seating postures.

<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>No. of Tests</th>
<th>Out of occupant position</th>
<th>Limitations</th>
</tr>
</thead>
<tbody>
<tr>
<td>McConnell</td>
<td>1995</td>
<td>2</td>
<td>Head turned left</td>
<td>Human volunteer tests:</td>
</tr>
<tr>
<td>Szabo</td>
<td>1994</td>
<td>1</td>
<td>Increased head-head restraint position</td>
<td>• With low impact velocities.</td>
</tr>
<tr>
<td>West et al</td>
<td>1993</td>
<td>1</td>
<td>Leaning forward</td>
<td>• Did not account for biomechanics of internal spinal structures</td>
</tr>
<tr>
<td>Matsushita et al</td>
<td>1994</td>
<td>4</td>
<td>• Head rotated (Left/Right) • forward lean • bending laterally</td>
<td></td>
</tr>
</tbody>
</table>

Even though these researchers performed the experiments with volunteers in out of occupant seating positions, studies were conducted in a very low number to report any significant data on kinematics of out of occupant posture impacts and also, human volunteer studies do not account for biomechanics of internal spinal structures.

The out of occupant seating posture will induce initial loads and deformations in the spinal structures, which can alter the nature of the injury when subjected to the rear end impacts. Currently, there is no data available in the orthopedic literature to assess the risk of injury based on kinetics (disc stresses, facet stresses and ligament strains) with such non-typical occupant postures.
3.3.5 Computational Modeling

Several research groups have investigated whiplash using computational models of the cervical spine for automotive research. In this section, finite element models that were used to study the rear end impact whiplash injury are reviewed. Major computational models reviewed all have their limitations which are discussed along with the model details (Table 3.5).

Before a decade, only one study of the full three-dimensional cervical spine as an FE model has been used to investigate the effects of whiplash. Kleinberger developed this three-dimensional C0-T1 model with highly simplified topology and assigned linear material properties to all the regions [86]. Also, there was no difference between the topology at different levels of vertebra in the cervical spine. However, when subjected to the dynamic whiplash loading, this model predicted promising results in terms of global head displacements and thus proved the potential of FE modeling for whiplash injury investigation.

Table 3.5: A summary of computational models that were used to study whiplash injury.
The Skull-T1 multibody model [87] developed by Stemper et al does not use elements to represent the intervertebral disc and facet joints, but rather uses elastic constraints between the vertebrae to model these joints. This method is able to accurately predict global kinematics but cannot assess injury to the disc. Also, the TNO model is a multi-body model rather than a finite element model, which means the model cannot predict stress or strain in the vertebrae (Figure 3-3) [87].
The neck FE model developed by Schmitt et al have rigid vertebra which would limit the vertebral stress or strain (Figure 3-4) [88]. In this study, they used the C1-T1 finite element model with fluid structural interaction in the spinal canal to study the pressure in the spinal fluid during a 4g rear impact. There was considerable amount of difference in the pressure value predicted by the model compared to the cadaver study. However, neither the cadaver study nor the FE model could conclude about nerve damage based on the pressure spike results.
Another FE model (C0-C7) from Nanyang Technological University group (NTU) had predicted head-neck segmental motions which were in good agreement with respect to experimental cadaveric tests [89]. However, they assigned simplified linear properties to the intervertebral discs. Moreover, this group mainly focused on predicting kinematics due to rear end impact and the main essence of FE model of predicting local tissue damage was missing in their studies.

In addition, all these isolated head-neck models simulated only posteroanterior loading which do not account for the thoracic straightening effects in the rear end impacts. To replicate the involvement of thoracic-seat back interaction, a full body model is necessary.

Human Model for Safety (HUMOS) and Total Human Model for Safety (THUMS) are the only two full body FE models existing in the literature. In HUMOS full
body model, the T1 and C7 vertebrae were considered to be rigid bodies connected with a numerical joint (no elements for disc and facet joint) to model the T1–C7 physical joint which do not allow the study of potential injury on or around cervico-thoracic junction and it was not validated for the rear-impact [90].

Figure 3-9: HUMOS computational model from Tropiano et al, 2004 [90].

The HUMOS FE model in a seated driving position was not validated for rear-impact, yet they present vertebra stress and ligament strains from a 15g rear impact [90-91].

FE THUMS model consists of ligaments modeled as shell elements and were assigned with the linear material properties [92, 93]. The purpose of the ligaments is to transfer the tensile load and the shell element formulation can allow compression. Also,
the intervertebral discs and ligaments were assigned with the linear material properties which may not predict local tissue damage accurately. In addition to this, both these models have less than 80,000 elements which is considered to be a coarser mesh compared to 298,313 elements in our cervical spine model alone.

![THUMS computational model from González et al, 2010][1]

Figure 3-10: THUMS computational model from González et al, 2010 [93].

In short, FE method offers the advantage that it can handle complex geometric configurations and material, contact and geometric nonlinearities. It provides feasibility in performing parametric analytical studies. But the models reviewed above have their own limitations in terms of either modeling degree of detail to predict the local tissue deformations or did not consider the main compressive loading effect produced by the seatback-spine interaction.
3.4 Summary

Based on the clinical research, there is an increased whiplash injury risk when the occupant is subjected to an out of occupant seating posture prior to impact. The facet joints have been reported as the most common source of neck pain. Intervertebral disc lesions were identified and can be at risk during the whiplash trauma. Cranio-vertebral ligaments are vulnerable to trauma especially when the occupant is subjected to an abnormal posture (head turned) before the rear end impact.

The overall position for the experimental studies is that testing on living human subjects is limited to conditions which cannot implicate local tissue damage. While testing on dummies, even under realistic conditions, does not reflect the full anatomical structure.

Human cadaveric specimens are rare, costly, and could present an ethical challenge to our society. The associated costs and complexity for the intact PMHS tests using high speed imaging technique is another drawback. In addition, the investigator using cadaveric tissue might not have control over the age and physical condition of the donor. In clinical observations and experimental research, tissue loads and deformations are often impracticable if not impossible to determine. Above all, the in vitro experiments do not account for the intervertebral disc and ligament stresses and strains.

Axial compression in the neck, together with the shear force, is responsible for the higher observed frequency of neck injuries in rear end impacts. This axial compression
occurs during the first phase of the rear end impact due to the straightening of thoracic spine and this axial compression reduces the shear stiffness of the cervical spine (loosening of cervical ligaments) and makes it easier for the shear type soft tissue injuries to occur.

All the computational models in the literature have their own limitations. Most of the computational models consists of isolated cervical spine simulations which do not account for the above mentioned phenomenon accurately. HUMOS and THUMS are the only two full body FE models existing in the literature. In HUMOS full body model, the T1 and C7 vertebrae were considered to be rigid bodies connected with a numerical joint (no elements for disc and facet joint) to model the T1–C7 physical joint which do not allow the study of potential injury on or around cervico-thoracic junction and it was not validated for the rear-impact. FE THUMS model consists of ligaments modeled as shell elements and were assigned with the linear material properties. The purpose of the ligaments is to transfer the tensile load and the shell element formulation can allow compression. There is no data in the literature which explored the risk factors associated with the rear end impacts especially when the subject is in out of occupant seating posture.

All the above factors in the literature were instrumental in developing a non-linear osseoligamentous head-pelvis finite element model that included head, full spine with ribcage and pelvis to study the effect of different initial seating postures on the ligament and disc injury caused due to the rear end impact.
Chapter 4

Materials and Method

4.1 Introduction

This section consists of a detailed, non-linear, three dimensional osseo-ligamentous human head-spine-pelvis finite element model development followed by the validations with static and dynamic loadings respectively. Then, a parametric analysis will be performed by varying initial occupant seated postures with the same rear end impact loading scenario. The effect of increased acceleration magnitudes will also be investigated on the potential injurious out of occupant postures.

4.2 Finite Element Model Development

The first objective of this study is to develop a detailed, non-linear, three dimensional osseo-ligamentous human head-spine-pelvis FE model which enables the
study of rear end impact loading. The three major steps involved in the development of the finite element model are:

**Pre-Processing:** This is the first step in FEA, in which a finite element model of the structure to be analyzed is constructed. Most FEA packages require a topological description of the structure's geometric features as input. This description can be in one, two or three dimensional form, representing line/polyline/spline, surface or structural elements, respectively. However, 3D models are predominantly used in most of the cases. Design files, CAD models and pre-existing digital scans can be imported into an FEA environment for a finite element model development. Once the finite element geometric model is complete, a meshing procedure is used to define and discrete the model into small elements. A finite element model in general is defined by the creation of a mesh network that includes the geometric arrangement of elements and nodes. When the FE model is created, the material properties are assigned to the individual components with appropriate interactions and constraints defined between components. Finally, boundary conditions and loads are assigned to the model.

**Analysis:** Next, a solver is used to analyze convergence of the model within the predefined boundary conditions, loads and constraints. To better simulate the physical conditions being tested, various types of analyses such as static or dynamic (time dependent) may be considered. In static simulations, inputs and outputs are independent of time and the system is solved to balance the load and boundary conditions. In dynamic simulations, time may affect the input and output parameters, thus the behavior of the
model varies with time. Examples of dynamic simulations are impact loading and low rate creep, in which the rate of loading affects mechanical response in the model.

**Post-Processing:** Once the simulation is complete, post-processing is performed for visualization of the analysis results. Outputs of the analysis serve to evaluate each node and each element for parameters such as displacement and stresses-strains, temperature, etc. respectively.

The pre-processing step in the development of the human head-pelvis finite element model is explained briefly (Figure 4-1):

![Figure 4-1: Steps involved in the pre-processing or creating the three dimensional human head-pelvis finite element model.](image)
4.2.1 **Geometry Creation:**

A set of CT and MRI scans of a normal human subject with 1mm slice thickness, free of anatomic abnormalities were procured. MIMICS (Materialize Inc., Leuven, Belgium) software was used to process and edit the 2D image data to construct the three dimensional geometry of head and cervical spine. The powerful segmentation tools of this software allowed us in segmenting the CT/MRI images, taking measurements and creating the resulting 3D geometry (shown in Figure 4-2). From here, the three dimensional geometry was exported in *STL format to IAFE MESH (University of Iowa, Iowa) software for the further processing.
Figure 4-2: Procedure depicting the import of CT images into MIMICS software for development of 3D models of C1-C7 spine segment. Top left, top right and bottom left images depict the frontal, axial and Sagittal views, respectively.

4.2.2 Meshing Procedure

The geometries in *STL format were transferred into IAFE MESH software for creating the mesh structure. It involved construction of a series of building blocks around the 3D geometrical structure, assignment of the desired mesh density and projection onto the surface representation (shown in figure 4-3), thereby creating a 3D FE model. Finally, the mesh quality module in the software was recruited to evaluate and improve the mesh quality. The head and ribcage finite element models were also created using the same procedure.

Figure 4-3: Procedure depicting the creation of vertebral finite element meshed model from the 3D geometry.
The meshing of the thoracic, lumbar spine and pelvis finite element models were created in a different manner as they were developed previously. The CT images, obtained in transverse sections were used to develop the mesh structure for components of the model. For this purpose, the four nodes characterizing a particular element were digitized (using Image J software) to obtain their X and Z coordinates with respect to the global coordinate system. The Y coordinate was taken as the depth of the corresponding transverse slice on the CT film. Under the assumption of symmetry, half of the model was digitized, while the other half was generated by reflection across the mid-sagittal plane. Next, the nodes were interconnected with quadrilateral shaped elements and the elements were assembled to generate three-dimensional mesh representing the geometry of the model (Figure 4-4). The approximate middle transverse plane of the L3-L4 disc was taken as horizontal. A lordotic curvature of approximately 9° was simulated at the L4-L5 level of the FE model based on the anthropometric data. A lordotic curvature of 34° was simulated across the L1-S1 segment. The thoracic spine and pelvis joint finite element models were created following the same procedure.
Figure 4-4: Procedure used in the development and use of an FE model of thoracolumbar spine and pelvis.

4.2.3 Assembly

Following the development of the FE model geometry, all the meshed regions were assembled together in the ABAQUS (Dassault Systèmes, Simulia Inc., Providence, RI) software. Appropriate material properties were assigned to groups of elements separately representing the mechanical characteristics of each physiological component.

Vertebral Body and Posterior Elements

The following pictures provide different views of the meshed bony structures used in this study. The skull is made up of all tetrahedral elements. It consists of 109543 elements and 30617 nodes (Figure 4-5). The C1 vertebra consists of 6351 hexahedral elements and 8446 nodes. The C2 vertebra consists of 27996 hexahedral elements and 32649 nodes. The C3 vertebra consists of 19080 hexahedral elements and 22911 nodes. The C4 vertebra consists of 19363 hexahedral elements and 23206 nodes. The C5 vertebra consists of 25497 hexahedral elements and 30090 nodes. The C6 vertebra consists of 32503 hexahedral elements and 37631 nodes. The C7 vertebra consists of 33091 hexahedral elements and 38634 nodes (Figure 4-6).
Figure 4-5: Lateral view of the meshed skull.
Figure 4-6: Different views of C1 through C7 meshed vertebrae of the FE model.

Thoracic and lumbar spine vertebrae were also meshed with hexahedral elements. Figure 4-7 represents the vertebral body mesh of the thoracic and lumbar spine. Each vertebra in these regions consists of 6032 hexahedral elements and 7552 nodes respectively. Sacrum consists of 20088 elements and 24755 nodes.

Figure 4-7: Meshed T1 vertebra, L1 vertebra and Sacrum.
The vertebral body and posterior elements were defined using three-dimensional solid continuum hexahedral elements. The vertebral bodies have been modeled as a cancellous (porous) bone core surrounded by a 0.5 mm thick cortical (dense) bone shell. Appropriate isotropic material properties were defined for each respective region.

**Intervertebral Disc**

The intervertebral disc was modeled as the annulus fibrosus and nucleus pulposus. The annulus fibrosus was modeled as a composite solid ground substance, reinforced by embedded fibers. The ground substance was made up of 3D solid hexagonal elements. The REBAR option (ABAQUS, Simulia Inc., Providence, RI) was used to define fibers that were oriented at alternating angles ±30° to horizontal in lumbar discs and ±65° to vertical in the cervical spine. The “no compression” option was used for the REBAR elements such that they would transmit only tension. Fiber thickness and stiffness increased in the radial direction. The mean collagen fiber content of the annulus was of 16%. The nucleus pulposus was modeled with C3D8 hexagonal elements and the volume of the nucleus was considered to be approximately 50 percent of the disc (Figure 4-8) [94]. The nucleus pulposus was modeled as an incompressible hyper-elastic material [95]. The annulus fibrosis was modeled as hyper-elastic solid material [96].
Figure 4-8: Meshed C5-C6 intervertebral disc depicting nucleus pulposus (blue) and annulus fibrosis (grey).

A unique feature of the cervical intervertebral disc is the presence of fissures or clefts that run along the uncinate processes medially. These fissures, called Luschka’s joints or uncovertebral joints, appear in the annular lamellae. In the lower cervical of FE model, these Uncovertebral (Luschka's) joints were simulated using GAPUNI elements around the area of the uncinate processes in the lower cervical spine [97].
Facet (Apophyseal) joints

The mechanical influence of the facet joints plays a crucial role in determining the biomechanical outcome of a spine model. In the model described here, the facet joints were simulated using surface to surface sliding contact interactions. The facets were oriented at an inclination of 72° from the horizontal plane in the lumbar spine and approximately 45° in the cervical spine (Figure 4-9). In the upper cervical spine at C1-C2 level, the facets were oriented at 25° from the horizontal plane. The facets were oriented almost vertically in the thoracic spine region.
Figure 4-10: Lateral view of the C3-C4 right facet joint of the FE model.

An initial gap was specified between these elements based on the CT scan geometry. Force was transmitted with nonlinear contact property, which exponentially adjusts the force transfer as the gap is closed (Figure 4-11). Upon full contact, the joint assumes the same stiffness as the posterior elements. This feature was modeled using ABAQUS 6.11 version.
In addition to the facet joints, sliding articulations were incorporated into upper cervical spine in: right and left occipito-atlantal joints (C0-C1 level), odonto-atlantal joint (between the anterior aspect of the odontoid process and the posterior aspect of the anterior ring of the atlas), and the transverse ligament-odontoid process articulation (between the transverse ligament and the base of the posterior aspect of the odontoid process).

**Ligaments**

Ligaments were modeled using three-dimensional two-node truss elements (T3D2). The ligament sites, area and orientations were determined from the literature. Hypoelastic material properties were assigned to each ligament, which allowed laxity to be simulated. As such, the ligament provided little stability under minimally applied external loads. This phenomenon allowed the model to simulate the “neutral zone” of each segment. The hypoelastic material definition was given by specifying a variable Young’s modulus and Poisson’s ratio along with the strain invariants at the specified strain rate. Ligament material properties were taken from the literature and in house experimental data [97-101]. The defining elements were aligned along their respective ligament fiber orientation. The ligamentum flavum and the longitudinal ligaments experience a pre-stress at rest, while all other ligaments were assumed to be unstressed.
initially. The Young’s modulus for all the ligaments was defined over a range of strains to simulate their nonlinear stress-strain curve characteristics. Ligament properties are responsible for the nonlinear kinematic response of the spine model.

Ligaments modeled in the upper cervical spine were cruciform ligament, alar ligament, apical ligament, transverse ligament, accessory ligament, nuchal ligament, tectorial membrane, anterior longitudinal ligament, occipito-atlantal capsular ligament and atlanto-axial capsular ligaments. All the seven major ligaments were modeled in the rest of the spine, including Anterior Longitudinal Ligament (ALL), Posterior Longitudinal Ligament (PLL), Interspinous Ligament (ISL), Supraspinous Ligament (SSL), Capsular Ligament (CL), Ligamentum Flavum (LF).

The developed finite element model assembly is shown in the figure 4-12, which consists of human head, whole spine, ribcage and pelvis with their respective joint articulations. It consists a total number of 506642 nodes and 507037 elements.
Figure 4-12: Individual three dimensional finite element models of head-cervical spine, thoracic spine with ribcage and the lumbar-pelvis segments were assembled to create a complete three dimensional osseo-ligamentous head-pelvis finite element model.

The material properties used for modeling the human head-pelvis FE model are listed in the following Tables (4.1, 4.2, 4.3, 4.4, 4.5, 4.6). Isotropic linear, 4-node tetrahedral elements (C3D4) were assigned for both the cortical and cancellous regions of skull. Isotropic linear, 8-node hexahedral elements (C3D8) were assigned for all the regions of vertebrae.

Table 4.1: Material properties of bony structures in skull and spine [97-101].
<table>
<thead>
<tr>
<th>Element Name Group</th>
<th>Elastic Modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Bony Regions in Skull</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Skull-Cortical</td>
<td>15000</td>
<td>0.3</td>
</tr>
<tr>
<td>Skull-Cancellous</td>
<td>500</td>
<td>0.3</td>
</tr>
<tr>
<td><strong>Bony Regions in Cervical Spine</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Vertebrae-Cortical</td>
<td>10000</td>
<td>0.3</td>
</tr>
<tr>
<td>Vertebrae-Cancellous</td>
<td>450</td>
<td>0.25</td>
</tr>
<tr>
<td>Vertebrae-Posterior</td>
<td>3500</td>
<td>0.25</td>
</tr>
<tr>
<td><strong>Bony Regions in Thoracic and Lumbar Spine</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Vertebrae-Cortical</td>
<td>12000</td>
<td>0.3</td>
</tr>
<tr>
<td>Vertebrae-Cancellous</td>
<td>100</td>
<td>0.2</td>
</tr>
<tr>
<td>Vertebrae-Posterior Cortical</td>
<td>12000</td>
<td>0.3</td>
</tr>
<tr>
<td>Vertebrae-Posterior Cancellous</td>
<td>100</td>
<td>0.2</td>
</tr>
</tbody>
</table>

The intervertebral discs were modeled as solid 8-node hexahedral elements. Both the nucleus and annulus were modeled as hybrid elements (C3D8H). The nucleus pulposus was modeled as an incompressible hyper-elastic solid material. The annulus fibrosis was modeled as hyper-elastic solid material. A Mooney-Rivlin formulation was used to model the hyper-elasticity. According to ABAQUS documentation, the strain energy potential is defined by equation 4.1.

\[
U = C_{10}(\bar{I}_1 - 3) + C_{01}(\bar{I}_2 - 3) + \frac{1}{2} \left( J^{el} - 1 \right)^2 \tag{4.1}
\]
Table 4.2: Mooney-Rivlin constants for nucleus pulposus model [95].

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>( C_{10} )</td>
<td>0.12</td>
</tr>
<tr>
<td>( C_{01} )</td>
<td>0.03</td>
</tr>
<tr>
<td>( D_1 )</td>
<td>0</td>
</tr>
</tbody>
</table>

Table 4.3: Mooney-Rivlin constants for annulus ground material model [96].

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>( C_{10} )</td>
<td>0.7</td>
</tr>
<tr>
<td>( C_{01} )</td>
<td>0.2</td>
</tr>
</tbody>
</table>

Annulus fibers were modeled using REBAR elements with no compression option to transmit only tension with Young's Modulus varying between 357-550 MPa and Poisson's ratio of 0.3. Tables 4.4, 4.5, and 4.6 describe the material properties for ligaments in the upper cervical spine, lower cervical spine, thoracic spine and lumbar spine respectively.

Table 4.4: Upper cervical spine (C0-C2) ligament material properties [100].

<table>
<thead>
<tr>
<th>Ligament</th>
<th>Young’s Modulus (MPa)</th>
<th>Poisson’s</th>
<th>Cross-</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ligament</td>
<td>Young’s Modulus (MPa)</td>
<td>Poisson’s Ratio</td>
<td>Cross-Sectional Area (mm²)</td>
</tr>
<tr>
<td>--------------------------------------</td>
<td>--------------------------------</td>
<td>-----------------</td>
<td>--------------------------</td>
</tr>
<tr>
<td>Transverse</td>
<td>18</td>
<td>0.3</td>
<td>20</td>
</tr>
<tr>
<td>Superior/inferior cruciform</td>
<td>0.3 (80 %), 10.0 (80 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Alar</td>
<td>0.3 (90 %), 10.0 (90 %)</td>
<td>0.3</td>
<td>22</td>
</tr>
<tr>
<td>Apical</td>
<td>6.0 (90 %), 10.0 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Accessory</td>
<td>6.0 (90 %), 10.0 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Nuchal</td>
<td>12.0 (90 %), 20.0 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Tectoral membrane/ligament</td>
<td>6.3</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Anterior Longitudinal</td>
<td>12.0 (90 %), 20.0 (90 %)</td>
<td>0.3</td>
<td>6</td>
</tr>
<tr>
<td>Posterior C0-C1 capsular</td>
<td>6.0 (90 %), 10.0 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Anterior C0-C1 capsular</td>
<td>6.0 (90 %), 10.0 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Posterior C1-C2 capsular</td>
<td>6.0 (90 %), 10.0 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Anterior C1-C2 capsular</td>
<td>0.2 (90 %), 1.25 (90 %)</td>
<td>0.3</td>
<td>5</td>
</tr>
</tbody>
</table>

Table 4.5: Lower cervical spine (C2-C7) ligament material properties [97, 99].
Table 4.6: Thoraco-Lumbar spine (T1-S1) ligament material properties [101].

<table>
<thead>
<tr>
<th>Ligament</th>
<th>Young’s Modulus (MPa)</th>
<th>Poisson’s Ratio</th>
<th>Cross-Sectional Area (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior Longitudinal</td>
<td>7.8 (&lt;12%), 20.0 (&gt;12%)</td>
<td>0.3</td>
<td>74</td>
</tr>
<tr>
<td>Posterior Longitudinal</td>
<td>10.0 (&lt;11%), 20.0 (&gt;11%)</td>
<td>0.3</td>
<td>14.4</td>
</tr>
<tr>
<td>Ligamentum Flavum</td>
<td>15.0 (&lt;6.2%), 19.5 (&gt;6.2%)</td>
<td>0.3</td>
<td>40</td>
</tr>
<tr>
<td>Inter-transverse</td>
<td>10.0 (&lt;18%), 58.7 (&gt;18%)</td>
<td>0.3</td>
<td>1.8</td>
</tr>
<tr>
<td>Interspinous</td>
<td>10.0 (&lt;14%), 11.6 (&gt;14%)</td>
<td>0.3</td>
<td>40</td>
</tr>
<tr>
<td></td>
<td>Percentages</td>
<td>Value1</td>
<td>Value2</td>
</tr>
<tr>
<td>----------------</td>
<td>---------------------------</td>
<td>--------</td>
<td>--------</td>
</tr>
<tr>
<td>Supraspinous</td>
<td>8.0(&lt;20%), 15.0(&gt;20%)</td>
<td></td>
<td>30</td>
</tr>
<tr>
<td>Capsular</td>
<td>7.5(&lt;25%), 32.9(&gt;25%)</td>
<td>0.3</td>
<td>34</td>
</tr>
</tbody>
</table>
4.3 Mesh Convergence Study

Before finalizing the mesh sizing, it is necessary to perform a convergence study. To determine the appropriate levels for use in the new spine model, the convergence was completed using a spine model with simplified geometry (Figure 4-13). In this simplified system, the vertebrae and intervertebral disc were approximated as elliptical structures. The annulus fibrosus (AF) and nucleus pulposus (NP) created as distinct geometries, and the NP was defined as 50% of the total disc volume.

![Cross section of simplified convergence model assigned with mesh size 3mm.](image)

Figure 4-13: Cross section of simplified convergence model assigned with mesh size 3mm.

The seed size for meshing was varied to four sizes (0.5, 1, 2, and 3 millimeters) of hexahedral element type to examine this convergence. All models were examined under the same constraints and loading conditions. Material properties for all components were taken from the Table 4-1 and 4-2. Load was applied as 60% body weight to the surface of
the superior vertebra (316N), while the lower surface of the inferior vertebra was held fixed. Tie constraints were used between each adjacent component. Deformation was measured at the center of the intervertebral disc, near the surface of the superior vertebra. This displacement was compared between each model to determine the ideal mesh size to be used for the spine FE model development.

Figure 4-14: Convergence study of simplified spine model using linear hexahedral elements.
Based on the conversion study completed, a mesh seed size ranging from 0.5 mm to 1 mm is ideal for our spine computational simulation. Same behavior was even observed for the dynamic loading (Appendix A). Using a greater number of elements would result in a model taking too much time to run without significantly impacting results. Fewer elements would yield an inaccurate solution.

### 4.4 Model Validation

The goal of the validation of a finite element model is to verify whether the model can predict experimental data. The developed human head-pelvis FE model was validated under two different loading conditions: static and dynamic. Firstly, the model was tested for its ability to replicate the crucial response of physical experiments performed under static loading conditions from cervical, thoracic and lumbar spine in vitro experiments. These in vitro experimental data were obtained from our own laboratory experiments and the literature. Dynamic validation was performed by comparing the model predictions to full body sled tests involving human volunteers and cadaver data.

#### 4.4.1 Static Validation

Static validation was done under flexion, extension, lateral bending, and axial rotation. A follower load technique (Patwardhan et al) was used to apply a physiological axial preload in the first step and the respective moments were applied in the subsequent
steps [102]. These loads and boundary conditions for the FE model were obtained from the respective cervical spine, thoracic spine and lumbar spine cadaveric experiments which are described briefly in this section.

For the validation of the FE model, the segmental kinematics across the spine were compared with those obtained from a cadaveric experiment under similar loading scenarios. For this purpose six fresh osseoligamentous spine segments, L1-S1, were used. The sacrum was fixed in a rigid base, while a loading fixture was secured to L1 vertebral body. To determine the load-displacement behavior of the specimen, a set of three LED markers were attached to each vertebral body (Figure 4-15). An Opto-Trak 3020 motion tracking system (Northern Digital Inc, Waterloo, Ontario, Canada) was used to capture segmental motion across each specimen over a range of ±10 Nm in all three planes of motion. Data from the literature using similar testing methods for the cervical spine and thoracic spine specimens respectively were obtained for validation [103-106]. A bending moment of 1.5 Nm for the cervical spine and 4 Nm for thoracic spine was used in these experiments.
Figure 4-15: Experimental set up for static load pure moment testing. L1-S1 cadaver spine with Infrared LED markers attached shown on L1, L2, L3, L4, L5 and S1 vertebrae respectively.

The mean angular motion with standard deviation error bars were calculated for the respective spinal motion segments in all the above experiments.

The loading for the human head-pelvis FE model was applied in the four steps. The bottom of the femur was fixed in all degrees of freedom and rest of the model was free to move in all directions. In the first step, a follower load technique was used to apply a physiological axial preload of 73.6 N for cervical spine, 200 N for thoracic spine and 400N for lumbar spine using connector elements.
In the second step, a moment of 1.5 Nm was applied at a node which was coupled to the occiput region of the skull (Figure 4-16) which created a bending moment of 1.5 Nm throughout the entire FE model.

In the third step, as 1.5 Nm was already loaded onto the thoracic spine, a bending moment of 2.5 Nm was applied at a node which was coupled to the T1 vertebral body to match the thoracic spine *in vitro* experimental loading of 4 Nm [106].

In the subsequent step, a bending moment of 6 Nm was applied to the node coupled to the L1 vertebra, which together with the other step loadings matched the *in vitro* experimental loading of 10Nm on the lumbar spine.

This loading procedure was followed in order to simulate the pure moment loadings in all the physiological planes of motion (extension, flexion, right lateral bending, left lateral bending, right axial rotation and left axial rotation) on the respective spine regions. This loading enables the head-pelvis FE model predictions to compare with the data obtained from the above mentioned *in vitro* experiments [103-106]. The experimental angular displacement data was compared to the FE model predictions. In addition to the range of motion validation, the intra-discal pressure and facet load predictions were compared with the previous studies in the literature.
Figure 4-16: Different views of the human head-pelvis FE model depicting the loading and boundary conditions of the static validation method. Inferior surfaces of the left and femur were constrained in all the six degrees of freedom.

Black color line which follows the curvature of the spine indicates the follower type axial compressive preload application on the spine. Moment application on the occiput region of skull is shown with a red dot. Similarly, moments were applied on the thoracic and lumbar spines in the respective loading steps.
4.4.2 Dynamic Validation

Dynamic validation was done by simulating the experiments performed by Davidson *et al.* [107] and Ono *et al.* [108]. The data from these studies (Japan Automobile Research Institute) are the most comprehensive published data that can be used to validate the human body models. In these experiments, seven healthy male volunteers with mean age of 25 years (±4 years) were subjected to rear end impacts. A rigid wooden seat was used with no head restraint. This seat was mounted on a sled sliding on a long rail at an angle of 10° with the horizontal as shown in Figure 4-17.

![Test set up used to perform the rear end impact loadings on the human volunteers. (Adapted from [107])](image)

The volunteers were informed to assume normal seating posture with Frankfort plane of the head oriented horizontally. An oil damper was placed at the end of the rail...
which produces the impact when sled was allowed to slide along the rail and engaged to it at an impact speed of 8 km/h. This sled acceleration (Figure 4-18) was used as input to the FE model.

![Sled Accn](image)

Figure 4-18: Dynamic validation sled acceleration input to the FE model [107].

Ono et al [108] used the same set of volunteers and set up used by Davidson et al [107] to perform the rear end impacts. They reported the experimental data that can be utilized to validate the rear end impact thoracic spine deformations. This group used the same impact speed of 8 km/h and provided the vertebra rotations of one representative volunteer. These intervertebral rotations from Ono et al [108] and the global head, T1 kinematics from Davidson et al [107] were used to validate the FE model in the rear end impact loading scenario.

The mass properties used for the bony geometry are shown in the following table (Table 4.7). The densities of the cortical, cancellous, and posterior elements were all scaled to get the proper mass value. The moment of inertia is automatically accounted by
geometry and density values. Discs were added with viscoelastic material properties adapted from the literature (Table 4.8, 4.9) [123,124].

Table 4.7: Bony mass properties [87, 109].

<table>
<thead>
<tr>
<th>Level</th>
<th>Mass (Kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SKULL</td>
<td>3.53</td>
</tr>
<tr>
<td>C1</td>
<td>0.22</td>
</tr>
<tr>
<td>C2</td>
<td>0.25</td>
</tr>
<tr>
<td>C3</td>
<td>0.24</td>
</tr>
<tr>
<td>C4</td>
<td>0.23</td>
</tr>
<tr>
<td>C5</td>
<td>0.23</td>
</tr>
<tr>
<td>C6</td>
<td>0.24</td>
</tr>
<tr>
<td>C7</td>
<td>0.22</td>
</tr>
<tr>
<td>T1</td>
<td>0.76</td>
</tr>
<tr>
<td>T2</td>
<td>0.76</td>
</tr>
<tr>
<td>T3</td>
<td>0.97</td>
</tr>
<tr>
<td>T4</td>
<td>0.90</td>
</tr>
<tr>
<td>T5</td>
<td>0.90</td>
</tr>
<tr>
<td>T6</td>
<td>0.90</td>
</tr>
<tr>
<td>T7</td>
<td>0.97</td>
</tr>
<tr>
<td>T8</td>
<td>1.04</td>
</tr>
</tbody>
</table>
Table 4.8: Prony series constants for nucleus pulposus [123, 124].

<p>| | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>T9</td>
<td></td>
<td>1.11</td>
</tr>
<tr>
<td>T10</td>
<td></td>
<td>1.39</td>
</tr>
<tr>
<td>T11</td>
<td></td>
<td>1.46</td>
</tr>
<tr>
<td>T12</td>
<td></td>
<td>1.74</td>
</tr>
<tr>
<td>L1</td>
<td></td>
<td>1.67</td>
</tr>
<tr>
<td>L2</td>
<td></td>
<td>1.67</td>
</tr>
<tr>
<td>L3</td>
<td></td>
<td>1.60</td>
</tr>
<tr>
<td>L4</td>
<td></td>
<td>1.81</td>
</tr>
<tr>
<td>L5</td>
<td></td>
<td>1.81</td>
</tr>
</tbody>
</table>

Table 4.9: Prony series constants for Annulus Fibrosis [123, 124].

<p>| | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>g_i</td>
<td>k_i</td>
<td>τ_i</td>
</tr>
<tr>
<td>0.638</td>
<td>0.0</td>
<td>0.141</td>
</tr>
<tr>
<td>0.156</td>
<td>0.0</td>
<td>2.21</td>
</tr>
<tr>
<td>0.120</td>
<td>0.0</td>
<td>39.9</td>
</tr>
<tr>
<td>0.0383</td>
<td>0.0</td>
<td>266</td>
</tr>
<tr>
<td>0</td>
<td>0.0</td>
<td>500</td>
</tr>
</tbody>
</table>
A rigid seat without head restraint was created in ABAQUS (6.11v) and the human head-pelvis FE model was positioned in the normal seating posture with the Frankfort plane of the head oriented horizontally (Figure 4-19), similar to the above mentioned experiments. The normal seating posture for the FE model was based on the measurements from the literature [81,110,111]. Figure 4-20 depicts the angles of posterior face of vertebral body (clockwise is +ve) measurements in the FE model compared to the experimental measurements done on PMHS subjects (using lateral cervical spine X-ray) in the normal seating posture [81]. A surface to surface hard contact was defined between the seat and the FE model. These contacts were defined between inferior surface of the pelvis and the seat bottom, and posterior region of the FE model and seat back. We are not considering the effect of muscles on whiplash injury in this study considering the occupant to be unaware of the impact.

<p>| | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>0.000</td>
<td>0.300</td>
<td>100</td>
</tr>
<tr>
<td>0.361</td>
<td>0.149</td>
<td>1000</td>
</tr>
<tr>
<td>0.108</td>
<td>0.150</td>
<td>5000</td>
</tr>
</tbody>
</table>
Figure 4-19: Loading and boundary conditions for the dynamic validation.

Figure 4-20: Comparison of FE model cervical vertebra angles (angle of posterior face of the vertebral body - clockwise is +ve) compared to those measured from the PMHS subjects using lateral cervical spine X-rays in the normal seating condition [81].

In another study, Deng et al used six full body cadavers on which 26 rear end impacts were performed with accelerations ranging from 5 to 9.9g [79]. They used a
HYGE mini-sled to accelerate a full body cadaver seated in a custom made rigid seat, with or without headrest. The average age, weight, and height of the cadavers were 72.3 years, 59.4 kg and 165 cm with standard deviation of (20.5 years, 11.4 kg, and 4.6 cm) respectively. Based on the available data, an impact severity of 7g peak sled acceleration was chosen for the FE model simulation. The response of the model was compared to tests in which the sled peak acceleration was ±1g of the 7g impact. The 7g peak sled acceleration pulse (Figure 4-21) was the only input given to the FE model.

![Sled Acceleration](image)

Figure 4-21: Sled acceleration pulse from Deng et al rear end impact full body cadaveric experiments were used as input for the FE model rear end impact simulation [79].

### 4.5 Influence of Different Seating Postures

The first experiment was to test the influence of different initial occupant seating postures on the injury patterns due to rear end full body impacts. For this, a 3D model of
a standard seat with headrest was designed in SOLIDWORKS (Dassault Systems, Massachusetts, USA) from geometrical measurements of the car seat used in the dummy experiments performed at University of Toledo (Figure 4-22). The standard seat with headrest was meshed using 3-Matic, which created a solid tetrahedral mesh. The seat with headrest was then imported into ABAQUS and assembled with the 3D human head-pelvis FE model in a normal seating posture (as mentioned in the above section).

Material properties of the standard seat with head restraint were derived from the literature. The seat with the headrest is made up of polyurethane foam.

Figure 4-22: Rear end impact test set up showing Hybrid III dummy seated in a standard car seat with normal headrest at University of Toledo.
An Ogden hyper-foam model was used to define the standard seat with head restraint material property. These material properties for the foam were defined using coefficients and they were given respective visco-elastic properties [112]. The visco-elasticity was defined using Prony series parameters derived from the literature. The hyper-foam Ogden model is given by equation 4.2. The equation for a Prony series expansion is defined by ABAQUS by equation 4.3 [98].

\[
U = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^2} \left[ \hat{\lambda}_1^{\alpha_i} + \hat{\lambda}_2^{\alpha_i} + \hat{\lambda}_3^{\alpha_i} - 3 + \frac{1}{\beta_i} \left( (J^e)^{\alpha_i\beta_i} - 1 \right) \right],
\]

\[
g_R(t) = 1 - \sum_{i=1}^{N} \tilde{g}_i^p (1 - e^{-\frac{t}{\tau_i}}),
\]

Table 4.10: Ogden coefficients for the hyper-foam material [112].

<table>
<thead>
<tr>
<th>(\mu_i) (MPa)</th>
<th>(\alpha_i)</th>
<th>(\beta_i)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.164861</td>
<td>8.88413</td>
<td>0</td>
</tr>
<tr>
<td>2.3017E-05</td>
<td>-4.81798</td>
<td>0</td>
</tr>
</tbody>
</table>

Table 4.11: Prony series constants for the hyper-foam material [112].

<table>
<thead>
<tr>
<th>(g_i)</th>
<th>(k_i)</th>
<th>(\tau_i)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.3003</td>
<td>0</td>
<td>0.010014</td>
</tr>
<tr>
<td>0.1997</td>
<td>0</td>
<td>0.1002</td>
</tr>
</tbody>
</table>
Figure 4-23: Stress-strain curve of foam material property assigned for the standard seat and head rest models.

FE model positioned in the normal seating posture in the standard seat with headrest is shown in Figures 4-24 & 4-25. A surface to surface hard contact was defined between the seat and the FE model as described previously. In addition to that, another surface to surface hard contact was defined between the skull posterior region and headrest. This assembly was subjected to the rear end impact of 8 km/h with a peak acceleration of 4g. Acceleration profile was applied at a point coupled to seat (Figure 4-24).
Figure 4-24: Sagittal plane view of meshed FE model with standard seat and headrest in the normal seating posture.
Figure 4-25: Anterior view of FE model with standard seat and headrest in the normal seating posture.

It was clinically identified that the axial rotation of the head or inclined head prior to impact has an increasing whiplash injury risk following automotive rear impact [53, 122]. The goal of this experiment was to investigate the whiplash injury pattern when the occupant is in such out of occupant posture and subjected to a rear end impact. For this purpose, different out of occupant seating postures were identified based on the literature [111- 114]. These are normal seating posture with body in upright position and head in the normal position (Frankfort plane parallel to the horizontal); body forward posture
(BF) with body leaned forward and head in the normal position; head turned posture (HT) with body in upright position and head in the rotated position; combination of body forward and head turned postures (BF-HT); head flexed posture or inclined (HF) with body in upright position and head in 20 degrees flexion.

4.5.1 **Body forward seating posture**

Three body forward prone conditions with body leaned forward by 15°, 20° and 25° were simulated. This was done by modifying the normal seated posture FE model such that the angle between the vertical and line passing along T1-T7 spinous process was 15°, 20° and 25° respectively. In all the three new models, the head was maintained such that the Frankfort plane was parallel to the horizontal [81]. The head-neck vertebral angles of all the body forward postures were maintained similar to the normal seating posture. These models (Figures 4-26, 4-27, 4-28) were subjected to a rear end impact magnitude similar to the normal seating posture loading in the above section.
Figure 4-26: Body forward posture with head in Frankfort plane and body leaned forward by 15 degrees with respect to the vertical (BF-15).
Figure 4-27: Body forward posture - 20 degrees (BF-20).
4.5.2 Head turned seating posture

It was clinically identified that the axial rotation of the head prior to impact has an increasing whiplash injury risk following automotive rear impact [53]. To investigate the effect of this particular posture on the injury pattern, the normal seating posture was modified to develop three head turned (HT) models of 15°, 30° and 45° respectively with body in the normal seated posture. Modification was done by constraining the T2 vertebra and rotating the head by desired angle. The level by level rotations of the head
and cervical spine were in agreement with the in vivo axial rotation data [125, 126]. Figure 4-29 shows the comparison of FE and in vivo distribution of rotation at each vertebral level of the cervical spine at 40 degrees head rotated posture. The following figures (4-30, 4-31, 4-32) depict the three progressively head rotated seated postures. They were subjected to the same rear end impact magnitude and their responses were compared with the normal seating posture.

Figure 4-29: FE model rotation angles compared to in vivo data during 40 degree head rotated posture [125].
Figure 4-30: Head turned posture with the body in the upright position and head rotated to the right by 15 degrees (HT-15).
Figure 4-31: Head turned posture - 30 degrees (HT-30).

Figure 4-32: Head turned posture - 45 degrees (HT-45).
4.5.3 Body forward and head turned posture (BF-HT)

From the above two seating postures, a combination of body leaned forward by 25 degrees and head turned by 45 degrees posture was created (Figure 4-33) which was named as body forward head turned posture (BF-HT). It was subjected to 4g rear end impact.

![Figure 4-33: Body forward-Head turned posture (BF-HT).](image)

4.5.4 Head flexed or inclined seating posture

The normal seating posture was modified to create the head flexed seated posture by keeping the body in the upright position and inclining head forward by 20 degrees as shown in Figure 4-34. It was subjected to a rear end impact magnitude similar to the normal seating posture loading for comparing the FE responses.
Figure 4-34: Head inclined or flexed posture with body in upright position and head inclined forward by 20 degrees with respect to the vertical (HF).

The comparison of the FE responses (facet stresses, ligament strains and intervertebral disc pressures) were made amongst the normal seating posture, body forward posture (BF-20), head turned posture (HT-45), head flexed and body forward head turned (BF-HT) posture to identify the high risk out of occupant seating position.

4.6 Effect of Increased Acceleration Magnitudes

For the last part of this experiment, the goal was to identify the high risk out of occupant seating postures from the above section results and test the effect of increased impact severity on the whiplash injury mechanism. For this purpose, an acceleration
profile of 10g (Figure 4-35) was applied as input to the high risk out of occupant posture FE models. Standard seat with headrest was used for all the simulations.

Figure 4-35: Sled acceleration pulse from Deng et al rear end impact full body cadaveric experiments were used as input for the high risk out of occupant seated posture FE model rear end impact simulation [79].

4.7 Data Analyses

Intradiscal pressures, facet stresses, facet capsule strains, anterior ligament strains, forces at cervico-thoracic junction, head translations and T1 vertical translations were analyzed to compare the results of different out of occupant seated postures with the normal seating posture (Table . The intradiscal pressures were taken at the centroid of the nucleus elements. The max stress reading was taken and recorded at each increment of the step. The intra-discal pressure reading was taken to determine the chance of the disc
trauma during rear end whiplash. Maximum facet stress readings were taken to compare the risk for facet pinching. Facet strains and anterior cervical spine ligament strains were also recorded and compared between the various cases.

Table 4.12: Parameters computed from the FE models.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Intervertebral Disc</strong></td>
<td></td>
</tr>
<tr>
<td>Maximum Intradiscal Pressure (IDP)</td>
<td>Mega Pascal (MPa)</td>
</tr>
<tr>
<td><strong>Anterior Ligaments</strong></td>
<td></td>
</tr>
<tr>
<td>Anterior Longitudinal Ligament Strain (ALL Strain)</td>
<td>(mm/mm)</td>
</tr>
<tr>
<td>Alar Ligament Strain (Alar Strain)</td>
<td>(mm/mm)</td>
</tr>
<tr>
<td><strong>Facet Joint</strong></td>
<td></td>
</tr>
<tr>
<td>Capsular Ligament Strain (CL Strain)</td>
<td>(mm/mm)</td>
</tr>
<tr>
<td>Facet Stress</td>
<td>Mega Pascal (MPa)</td>
</tr>
<tr>
<td><strong>Cervico-Thoracic Junction (C7-T1)</strong></td>
<td></td>
</tr>
<tr>
<td>Axial Force</td>
<td>Newton (N)</td>
</tr>
<tr>
<td>Shear Force</td>
<td>Newton (N)</td>
</tr>
<tr>
<td><strong>Translations</strong></td>
<td></td>
</tr>
<tr>
<td>Head Horizontal Translation</td>
<td>Millimeter (mm)</td>
</tr>
<tr>
<td>T1 Vertical Translation</td>
<td>Millimeter (mm)</td>
</tr>
</tbody>
</table>
Chapter 5

Results

5.1 Introduction

The first step of this project was to validate the FE model under static and dynamic loading conditions. Both loading conditions and data were obtained from the literature as described in the previous chapter. The first section presents these validation results. The main objective of this project was to study the effect of non-typical seating postures on the injury outcome compared to the normal seating posture. The second section in this chapter proceeds with the analysis of these results and identifying the high risk out of occupant postures. Finally, the results of increased impact severity on the identified high risk postures will be presented.

5.2 Validation Results

For the validation under static loading, the FE model range of motion predictions were compared to the experimental data. Cervical spine range of motion data at 1.5 Nm
was compared with Leahy et al and Faizan et al as shown in the figures (5-1 and 5-2) [103, 105]. Thoracic spine range of motion data at 4 Nm was compared with Brocc et al (Figure 5-3 to 5-6). Data from our own experiments were used to compare the lumbar spine range of motion at 10 Nm (Figures 5-7 to 5-10). FE model predictions in all the three planes of physiological loading fell within one standard deviation of the experimental data. We can conclude that this model behaves accordingly under static loading conditions.

Figure 5-1: Comparison of the FE model and experiment segmental kinematics in the upper cervical spine at 1.5 Nm [103].
Figure 5-2: Comparison of the FE model and experiment segmental kinematics in the upper cervical spine at 1.5 Nm [105].

Figure 5-3: Comparison of the FE model and experiment segmental kinematics across T4-L1 spine subjected to 4 Nm bending moment in extension.
Figure 5-4: Comparison of the FE model and experiment segmental kinematics across T4-L1 spine subjected to 4 Nm bending moment in flexion.
Figure 5-5: Comparison of the FE model and experiment segmental kinematics across T4-L1 spine subjected to 4 Nm bending moment in right bending.

Figure 5-6: Comparison of the FE model and experiment segmental kinematics across T4-L1 spine subjected to 4 Nm bending moment in left rotation.
Figure 5-7: Comparison of the FE model and experiment segmental kinematics across L1-S1 spine subjected to 10 Nm bending moment in extension.
Figure 5-8: Comparison of the FE model and experiment segmental kinematics across L1-S1 spine subjected to 10 Nm bending moment in flexion.

![Right Bending Graph](image)

Figure 5-9: Comparison of the FE model and experiment segmental kinematics across L1-S1 spine subjected to 10 Nm bending moment in right bending.
Figure 5-10: Comparison of the FE model and experiment segmental kinematics across L1-S1 spine subjected to 10 Nm bending moment in left rotation.

We can conclude that this model behaves close to experiments under static loading conditions. The next step was to validate the model under dynamic loading conditions. The experimental data was derived from Davidson et al [107]. A detailed set of FE model responses that compare the responses of the experiment are displayed in Figures 5-11 to 5-13. The complete set of volunteer responses are provided in the reference [107] and are shown here by the upper and lower bounds. For the model validation, several responses have been analyzed. The head rotation angle, T1 rotation angle and the head rotation with respect to T1 were plotted.
Figure 5-11: Comparison of head rotational response to a 4g rear end impact human volunteer response (extension is negative).
Figure 5-12: Comparison of T1 rotational response to a 4g rear end impact human volunteer response (extension is negative).
Figure 5-13: Comparison of head with respect to T1 rotational response to a 4g rear end impact human volunteer response (extension is negative).

Overall, the FE model responses were in good agreement with the volunteer data and were within the upper and lower limits. Ono et al used the same set of volunteers and set up with the same impact speed of 8 km/h and provided the vertebra rotations of one representative volunteer. FE model responses were compared to these reported vertebral rotations as displayed in the figures 5-14 to 5-21.

Figure 5-14: C6 vertebral rotation due to 8 km/h rear end impact. (extension is positive)
Figure 5-15: C5 vertebral rotation due to 8 km/h rear end impact. (extension is positive)

Figure 5-16: C4 vertebral rotation due to 8 km/h rear end impact. (extension is positive)
Figure 5-17: C3 vertebral rotation due to 8 km/h rear end impact. (extension is positive)

Figure 5-18: C2 vertebral rotation due to 8 km/h rear end impact. (extension is positive)
Figure 5-19: T1 vertebral rotation due to 8 km/h rear end impact. (extension is positive)

Figure 5-20: T2 vertebral rotation due to 8 km/h rear end impact. (extension is positive)
The individual spinal vertebra rotation angles of the FE model were in the same trend with the experimental rotational angles measured by Ono et al [108]. The model went through a transition from S-shaped curvature to the full extension phase (Figure 5-22) as observed in the volunteer experiments.
In the Initial response phase (0-50 msec), the upper thoracic spine was pushed by the seat back. The next phase (50-100 msec) involved the neck showing an S-shape formation, and the head a slight initial flexion. Initial extension phase (100-150 msec) began with the head moving into extension where as the neck was already in the extension phase. After 150 ms, head continues to extend more and the cervical vertebrae...
maintained the same extension alignment to form the maximum extension phase (150-200ms).

In addition to validating the global response of the head and T1 in the model, it is also valuable to validate the relative vertebral rotations at each spinal level with appropriate validation corridors. If the intervertebral motions in the model are accurate, the soft tissue strains predicted by the model will also be accurate [115].

For further validation, the FE model responses (Figure 5-23 to 5-27) were compared with the data obtained from the full body cadaveric rear end impact experiment by Deng B et al [79]. At the same time, the comparison was also done with the isolated head-neck FE model by Fice et al. The grey lines show the experimental response corridors.
Figure 5-23: C1-C2 rotation of the FE model during 7g rear end impact. (Extension is positive)

Figure 5-24: C2-C3 rotation of the FE model during 7g rear end impact. (Extension is positive)
Figure 5-25: C3-C4 rotation of the FE model during 7g rear end impact. (Extension is positive)
During the early stages of the impact (<150 msec), the cervical spine segments (C1-C2, C2-C3, C3-C4, C4-C5) in the cadavers went into relative flexion and then the extension motion followed. This trend was very well observed in our FE model responses compared to the isolated head neck FE model used by Fice et al [115]. The peak flexion angle predicted by their model was on the low end of the experimental results whereas our FE model predicted the peak flexion angle reasonably well within the corridors.
Overall, FE model was in good agreement with the global kinematics and the intervertebral rotations of the experimental studies.

5.3 Test Results

Once the model was determined to be validated under static and dynamic loading, it was used to test the effect of out of occupant seating postures on the outcome of whiplash injury. Intradiscal pressures, facet stresses, facet capsule strains, and anterior ligament strains were analyzed to compare the results of different out of occupant seated postures with the normal seating posture. To enable the comparison between different occupant seating postures, the stresses and strains were normalized to consider the effect of only rear end impact. The effects of the change in posture on ligament strains are discussed in the next chapter. The results of the rear end impact with occupant in the normal seating posture were analyzed first. Figure 5-28 depicts the motion sequence of the FE model with standard seat and headrest, in the normal seating posture during the 4g rear end impact.

For the first 65 ms after the impact, the head remained in its baseline position (Latency phase) after which the translation motion of head was observed until 85 ms without any rotational component (Translation phase). In this phase, head translated posteriorly with respect to T1 by 16 mm which was computed at the head center of gravity.
Combined translation with rotation was observed between 85-140 ms (Extension phase). A total head translation of 41 mm with respect to T1 was observed by the end of this phase. Head to head rest contact also occurred in the same phase. After 140 ms
(rebound phase), the head started to rebound and return to its original position. The kinematic response of the FE model was in trend and agree with the motion sequence of the tested volunteers by the time of occurrence of motion as shown in the Table 3-3 [59].

During the early stages of this rear end impact, the lower cervical vertebrae began to extend earlier and faster than the upper cervical vertebrae due to which a slight “S” shape curvature was created. This S-shape curvature lasted about 100 msecs.

The intradiscal pressure (IDP) variation for the complete sequence of motion was plotted and displayed in the Figure 5-29. A peak IDP of 1.4 MPa was observed for the C6-C7 level which had the maximum value compared to other levels. This peak value was achieved at 120 msecs during the extension phase. This peak IDP occurred 80 msecs before the muscle activation time since muscles are said not to be activated before 200 msecs.
Figure 5-29: Intradiscal pressure variation in the cervical spine during 4g rear end impact.

### 5.3.1 Influence of Body Forward Seating Posture

Four different positions were simulated to test the effect of body forward posture on the whiplash injury outcome. Normal seating posture along with three other body forward postures (BF-15, BF-20 and BF-25) were subjected to the same rear end impact loading of 8 km/h. Results showed as the body forward position increased, the range of motion increased. Figure 5-30 & 5-31 show global head and T1 rotation angles that were compared among all the four cases. BF-25 had the highest global head and T1 rotations compared to other iterations. In all the three body forward posture cases, there was a delay in the head to headrest contact occurrence compared to the Normal posture.
Figure 5-30: Comparison of global head rotation angle as a function of increasing body forward posture during the 4g rear end impact.
Figure 5-31: Comparison of global T1 rotation angle as a function of increasing body forward posture during the 4g rear end impact.

Body forward posture (BF-25) resulted in the highest intra-discal pressures at all levels (Figure 5-32). As the body forward position increased, there was an increase in the intradiscal pressure values. The C6-C7 disc had the highest pressures in all of the cases in the lower cervical spine. This is due to the greater relative motion at C6-C7 level when compared to the upper levels. A peak value of IDP close to 1.8 MPa was observed at C6-C7 level in the BF-25 posture. There was a percentage increase of 7%, 22% and 30% in the C6-C7 level intradiscal pressure of BF-15, BF-20 and BF-25 respectively compared
to the normal seating posture. The next highest pressures were found at the C5-C6 level in all of the cases. The increase in the intradiscal pressures were greater between BF-15 and BF-20 positions. The rise was minimal in between BF-20 and BF-25 positions.

Figure 5-32: Comparison of lower cervical spine IDP values in FE models with different body forward seating postures subjected to the same rear end impact loading.

Peak values of IDP were observed in the time frame of extension phase, where head and neck both were in extension in all the body forward cases.
Figure 5-33 shows the peak stresses on the bony surfaces of facet joints taken at the lower cervical spine for all the cases during 4g rear end impact. As the body forward position increased, there was an increase in the facet stress values. BF-25 position showed consistently greater peak facet stresses. The C6-C7 facets showed the greatest peak facet stresses in all the cases due to the higher range of motion at that level. BF-25 position showed a peak facet stress of approximately 2.75 MPa at the C6-C7 level. There was a percentage increase of 13%, 25% and 48.1% in the C6-C7 level facet stresses of BF-15, BF-20 and BF-25 respectively compared to the normal seating posture. The next highest facet stresses were observed at C5-C6 level with a peak value of 1.8 MPa in the BF-25 position followed by C4-C5 level. The increase in facet stresses was greater between BF-15 and BF-20 positions similar to the trend observed for IDP values.
Figure 5-33: Comparison of lower cervical spine facet stresses in FE models with different body forward seating postures subjected to the same rear end impact loading.

Peak values of facet stresses were observed during the formation of s-shape curvature in the upper cervical spine and during the extension phase in the lower cervical spine. The increase in slope at C6-C7 and C5-C6 compared to other levels was due to the significant increase in the relative motion at those levels during body forward postures.

Figure 5-34 shows the peak facet capsular strains in the cervical spine when compared to the varying body forward seating postures. As the body forward position
increased, there was an increase in the facet capsular strain values. C3-C4 facet capsule experienced the greatest amount of peak strains in all of the cases, followed by C2-C3 and then C4-C5. BF-25 had a peak facet capsule strain of approximately 0.154 mm/mm at C3-C4 level. At this level, there was a percentage increase of 6%, 13.1% and 13.2% in facet capsular strains of BF-15, BF-20 and BF-25 respectively compared to the normal seating posture. The increase in facet capsular strains were greater between BF-15 and BF-20 positions.

![Capsular Strain](image.png)

Figure 5-34: Comparison of cervical spine facet capsular strains in FE models with different body forward seating postures subjected to the same rear end impact loading.

The peak capsular strains occurred during the formation of S-shaped curvature where upper cervical spine was in flexion compared to the lower cervical spine. This
explains the reason for the presence of higher capsular ligament strains in the upper cervical levels.

Figure 5-35 shows the peak anterior longitudinal ligament (ALL) strains in the cervical spine when compared with varying body forward seating postures. The ALL at C6-C7 showed the greatest peak ligament strains in BF-20 and BF-25 positions, followed by ALL at C5-C6 whereas it was the opposite in Normal and BF-15 positions. In all the cases, ALL at C4-C5 followed in depicting greatest peak ligament strains after ALL at C5-C6 and C6-C7 levels. The model with BF-25 position showed a peak ALL strain of approximately 0.095 mm/mm at ALL 67. There was a percentage increase of 8%, 32.5% and 34.1% in the C6-C7 level ALL strains of BF-15, BF-20 and BF-25 respectively compared to the normal seating posture. As the body forward position increased, there was an increase in the ALL strain values. The increase in these strains was greater between BF-15 and BF-20 positions.
Figure 5-35: Comparison of cervical spine ALL strains in FE models with different body forward seating postures subjected to the same rear end impact loading.

Peak values of ALL strain were observed in the time frame of extension phase, where head and neck both were in extension in all the cases.

5.3.2 **Influence of Head Turned Seating Posture**

The next test was to investigate the effect of head turned posture on the whiplash injury outcome. Normal seating posture along with three other head turned postures (HT-15, HT-30 and HT-45) were subjected to the same rear end impact loading of 8 km/h. The same parameters were used to study and analyze. Results showed as the head turned position increased, the range of motion increased. The head-headrest contact in the case of HT-45 occurred after 15 ms of the normal seating posture. Figure 5-36 shows the peak ALAR ligament strains in the cervical spine when compared with varying head turned seating postures. As the head turned position increased, there was an increase in the ALAR strain values. The model with HT-45 position showed a peak ALAR strain of approximately 0.48 mm/mm. These peak strains were observed in the time frame of extension phase when both head and neck were in extension. There was a percentage increase of 16%, 19% and 30% in ALAR strains with HT-15, HT-30 and HT-45 respectively when compared to the normal seating posture. The increase in these strains was greater between HT-30 and HT-45 positions.
Figure 5-36: Comparison of cervical spine ALAR strains in FE models with different head turned seating postures subjected to the same rear end impact loading.

Head turned posture (HT-45) resulted in the highest intra-discal pressures at all levels (Figure 5-37). As the head turned position increased, there was an increase in the intradiscal pressure values. The C6-C7 disc had the highest pressures in all of the cases in the lower cervical spine. A Peak value of IDP close to 1.8 MPa was observed at C6-C7 level in the HT-45 posture. This peak value was observed at C6-C7 level due to the greater relative motion at this level compared to other levels in all the cases. There was a percentage increase of 5%, 13% and 26% in the C6-C7 level intradiscal pressure of HT-15, HT-30 and HT-45 respectively compared to the normal seating posture. The next highest pressures were found at the C5-C6 level in all the cases. The increase in all the intradiscal pressures was greater between HT-30 and HT-45 positions.
Figure 5-37: Comparison of lower cervical spine IDP values in FE models with different head turned seating postures subjected to the same rear end impact loading.

Peak values of IDP were observed in the time frame of extension phase, where head and neck both were in extension in all the cases.

Figure 5-38 shows the peak facet stresses taken at the lower cervical spine for all the cases during 4g rear end impact. As the head turned position increased, there was an increase in the facet stress values. HT-45 position showed consistently greater peak facet stresses. The highest facet stresses were observed at C6-C7 level with a peak value of 3 MPa in the HT-45 position followed by C5-C6 level. The increase in all the facet stresses was greater between HT-30 and HT-45 positions.
Figure 5-38: Comparison of lower cervical spine facet stresses in FE models with different head turned seating postures subjected to the same rear end impact loading.

Peak values of facet stresses were observed during the extension phase time frame where both head and neck were in extension.

Head turned posture (HT-45) resulted in the highest ALL strains at all levels (Figure 5-39) when compared with varying head turned seating postures. The ALL at C6-C7 showed the highest peak ligament strains in HT-15, HT-30 and HT-45 positions, whereas ALL at C5-C6 depicted the highest peak ligament strains in Normal seating posture. Overall, the lower cervical spine motion segments (C5-C6, C6-C7 and C4-C5) had the higher peak ALL strains when compared to the upper segments. The model with HT-45 position showed a peak ALL strain of approximately 0.1 mm/mm at ALL 67. There was a percentage increase of 14%, 28% and 51% in the C6-C7 level ALL strains of
HT-15, HT-30 and HT-45 respectively when compared to the normal seating posture. As the head turned position increased, there was an increase in the ALL strain values. The increase in these strains was greater between HT-30 and HT-45 positions.

Figure 5-39: Comparison of ALL strains in FE models with different head turned seating postures subjected to the same rear end impact loading.

Peak values of ALL strain were observed in the time frame of extension phase, where head and neck both were in extension in all the cases.
Figure 5-40 shows the peak facet capsular strain in the cervical spine when compared to the varying head turned seating postures. As the head turned position increased, there was an increase in the facet capsular strain values. C3-C4 facet capsule experienced the highest amount of peak strain in all of the cases, followed by C2-C3 and then C4-C5. HT-45 had a peak facet capsule strain of approximately 0.25 mm/mm at C3-C4 level. At this level, there was a percentage increase of 10%, 63% and 83% in facet capsular strains of HT-15, HT-30 and HT-45 respectively when compared to the normal seating posture. The increase in facet capsular strains was greater between HT-30 and HT-45 positions.

![Capsular Strain](image)

**Figure 5-40**: Comparison of cervical spine facet capsular ligament strains in FE models with different head turned seating postures subjected to the same rear end impact loading.
The peak capsular strains occurred during the formation of S-shaped curvature where upper cervical spine was in flexion compared to the lower. This explains the reason for the presence of higher capsular ligament strains in the upper cervical levels.

5.3.3 **Comparison of Out of Occupant Seating postures**

Five different positions were compared to test the effect of out of occupant seated postures on the increase in risk of whiplash injury. The postures compared were: Normal seating posture, Head flexed posture (HF), Head turned posture (HT-45), body forward posture (BF-25) and the body forward-head turned posture (BF-HT). Same parameters were used to compare these results.

The vertical upward motion of the C6 vertebrae due to the thoracic ramping was quantified and compared in all the cases (Table 5.1). Also, the posterior horizontal translation of head with respect to T1 was computed in all the cases. Highest vertical ramping was observed for the postures with body forward position. Increase in posterior head translation with respect to T1 was observed in all the cases. The largest increase was observed in the BF-HT posture followed by BF-25 and HT-45 postures.

Table 5.1: Comparison of C6 vertical translation and posterior head horizontal translation with respect to T1 for different occupant seating postures.

<table>
<thead>
<tr>
<th>Translations (mm)</th>
<th>Normal</th>
<th>BF-25</th>
<th>HT-45</th>
<th>HF</th>
<th>BF-HT</th>
</tr>
</thead>
<tbody>
<tr>
<td>C6 Vertical</td>
<td>11</td>
<td>23</td>
<td>11</td>
<td>11</td>
<td>23</td>
</tr>
</tbody>
</table>
The peak intradiscal pressure was compared and displayed in Figure 5-41 for all the cases. At all the levels, all the four out of occupant seated postures had increased intradiscal pressure values when compared to the normal seating posture. The greatest peak intradiscal pressure for normal seating posture was observed at C6-C7 level. BF-HT posture had the highest peak intradiscal pressure value followed by HT-45 posture and BF-25 posture at this C6-C7 level. A Peak value of IDP close to 2.1 MPa was observed at C6-C7 level for the BF-HT posture. The next highest peak intradiscal pressure was found at C5-C6 level for the BF-HT posture followed by BF-25 posture. At C4-C5 level, a similar trend to the one observed at C6-C7 level was noticed. At C3-C4 level, BF-HT posture had the highest IDP value followed by head turned posture. It can be observed that body forward position, head turned position and the combination of both (BF-HT) have potential high risk on the intervertebral discs of lower cervical spine compared to head flex posture.
Figure 5-41: Comparison of intradiscal pressure in FE models with different out of occupant seating postures subjected to the same rear end impact loading.

Peak facet stresses for the out of occupant seated postures were analyzed and displayed in the Figure 5-42. In all the levels of the lower cervical spine, BF-HT posture, body forward posture and head turned posture consistently showed higher peak facet stresses compared to the normal seating posture whereas head flexed posture had similar facet stress levels to that of normal in C3-C4 and C4-C5 levels. BF-HT posture had the highest peak facet stress value of 3.2 MPa at C6-C7 level followed by BF-25 and HT-45 postures. In other levels, BF-HT posture had the highest peak facet stress values when compared to body forward, head turned, head flex posture and the normal seating posture.
Thus, head turned posture, body forward posture and the combination (BF-HT) have potential high risk of facet joint pinching.

Figure 5-42: Comparison of facet stresses in FE models with different out of occupant seating postures subjected to the same rear end impact loading.

Figure 5-43 shows the peak facet capsular strain in the cervical spine when compared to the varying out of occupant seating postures. C3-C4 facet capsule in the BF-HT posture experienced the highest amount of peak strain in all of the cases followed by C2-C3 and then C4-C5. A peak facet capsule strain of approximately 0.3 mm/mm was observed at C3-C4 level. Head turned posture had the next highest peak strain after BF-HT posture at all the levels except C6-C7 where body forward posture had the next highest value. The potential risk increased in facet capsular strains when the body was subjected to head turned, body forward and the combination of the two postures.
Figure 5-43: Comparison of facet capsular strain in FE models with different out of occupant seating postures subjected to the same rear end impact loading.

Figure 5-44 displays the peak anterior longitudinal ligament (ALL) strains in the when compared with varying out of occupant seating postures. The ALL at C6-C7 showed the highest peak ligament strains in the BF-HT posture (0.12 mm/mm), followed by ALL at the same level in head turned posture (0.11 mm/mm). In all the levels, BF-HT posture had the highest ALL strains compared to all other cases. In C2-C3, C4-C5 and C5-C6 levels, body forward posture depicted next highest peak ligament strains compared to other cases whereas head turned posture had the next greatest ALL strains at
C3-C4 and C6-C7 levels. The increase in potential risk of ALL injury was greater for head turned, body forward and the combination of both the postures.

Figure 5-44: Comparison of anterior ligament strain (ALL) in FE models with different out of occupant seating postures subjected to the same rear end impact loading.

Alar ligament strains were computed and displayed for all the out of occupant seated postures (Figure 5-45).
Figure 5-45: Comparison of ALAR ligament strain in FE models with different out of occupant seating postures subjected to the same rear end impact loading.

BF-HT posture had the highest peak ALAR ligament strain compared to all the cases followed by head turned posture. The model with BF-HT posture and head turned position showed peak ALAR strains of approximately 0.69 mm/mm and 0.48 mm/mm respectively. There was a percentage increase of 109%, 48% and 17% in ALAR strains with BF-HT, HT-45 and BF-25 respectively compared to the normal seating posture. A 5% percentage decrease in ALAR strain was observed in HF position when compared to that of normal posture. This indicates that the BF-HT and head turned positions are at higher risk of ALAR ligament injury.

### 5.3.4 Influence of Increased Impact Acceleration

It was evident from the above section that body forward, head turned and the combination of the two postures (BF-HT) had increased risk of injury when compared to other cases. Hence, these out of occupant seated postures were subjected to increased severity of the rear end impacts. The objective was to study the effect of increased impact severity on the whiplash injury outcome due to these high risk out of occupant seating postures. A 10g rear end impact loading was used for this purpose. Intradiscal pressure, facet capsular ligament strains, and anterior ligament strains were analyzed.
Figure 5-46 shows the intradiscal pressure responses of body forward posture with the 10g impact compared to the lower acceleration magnitude (4g). In all the cervical spine levels, there was an increase in the intradiscal pressure values with the increase in the impact acceleration. The highest peak intradiscal pressure of 5.36 MPa was observed in the C4-C5 level for the 10g impact while peak value of 2.6 MPa was observed in C6-C7 level for the 4g impact.

Figure 5-46: Intradiscal pressures for varying acceleration magnitudes with BF posture.

The intradiscal pressure responses of the head turned posture (HT-45) with the 10g impact compared to the lower acceleration magnitude (4g) is shown in Figure 5-47.
There was an increase in the intradiscal pressure values with the increase in the impact severity. C5-C6 level intervertebral disc had the highest peak IDP of 4.1 MPa during 4g whereas C4-C5 had the highest peak pressure of 5.3 MPa during 10g impact acceleration.

The intradiscal pressure responses of the body forward head turned posture (BF-HT) with the 10g impact compared to the lower acceleration magnitude (4g) is shown in Figure 5-48.
There was an increase in the intradiscal pressure values with the increase in the impact severity. C6-C7 level intervertebral disc had the highest peak IDP of 2.1 MPa during 4g whereas C4-C5 had the highest peak pressure of 5.6 MPa during 10g impact acceleration. Overall, at 10g impact acceleration, body forward head turned position (BF-HT) had the higher IDP values when compared to that other two positions.

Capsular ligament strains were plotted and displayed (Figure 5-49) for the body forward seating posture with varying impact acceleration magnitudes. With the increase in the impact acceleration magnitude, the peak capsular ligament strains increased for all the levels of the lower cervical spine. A highest peak capsular strain of 0.43 mm/mm was
observed at C3-C4 during the 10g impact whereas 0.15 mm/mm peak strain was observed at the same level during 4g impact. The next highest strain was observed in both the impact acceleration levels at C2-C3 level.

Figure 5-49: Capsular strains for varying acceleration magnitudes with BF posture.

A similar trend of increased capsular strains with the increase in the impact severity was observed with the head turned posture (Figure 5-50).
Figure 5-50: Capsular strains for varying acceleration magnitudes with HT posture.

A highest peak capsular strain of 0.31 mm/mm was observed at C3-C4 during the 10g impact acceleration and a peak capsular strain of 0.24 mm/mm was observed at the same level. At 10g impact, the next highest peak strain was observed at C5-C6 level followed by C2-C3. C6-C7 had the lowest peak strain levels for both 4g and 10g impacts.
Capsular ligament strains were plotted and displayed (Figure 5-51) for the body forward head turned (BF-HT) seating posture with varying impact acceleration magnitudes. With the increase in the impact acceleration magnitude, the peak capsular ligament strains increased for all the levels of the lower cervical spine. A highest peak capsular strain of 0.44 mm/mm was observed at C3-C4 during the 10g impact whereas 0.29 mm/mm peak strain was observed at the same level during 4g impact. The next highest strain was observed in both the impact acceleration levels at C2-C3 level. Overall, body forward head turned posture had the highest peak capsular ligament strain compared to the body forward and head turned postures during the 10g impact.

For both the body forward and head turned postures, anterior longitudinal ligament (ALL) strains were computed and displayed (Figures 5-52 & 5-53). With the
body forward posture, highest peak ALL strain was observed at C5-C6 level followed by C6-C7 level for the 10g impact whereas it was vice versa during 4g impact. The highest peak strain of ALL at C5-C6 was close to 0.12 mm/mm.

![ALL-Strain](image)

Figure 5-52: ALL ligament strains for varying acceleration magnitudes with BF posture.

For the head turned posture, highest peak ALL strain was noticed at C6-C7 level followed by C5-C6 level during 10g impact and the same trend was observed even during the 4g impact. Highest peak strain value of 0.17 mm/mm during 10g and 0.11 mm/mm during 4g were observed at C6-C7 level.
Figure 5-53: ALL ligament strains for varying acceleration magnitudes with HT posture.

Figure 5-54: ALL ligament strains for varying acceleration magnitudes with BF-HT posture.
For the body forward head turned posture (BF-HT), anterior longitudinal ligament (ALL) strains were computed and displayed (Figure 5-54). The highest peak ALL strain was observed at C4-C5 level followed by C5-C6 level for the 10g impact. The highest peak strain of ALL at C4-C5 was close to 0.13 mm/mm.

For head turned and BF-HT postures, ALAR ligament strains were also computed and plotted (Figures 5-55 & Figure 5-56). ALAR ligament strains increased with the increase in impact severity with the head turned posture. A highest peak strain of 1.5 mm/mm was observed with the head turned posture during 10g impact and peak strain of 0.48 mm/mm was noticed during 4g.

Figure 5-55: Alar ligament strains for varying acceleration magnitudes with HT posture.

ALAR ligament strains increased with the increase in impact severity with the BF-HT posture. A highest peak strain of 1.6 mm/mm was observed with the body
forward head turned posture during 10g impact and peak strain of 0.68 mm/mm was noticed during 4g.

![Figure 5-56: Alar ligament strains for varying acceleration magnitudes with BF-HT posture.](image)

Overall, combination of body forward and head turned posture had the highest peak alar strains compared to the head turned and body forward postures at 10g impact.
Chapter 6

Discussion

6.1 Overview

A detailed human head-pelvis FE model was developed and validated in static loading conditions (Flexion, extension, lateral bending and axial rotation) based on the kinematics. The model was also validated based on global and intervertebral kinematics for different rear end impact crash scenarios. The main focus was to build the model with accurate geometry and formulate it with the experimentally obtained material properties, and no aspect of the model was modified to improve the fit to a specific data set. The model was then used to investigate intervertebral disc stresses and ligament strain parameters when the occupant is subjected to an out of occupant seating posture during rear impacts. Finally, the influence of increased impact severity on the parameters quantifying the injury outcome (disc stresses and ligament strains) during the rear end impact was investigated.
6.2 Model Development

In this thesis a detailed 3D osseoligamentous human head-pelvis FE model has been developed. The model consists of a skull, vertebrae, viscoelastic discs, frictionless facet joints, nonlinear ligaments, ribcage and pelvis.

HUMOS and THUMS are the only two full body FE models existing in the literature. In HUMOS full body model, the T1 and C7 vertebrae were considered to be rigid bodies connected with a numerical joint (no elements for disc and facet joint) to model the T1–C7 physical joint which do not allow the study of potential injury on or around cervico-thoracic junction. Moreover, it was not validated for the rear-impact. FE THUMS model consists of ligaments modeled as shell elements and were assigned with the linear material properties. The purpose of the ligaments is to transfer the tensile load and the shell element formulation can allow compression. Both these models have less than 80,000 elements which is considered to be a coarse mesh compared to 298,313 elements in our cervical spine model alone. A mesh convergence study was performed for the selection of optimal quality mesh size for the developed FE model. This produces the confidence of predicting stresses and strains more accurately than the coarse mesh models.

6.3 Model Validation

The osseoligamentous 3D human head-pelvis FE model was first validated with static loading in flexion, extension, lateral bending and axial rotations. Pure moment loading was used to obtain these physiological rotations. Due to the lack of any
The experimental work on the full spine kinematics exclusively, regional spine (cervical, thoracic and lumbar) pure moment testing kinematic data was used to compare the FE model predictions. The model agreed reasonably well within one standard deviation of the experimental data. The intradiscal pressures predicted by the FE model were comparable with the previously validated cervical and lumbar spine FE models as well as experiments in the literature as shown in the tables in Appendix A.

Dynamic validation was performed by subjecting the FE model to the rear end impact loading from both human volunteer and full body cadaver experiments. The FE model responses for the rear end impact from human volunteer experiments predicted the global parameters like head angle, T1 angle and head-T1 angle which were well within the volunteer corridors. The vertebral rotations from FE model were having similar trend compared to the volunteer data. These intervertebral rotations from the experiment were representative of one individual volunteer due to which a close fit was not observed.

To be more confident about the intervertebral rotations or segmental level kinematics of the FE model, another rear end impact from full body cadaveric sled experiments was simulated. This was necessary as the soft tissue strains predicted by the FE model would be more accurate if the intervertebral motions in the model were accurate. The model responses were compared with the volunteer set of corridors along with an isolated head-neck FE model from the literature [115]. The predictions made by the FE model had similar trend and was well within the validation corridors whereas it was not the case with the isolated head-neck FE model. The peak flexion angle predicted by their model was on the low end of the experimental results whereas our FE model predicted the peak flexion angle reasonably well within the corridors. Overall, the
kinematic response of the FE model was a good fit to the volunteer data at 4g and the cadaver tests performed at 7g.

### 6.4 Whiplash soft tissue strains

The goal of the finite element study was to test the effect of different initial occupant seating postures on the injury patterns due to rear end full body impacts. For this, FE model was first tested in the normal seating posture in the standard seat with headrest due to rear end impact of 8 km/h and peak sled acceleration of 4g. During the early stages of this rear end impact, thoracic spine is pushed by the seat back causing the straightening of the thoracic spine curvature. This straightening created a peak axial compressive force of 101 N on the neck at the cervico-thoracic junction (C7-T1). The seatback-thoracic spine interaction induced rotations in the neck in such a way that lower cervical vertebrae extended earlier and faster than the upper cervical vertebrae. This shearing motion and the posterior translation (section 5.3) of the head caused an S-shaped curvature in the cervical spine. A peak shear force of 51.4 N was observed at the cervico-thoracic junction. Most of the previous computational models consist of isolated cervical spine simulations which do not account for the axial compression phenomenon in the neck that occurs during the first phase of the rear end impact. Yang et al performed PMHS experiments and proposed that the complex loading (axial compression together with the shear force) makes it easier for the soft tissue injuries to occur and it is responsible for the higher observed frequency of neck injuries in rear end impacts [78-81].
Capsular ligament strains were compared to the reported values for sub-traumatic failure of isolated facet joint exposed to tensile load by Winkelstein et al and Siegmund et al. These strains were in the range of 35%-65% [117, 118]. All other ligament strains were compared to a failure corridor that represented the average ± standard deviation of strain at the ultimate stress from in vitro tensile tests of dissected cervical spine ligaments [119, 120].

Intradiscal pressures (IDP) were compared to the maximum intradiscal pressure values due to physiological loading of the cervical spine which do not exceed 1 MPa [121]. Intradiscal pressures with all the out of occupant seated postures consistently showed higher values compared to the maximum physiological loading. This can be attributed to the increased segmental kinematics with change in the initial occupant seated posture compared to the normal seating posture. The IDP trend increased caudally in the cervical spine with maximum value at C6-C7 level for all the postures.

Based on the clinical and epidemiological evidence, the axial rotation of the head or inclined head and the body forward postures prior to impact had an increasing whiplash injury risk following automotive rear impact [53, 55-57, 85].

The first part of the experiment was to study the influence of progressive body forward position on the injury outcome due to rear end full body impacts. Simulation of these out of occupant seating postures and the respective angles were taken from the literature [81,111- 114, 125].

Results showed that as the body forward position increased, there was a non linear increase in the ALL and CL strains at all the levels of the lower cervical spine. This can be attributed to the increase in the complex loads (axial and shear forces) with the
increase in body forward posture. The axial force on the cervico-thoracic junction increased from 101 N during normal seating posture to 212 N during body forward by 25 degrees (BF-25) posture. Even, the shear force increased from 51.43 N to 119.6 N. Also, with the torso bending forward, the distance between the headrest and head increased due to which the range of motion increased.

The focus of this first study was to identify the pattern of increase in the strains with the progressive increase in the body forward positions. There was a sharp rise in strain parameters between BF-15 and BF-20 whereas there was not much difference between the BF-20 and BF-25 positions. These results reveal that there is an increased whiplash injury risk when the occupant is subjected to a body forward leaning position of more than 20 degrees (BF-20). The values of ligament strains did not see any noticeable increase due to changes in the posture itself, as the head and neck positions for all the body forward cases were maintained in the Frankfort plane that was parallel to the horizontal.

The next test was to investigate the effect of progressive head turned posture on the whiplash injury outcome. Results showed as the amount of head rotation increased, there was a non linear increase in the alar, ALL and CL ligament strains in the cervical spine. This can be attributed to increase in the complex loading (axial and shear forces) with the increase in head turned posture. The axial force on the cervico-thoracic junction increased from 101 N during normal seating posture to 164 N during head rotated by 45 degrees (HT-45) posture. Shear force increased from 51.43 N to 159 N. Also, when the head was rotated, the distance between the headrest and head increased due to which the range of motion increased.
This explains the increase in the FE predictions of HT postures compared to the normal posture. Ligament strains increased in a non-linear fashion with the increase in the head rotated position and had a sharp rise between HT-30 and HT-45. This indicates that any rotated angle greater than HT-30 can cause an increased risk of rear end impact injury.

Capsular ligament strains had an increase due to head rotated posture itself. For HT-15 posture, the increase in capsular ligament strain values at the cervical levels were less than 1%. Table 6.1 shows the increase in capsular ligament strains just due to the posture variation for HT-30 and HT-45 postures. These strains were in trend with the maximum capsular ligament strains observed in the cervical FE model due to axial rotation of the cervical spine under physiological loading [97].

Table 6.1: Increase in CL strains due to change in head turned posture.

<table>
<thead>
<tr>
<th>Level</th>
<th>Capsular ligament strains due to HT-30 Posture (%)</th>
<th>Capsular ligament strains due to HT-45 Posture (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C2-C3</td>
<td>16</td>
<td>21</td>
</tr>
<tr>
<td>C3-C4</td>
<td>18</td>
<td>25</td>
</tr>
<tr>
<td>C4-C5</td>
<td>19</td>
<td>26</td>
</tr>
<tr>
<td>C5-C6</td>
<td>19</td>
<td>26</td>
</tr>
<tr>
<td>C6-C7</td>
<td>12</td>
<td>19</td>
</tr>
</tbody>
</table>

Taking the effect of change in the capsular ligament strain due to head turned posture into account, the CL strains at C2-C3, C3-C4, C4-C5, C5-C6 levels were more than the threshold of 35% strain indicating potential risk of capsular ligament injury.
ALL strains had a small increase due to increase in the head rotated posture itself. For HT-15 posture, the increase in the ALL strain values at the cervical levels were very less. Table 6.2 shows the increase in ALL strains just due to the posture variation for HT-30 and HT-45 postures. Even after taking the ALL strains due to head turned posture itself into account, the strains did not go beyond threshold for potential injury.

Table 6.2: Increase in ALL strains due to change in head turned posture.

<table>
<thead>
<tr>
<th>Level</th>
<th>ALL strains due to HT-30 Posture (%)</th>
<th>ALL strains due to HT-45 Posture (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C2-C3</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>C3-C4</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>C4-C5</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>C5-C6</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>C6-C7</td>
<td>4</td>
<td>6</td>
</tr>
</tbody>
</table>

Alar strains had an increase of 4%, 13.5% and 21.3% due to the change in postures (HT-15, HT-30 & HT-45) itself respectively. Out of all these, only HT-45 went beyond threshold strain (55%) of after taking the increase due to posture itself into account indicating potential alar ligament injury.

In the next experiment, head rotated, head flexed, body forward and combination of body forward and head turned (BF-HT) postures were compared to the normal seating posture. From the results of body forward and head turned postures, it was evident that there is an increased risk compared to other cases. This led to the formulation of the combination of these two high risk postures (body forward head turned posture: BF-HT)
to observe the potential increase in risk factors. BF-HT posture had the highest C6 vertical translation and posterior head translation compared to other cases. This increase can be attributed to the increase in thoracic kyphosis due to forward bending of the occupant.

The intradiscal pressures showed increased loading on the intervertebral discs in all the out of occupant seating postures compared to the physiological loading. The occupant seating postures in which head was rotated (BF-HT & HT) had the highest alar ligament strains followed by the body forward posture. The increase in stress and strain levels for BF-HT posture compared to other cases can be attributed to the increased complex loading values (Axial force: 220 N, Shear force: 168 N) than both body forward as well as head turned postures.

Capsular ligament strains had an increase due to BF-HT posture itself. Capsular ligament strain values at the cervical levels were shown in Table 6.3. Also, the increase in ALL strains due to BF-HT posture were documented in the same table. Alar ligament strain including the effect of BF-HT posture itself along with the rear end impact (0.90 mm/mm) was within the failure corridors (0.55-1.95 mm/mm) with BF-HT posture showing high risk of alar ligament injury. Taking the effect of change in the capsular ligament strain due to BF-HT posture into account, the CL strains at C2-C3, C3-C4, C4-C5, C5-C6 levels were more than the threshold of 35% strain indicating potential risk of capsular ligament injury. Even after taking the ALL strains due to BF-HT posture itself into account, the strains did not go beyond threshold for potential injury.
Table 6.3: Increase in CL & ALL strains due to BF-HT posture.

<table>
<thead>
<tr>
<th>Level</th>
<th>Capsular ligament strains due to BF-HT Posture (%)</th>
<th>ALL strains due to BF-HT Posture (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C2-C3</td>
<td>20</td>
<td>4.1</td>
</tr>
<tr>
<td>C3-C4</td>
<td>25</td>
<td>2.2</td>
</tr>
<tr>
<td>C4-C5</td>
<td>26</td>
<td>1.4</td>
</tr>
<tr>
<td>C5-C6</td>
<td>26</td>
<td>2.1</td>
</tr>
<tr>
<td>C6-C7</td>
<td>19</td>
<td>6.2</td>
</tr>
</tbody>
</table>

The intradiscal pressures as well as facet stresses increased with the axial rotation of head with the posture (HT-45 & BF-HT) alone. After the rear end impact started, these values decreased initially and then increased to reach the maximum values with the increase in extension angle at the respective cervical levels.

The next test investigated the effect of increased impact severity on the identified high risk out of occupant postures. With the increase in the impact acceleration magnitude from 4g to 10g, all the parameters showed an increased behavior in the out of occupant seating postures (HT, BF & BF-HT). Values of ligament strains described here include the strains due to posture variation combined with rear end impact loading.

Capsular ligament strains increased to a peak value of 0.69 mm/mm in the BF-HT posture, 0.44 mm/mm in the body forward posture and 0.56 mm/mm in the head turned posture respectively at C3-C4 level. At the 10g impact, BF-HT, head turned and body
forward positions had the highest peak capsular strain values and were more than the threshold of 35%, indicating potential risk of facet capsular injury. It has been proposed that ligament injuries may occur in the initial S-shape phase of a rear impact [65]. Even here, peak capsular strains occurred during the formation of S-shaped curvature where upper levels of cervical spine was in flexion compared to the lower. This explains the reason for presence of higher capsular ligament strains in the upper levels. The locations of this capsular ligament injury were in agreement with the clinical literature reported by Lord et al [46].

Alar ligament strains also increased in both the head turned and BF-HT postures with the increase in the acceleration magnitude to a peak strain value of 1.7 mm/mm and 1.8 mm/mm respectively. Also, BF-HT posture had peak alar ligament strain of 0.90 mm/mm and head turned posture had 0.69 mm/mm with 4g sled acceleration. For the Alar ligaments, the failure strain corridors ranged from 0.55 mm/mm to 1.95 mm/mm with a mean of 1.25 mm/mm. The Alar ligament strains predicted by the FE model in HT-45 and BF-HT postures during 4g as well as 10g impact exceeded the mean failure strain indicating the increased whiplash injury risk compared to the normal seating posture. This was in agreement with the study by Krafft et al in which they reported that the patients with the turned head prior to the impact had long term symptoms (symptoms persisted more than 2 months) [122]. Even Kaale et al identified more high grade lesions with significance in alar ligament lesions among those with rotated head position than the neutral head position [53].
6.5 Limitations and Recommendations

The limitations of this model include the lack of muscle behavior in the model. Since muscles are said not to be activated before 200 ms and it has been demonstrated that the reflex muscle contraction in the unaware occupant in whiplash does not occur in sufficient time to alter markedly spinal kinematics resulting in soft tissue injury. Therefore, they were not modeled for this current study.

Another limitation was the development of 3D osseoligamentous head-pelvis FE model from single patient CT scans (no abnormalities) due to which the FE responses may not be applicable to all the demographic population. The size and mass of the neck as well as anatomical structures vary among different people based on different parameters like age, gender and body mass index. Mass properties applied to this FE model represent 50th percentile female population. In addition to this, follower type preload was also simulated which accounts for the *in vivo* physiological load on the spine [102]. Above all, FE responses were in agreement with the clinical literature as mentioned earlier indicating that the injury predictions of this FE model could help explain in part post whiplash associated pain.

Nilson *et al* investigated the effect of seat belt usage on car occupant response due to rear end impacts and reported that seat belt usage did not have much effect on the occupant motion [127]. Seatbelt in rear end impact plays an essential role in restricting the body motion after rebound which occurs in the later stages of the rear end impact [24, 63]. As the results (stresses and strains) presented in this study were taken before this phase, seatbelt was not modeled in this study.
Also, legs were not included in the model as the focus of the study was to investigate the neck injury. However, femur was included to enable the accurate positioning of the model relative to the seat bottom.

Future studies can extend this model to a full body model and investigate the effects of different impact loading scenarios on other joint injuries as well. More number of possible configurations considering different occupant seating postures, varying head restraint positions, occupant protection systems (HANS device etc.) and active head restraints should be evaluated.

Since it has been identified clinically that out of occupant seating postures can increase the whiplash injury risk and only few rear end impact experiments reproducing injury are available in the literature with even normal seating posture, additional experimental studies with PMHSs are recommended. These experiments should address the following aspects: a) global head, T1 and pelvis kinematics, b) vertebral rotations, c) facet joint kinematics, d) ligament kinematics, and e) injury monitoring.

6.6 Conclusions

In this thesis, a detailed 3D osseoligamentous human head-pelvis FE model has been developed. The model consists of a skull, vertebrae, viscoelastic discs, frictionless facet joints, nonlinear ligaments, ribcage and pelvis.

The author successfully validated the human head-pelvis FE model under quasi-static and dynamic loading. Quasi-statically, the model agreed reasonably with the experimental data. Dynamically, the FE model exhibited responses which were in good
agreement with the rear end (4g to 10g) experimental data. The validated model was then used to test the effect of whiplash injury due to rear end impacts.

Standard seat with headrest was modeled to study the influence of seating posture on the whiplash injury outcome due to rear end impacts to increased impact severities. The responses of standard seat with headrest in normal seating posture were in agreement with the motion sequence characterized for human volunteers in the experiment. With the progressive increase in body forward and the head turned postures, there was a non-linear increase in the range of motion, intervertebral disc pressures, facet stresses and ligament strains. Intervertebral discs had increased pressures with the out of occupant seating postures when compared to the physiological values indicating high risk of injury. Results supported the hypothesis that body forward posture increased the risk of capsular ligament injury and head turned posture increased the risk of alar ligament injury which were in agreement with the clinical literature. Also, the head turned posture increased the injury risk of capsular ligaments. In addition, combination of both head turned and body forward postures had higher risk of injury compared to the individual postures.
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end impacts: a comparison with the TNO rear impact dummy (TRID) model.


# Appendix A

## FE Model Predictions

Table A.1: Intradiscal pressure (MPa) predictions of head-pelvis human model for cervical spine(C0-C7) under quasi-static loading [128]

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>C3-C4</td>
<td>C4-C5</td>
<td>C5-C6</td>
<td>C6-C7</td>
</tr>
<tr>
<td>Flexion</td>
<td>0.62</td>
<td>0.66</td>
<td>0.54</td>
<td>0.58</td>
</tr>
<tr>
<td></td>
<td>0.32 (0.12-0.43)</td>
<td>0.23 (0.00-0.56)</td>
<td>0.24</td>
<td></td>
</tr>
<tr>
<td>Extension</td>
<td>0.38</td>
<td>0.45</td>
<td>0.28</td>
<td>0.21</td>
</tr>
<tr>
<td></td>
<td>0.32 (0.12-0.43)</td>
<td>0.23 (0.00-0.56)</td>
<td>0.02</td>
<td></td>
</tr>
<tr>
<td>Left Bending</td>
<td>0.42</td>
<td>0.45</td>
<td>0.27</td>
<td>0.21</td>
</tr>
<tr>
<td></td>
<td>0.16 (0.08-0.31)</td>
<td>0.17 (0.00-0.38)</td>
<td>0.11</td>
<td></td>
</tr>
<tr>
<td>Right Bending</td>
<td>0.45</td>
<td>0.33</td>
<td>0.28</td>
<td>0.19</td>
</tr>
<tr>
<td></td>
<td>0.16 (0.08-0.31)</td>
<td>0.17 (0.00-0.38)</td>
<td>0.11</td>
<td></td>
</tr>
<tr>
<td>Left Rotation</td>
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<td>0.55</td>
<td>0.39</td>
<td>0.24</td>
</tr>
<tr>
<td></td>
<td>0.25 (0.14-0.36)</td>
<td>0.16 (0.04-0.49)</td>
<td>0.14</td>
<td></td>
</tr>
<tr>
<td>Right Rotation</td>
<td>0.41</td>
<td>0.53</td>
<td>0.37</td>
<td>0.26</td>
</tr>
<tr>
<td></td>
<td>0.25 (0.14-0.36)</td>
<td>0.16 (0.04-0.49)</td>
<td>0.14</td>
<td></td>
</tr>
</tbody>
</table>
Table A.2: Intradosal pressure (MPa) predictions of head-pelvis human FE model for lumbar spine (L1-S1) under quasi-static loading

<table>
<thead>
<tr>
<th></th>
<th>L1-L2</th>
<th>L2-L3</th>
<th>L3-L4</th>
<th>L4-L5</th>
<th>L5-S1</th>
</tr>
</thead>
<tbody>
<tr>
<td>EXT</td>
<td>0.93</td>
<td>0.82</td>
<td>0.73</td>
<td>0.64</td>
<td>0.70</td>
</tr>
<tr>
<td>FLEX</td>
<td>1.10</td>
<td>0.83</td>
<td>0.84</td>
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<td>0.74</td>
</tr>
<tr>
<td>BENDING</td>
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<td>0.87</td>
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</tr>
<tr>
<td>ROTATION</td>
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<td>0.71</td>
<td>0.65</td>
<td>0.53</td>
<td>0.56</td>
</tr>
</tbody>
</table>

Figure A-1: Convergence study results for dynamic compressive loading simulation with the simplified motion segment.