A biomechanical evaluation of three atlantoaxial transarticular screw salvaging fixation techniques

Tejaswy Potluri

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A Thesis

entitled

A Biomechanical Evaluation of Three Atlantoaxial Transarticular Screw Salvaging Fixation Techniques

by

Tejaswy Potluri

Submitted to the Graduate Faculty as partial fulfillment of the requirements for The Master of Science in Bioengineering

_____________________________
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May 2010
Injuries to the neck, or cervical region, are very severe types of injuries since there is a potential risk of damage to the spinal cord. Any neck injury can have devastating if not life threatening consequences. The upper cervical spine consists of three vertebrae: The occiput (C0), the atlas (C1) and the axis (C2). Stabilization of the atlanto-axial complex following a neck injury is a challenging procedure because of its complicated anatomy. Several stabilization techniques have been reported for C1-C2 fixation. These techniques include posterior wiring and bone graft, posterior transarticular screws or a combination. Fixation with transarticular screws has been the gold standard in effectively stabilizing the segment. However, the drawback of using the transarticular screws is that they have a potential risk of vertebral artery injury due to a high riding transverse foramen of C2 vertebra, screw malposition or fracture of the C1 lateral mass. In such cases, it is not recommended to proceed with inserting the contralateral
transarticular screw and the surgeon should find an alternative to fix the contralateral side. Many studies are available comparing different atlanto-axial stabilization techniques but none of them compared the techniques to fix the contralateral side while using the transarticular screw on one side. The current options are C1 Lateral Mass Screw and C2 Pedicle Screw or C1 Lateral Mass Screw and C2 Intralaminar Screw or C1-2 Sublaminar Wire.

The purpose of this study is to compare the biomechanical stability of the three C1-C2 transarticular screw salvaging fixation techniques through \textit{in vitro} testing and finite element modeling. An in vitro testing using nine cadaver specimens was done to compare the degree of stability afforded by the three salvaging fixation techniques at C1-C2 level. To compare the efficiency of bilateral instrumentation over unilateral case, unilateral transarticular screw was also investigated.

A finite element model of the C0-C3 has been developed and applied to study as well as compare the effects of the three C1-C2 transarticular screw fixation techniques on the biomechanical behavior of the upper cervical spine. The intact model was validated by comparing with previously published experimental results as well as with the \textit{in vitro} results. Finite element models representing the following combinations with C1-2 Transarticular Screw on one side and: 1) C1 Lateral Mass Screw and C2 Pedicle Screw (TS+C1LMS+C2PS) 2) C1 Lateral Mass Screw and C2 Intralaminar Screw (TS+C1LMS+C2ILS) 3) C1-2 Sublaminar Wire (TS+WIRE) on the other side were evaluated. Various parameters like range of motion, facet loading and implant stresses were evaluated. Results show that all the three bilateral fixation procedures significantly reduced motion in all the loading modes when compared to unilateral fixation. When the
three bilateral techniques were compared, TS+C1LMS+C2PS and TS+C1LMS+C2ILS afforded same stability and instrumentation stresses in all the loading modes. In addition, both the techniques were highly stable in axial rotation mode when compared to TS+WIRE. TS+WIRE resulted in higher stresses when compared to the other two bilateral techniques in all the loading modes.
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Chapter 1

Introduction

The cervical spine contains 33 synovial articulations: 14 facet joints, 10 neurocentral joints, 5 discovertebral joints, 2 occipito-atlantal joints, one atlanto-dental joint, and an articulation between the transverse ligament and the posteroinferior aspect of the odontoid process.[1] Appropriately, reported prevalence rates of musculoskeletal conditions involving the cervical spine are relatively high. The National Center for Health Statistics[2] reported that 8.2% of the United States general population had experienced prolonged (greater than 2 weeks) neck pain, with females reporting higher prevalence rates than males. These high prevalence rates create a large financial and medical burden on society. In 1985, approximately 5.2 million physician consults were directly attributable to cervicogenic pain. Hospitalization burdens are equally high, with 1.4 million hospital days needed to treat cervical pain-related lesions between 1985 and 1988. In light of this, the motivation for investigating the biomechanical characteristics of the upper and lower cervical spine cannot be understated.
The upper cervical spine also called the occipito-atlanto-axial complex (C0-C1-C2) consists of occiput (C0), atlas (C1), and axis (C2). At the base of the skull, the occiput (C0) articulates on the atlas (C1) through the convex occiput condyles (OC) (Figure 2-2). The atlas (C1) is a ring-like structure with anterior and posterior arches where the articular facets and transverse processes are located. However, it lacks a vertebral body and a spinous process. The axis (C2) is characterized by an odontoid process or dens which articulates with the posterior aspect of the anterior arch of C1. The C0-C1-C2 complex is the most complicated joint of the skeleton, both anatomically and kinematically[3, 4]. The atlanto-axial articulation accounts for 50% of the rotation (47º) and 12% of flexion/extension of the cervical spine.[4]

Approximately 5-10% of unconscious patients who present to the emergency as the result of a motor vehicle accident or fall have a major injury to the cervical spine. Most cervical spine fractures occur predominantly at two levels. One third of injuries occur at the level of C2, and one half of injuries occur at the level of C6 or C7. Most fatal cervical spine injuries occur in upper cervical levels, either at craniocervical junction, C1 or C2.

Two of the most commonly seen fractures in the upper cervical spine involve the odontoid process and the ring of the atlas. Fractures of the odontoid process of the second cervical vertebra comprise 7-13% of all cervical spine fractures. Fractures of the atlantal ring account for approximately 2% of all spinal injuries: 10% of the injuries involving the cervical spine and 25% of all injuries involving the atlanto-axial complex.

Posterior fusion at C1-C2 segment is indicated in fractures, degenerative osteoarthritis, rheumatoid arthritis, metastasis, ligamentous injuries, congenital
anomalies, infections and tumors. Atlantoaxial instability can cause severe neurological injury or death. Different strategies to fix the atlantoaxial complex can be found in the literature. Many spine surgeons consider bilateral transarticular screw technique the gold standard for posterior fusion of C1-C2. However, transarticular screw fixation is technically demanding and requires considerable experience. Approximately 10-23% of patients requiring atlantoaxial arthrodesis have anatomic variations in the path of the vertebral artery and in the osseous anatomy on at least one side and may not be suitable candidates for transarticular screw placement.\(^{[5-7]}\) Due to the risk of the vertebral artery injury, there is a need for the surgeon to find an alternative method for the fixation of contralateral side. This study focuses on comparing the efficiency of different options available to fix the contralateral side at C1-C2.

1.1 Purpose and Goals of the Study

The studies that have been reported in the literature to fix the atlanto-axial complex involve cadaver models wherein significant anatomic variation may play a role in the outcomes of those investigations. All the studies in the literature have compared the efficiency of similar fixation techniques when used bilaterally or unilaterally, studies focusing on comparing the different combinations of fixation techniques at C1-C2 level are non-existent. Furthermore, changes in the loading profile and the instrumentation stresses due to the stabilization of atlanto-axial junction have not been reported. This is an inherent limitation of the cadaver model.
The goals of this study involve: (1) Comparing the biomechanical efficiency of the three different combinations of atlanto-axial salvaging fixation techniques through an *in vitro* study (2) Development and validation of a finite element model for investigating the effect of the three techniques with respect to kinematic motion reductions and stress changes.
Chapter 2

Literature Review

This chapter covers three main topics: (1) functional anatomy of human cervical spine, (2) kinematics of upper cervical spine, and (3) instability of the atlanto-axial joint and stabilization techniques.

2.1 Functional Anatomy of Human Cervical Spine

The spinal column is a complex structure, whose main functions are to hold us upright, support the head, and protect the spinal cord and blood vessels. The spine is divided into five regions (Figure 2.1):

- The cervical spine,
- The thoracic spine connecting to the ribs,
- The lumbar spine,
- The sacrum, and
- The coccyx.
Figure 2-1: The human spine with the cervical spine extending from C1 to C7, the thoracic spine from T1 to T12, the lumbar spine from L1 to L5, the sacrum and the coccyx. (Copyright 1962 & 1992. ICON Learning Systems, LLC, a subsidiary of MediMedia USA Inc. Reprinted with permission from ICON Learning Systems, LLC, illustrated by Frank H. Netter, M.D. All rights reserved.)
The human cervical spine is the upper part of the spine, which contains the top seven vertebrae. It supports the head and allows for a wide range of motion. It can be divided into upper cervical and lower cervical spine. This study focuses mainly on the upper cervical spine, which has a different anatomy than the rest of the vertebral column and is the most flexible part of the cervical spine.

2.1.1 Upper Cervical Spine

The upper region is also commonly called the “cervicovertebral junction” or the “craniovertebral junction” (CVJ). It is composed of three bony structures: the occipital bone (the occiput), the atlas, and the axis (Figure 2.2). The occiput is not technically part of the vertebral column but is often denoted as the zeroeth cervical vertebra (C0). At the base of the skull, the occiput (C0) articulates on the atlas (C1) through the convex occiput condyles (OC) (Figure 2.2). The two consecutive vertebrae, the joints connecting them and the ligaments in between make up a motion segment. Hence, the cervical spine contains eight motion segments. These are named after the two vertebrae: C0-C1, C1-C2, etc. The last motion segment in the cervical spine is C7-T1, where T1 is the first vertebra in the thoracic spine.
2.1.1.1 Atlas

The first cervical vertebra (figure 2.3) is referred to as the atlas (C1) and is named after the giant who carried the earth on his shoulders; similarly the atlas holds up the head. It is a ring-like structure with anterior and posterior arches where the articulating facets and transverse processes are located. However, it lacks a vertebral body and a spinous process. Its relatively wide profile (when compared to the cervical vertebrae)\(^9\) and ring-like structure serves to support the skull and can be thought of having five equal parts. The anterior arch lies anterior to the plane of the other cervical vertebrae bodies and the odontoid process of the axis (C2). This also has an anterior tubercity which serves as an attachment for the anterior longitudinal ligament and the longus colli muscles. Posteriorly, this arch has an articulating facet that serves to provide a joint area with the dens (odontoid process). The lateral masses of the atlas have both superior and inferior articular facets. The superior facets are elongated, kidney-shaped, and concave, and serve to receive the occipital condyles. It has been shown that the right and left superior atlantal facets can demonstrate gross asymmetry\(^{10}\). The inferior facets
are flatter and more circular and permit axial rotation. Transverse processes extend laterally from each lateral mass. Within each transverse process is a foramen that is bisected by the vertebral artery. The final area of the atlas is the posterior arch which serves to connect the right and left halves of the lateral masses. The posterior arch also contains a tubercle which acts as an attachment for the ligamentum nuchae. Immediately posterior to the superior facets is a groove that is apparent in each half of the posterior arch. This groove is traversed by the vertebral artery after it passes through the transverse foramen of the atlas.

Figure 2-3: Anatomy of the 1st cervical vertebrae (Copyright 1962 & 1992. Icon Learning Systems, LLC, a subsidiary of MediMedia USA Inc. Reprinted with permission from ICON Learning Systems, LLC, illustrated by Frank H. Netter, MD. All rights reserved.)

2.1.1.2 Axis

The second cervical vertebra (Figure 2.4) is referred to as the axis (C2) probably because it provides the axis for axial rotation in the upper cervical spine. C2 is different from other cervical vertebrae because of its odontoid process (Figure 2.4). Projecting superiorly from the vertebral body of the axis, the odontoid process articulates with the
anterior arch of the atlas. The odontoid process is fixed at its base and is held in its place by the transverse ligament of the atlas. Huge variability has been shown in the dimensions of the dens\textsuperscript{[11]}. The superior facets are large, convex shaped and are located posterolaterally to the odontoid process. The superior facets emerging from the vertebral body and pedicles articulate with the inferior facets of C1. The inferior facets of C2 project posteriorly as one traverse inferiorly and articulate with the superior facets of C3. The second cervical nerve emanates posterior to this synovial joint due to the extreme anterior placement of the superior facets, and thus does not groove the transverse process. C2 has a large bifid spinous process that is the attachment site delineating the craniovertebral and the atlas. The inferoposterior aspect of the dens articulates with the transverse process\textsuperscript{[12]}.

\textbf{Figure 2-4}: Anatomy of the 2\textsuperscript{nd} cervical vertebrae (Copyright 1962 & 1992. Icon Learning Systems, LLC, a subsidiary of MediMedia USA Inc. Reprinted with permission from ICON Learning Systems, LLC, illustrated by Frank H. Netter, MD. All rights reserved.)
2.1.2 The Joints

In the upper cervical spine there are two joints, the occipitoatlantal joint and the atlantoaxial joint. These joints differ from the joints in the lower cervical spine since they do not have a disc. The difference in structures gives distinct characteristics. The occipitoatlantal joint is most flexible in flexion-extension, the so-called ‘yes-motion’ of the head. The main function of the atlantoaxial joint is to allow rotation, a ‘no-motion’ of the head. Lateral bending is distributed evenly between the spinal joints.[13]

2.1.2.1 The Occipitoatlantal Joint

The atlantooccipital joint connects the top of the cervical spine to the base of the skull. This joint is formed by the superior facets of C1, the anterior and posterior atlantoaxial membranes that span between the anterior and posterior arches of C1, and the skull’s foramen magnum. This joint is involved primarily with nodding (i.e., capital flexion) as well as sideways tilting of the head.

2.1.2.2 The Atlantoaxial Joint

The atlantoaxial joint is formed by the facets between C1 and C2 and comprise the joining of the superior facets of C2 with articular surfaces on the anterior arch of C1. This joint is primarily responsible for rotation of the head.

2.1.2.3 The Articulations

There are six synovial articulations in the occipitoatlantoaxial complex: the paired atlantooccipital joints, the paired atlantoaxial joints, the joint between the odontoid
process and the anterior arch of the atlas, and the joint formed by the transverse ligament and the posterior aspect of the odontoid process. The last two joint articulations will be discussed in this section:

2.1.2.3.1 Atlanto-Dental Joint

The odontoid process articulates anteriorly with the posterior aspect of the anterior atlantal ring (Figure 2.7). The joint is actually a bursal joint, with absence of specific capsular ligaments.

2.1.3 The Ligaments

Ligaments join the vertebrae, stabilize the joints in the spine and restrict motion. Ligaments specific to the upper cervical spine are the apical ligament, the alar ligament, the transverse ligament (TL), the tectorial membrane (TM), the anterior atlantooccipital membrane (AAOM), and the posterior atlantooccipital membrane (PAOM), accessory ligament, anterior longitudinal ligament, and nuchal ligament.

2.1.3.1 Accessory Ligament

The accessory atlantoaxial ligaments are bilateral structures that run between the base of the odontoid process and the lateral masses of the atlas.
2.1.3.2 Alar Ligaments

Attachment of the cervical spine to the skull is also achieved by the paired alar ligaments. These ligaments run bilaterally from the occipital condyles inferirolaterally to the tip of the odontoid process. The alar ligaments also contain fibers that run bilaterally from the odontoid process anterolaterally to the atlas. These ligaments limit over-rotation of the craniovertebral junction.

2.1.3.3 Anterior Longitudinal Ligament

The most anterior of the major ligaments is the anterior longitudinal ligament. This ligament extends inferiorly from the anterior margin of the foramen magnum to the superior surface of the anterior arch of the atlas at the anterior tuberosity. The ligament continues inferiorly to the anterior aspect of the axial body (Figure 2.5).

2.1.3.4 Apical Ligament

The apical dental ligament extends from the anterior portion of the magnum foramen to the tip of the odontoid process. Some controversy exists as to its functional significance (Figure 2.5).

2.1.3.5 Cruciform and Transverse Ligament

The cruciform ligament is usually denoted as containing two separate entities: the atlantal transverse ligament and the inferior/superior fascicles. The transverse ligament attaches between the medial tubercles of the lateral masses of the atlas, passing posterior to the odontoid process (Figure 2.6). The superior fascicles run inferiorly from the
posterior aspect of the anterior portion of the foramen magnum to the transverse ligament. The inferior fascicles project inferiorly from the transverse ligament to the posterior aspect of the body of the axis.

2.1.3.6 Nuchal Ligament

The nuchal ligament (ligamentum nuchae) extends from the base of the occiput to the posterior tubercle of the axis, continuing inferiorly to the spinous process of the subaxial vertebrae[^14^].

2.1.3.7 Tectorial Membrane/Ligament

Extending from the body of the axis to the inner surface of the occiput, the tectorial membrane is the most posterior ligament and actually represents the cephalad extension of the subaxial posterior longitudinal ligament. The tectorial membrane has been implicated as a check against extreme flexion motion.

---

**Figure 2-5:** Coronal view of the arrangement of the alar, apical and transverse ligaments. The transverse ligament forms an articulation with the posterior aspect of the odontoid process (taken from Ghanayem et al.[^15^]).
Figure 2-6: Posterior (left) and anterior (right) views of the occipito-atlanto-axial complex demonstrating various ligamentous structures: the transverse ligament (TR), alar ligament (AL) and apical ligament (AP). Notice the inferior and superior (retracted at right) fascicles blending with the transverse ligament to form the cruciform ligament (taken from Heller and Pedlow[12]).

2.2 Kinematics of the cervical spine

In this section, the range of motion of the cervical spine and the coupling motions are discussed. Most of the axial rotation occurs in the region between C1 and C2, while C0-C1 allowed for most of the flexion/extension and very little axial rotation. About 60% of the axial rotation of the entire cervical spine and occiput occurred in the upper region (C0-C1-C2) and the remainder occurred in the lower region (C2-T1)[4]. There was no significant translatory movement at the C0-C1-C2 complex. In the lower cervical spine, most of the flexion/extension occurred in the central region. The geometry of articulations and soft tissue linkages produce “coupled” motion in which case application of load on one plane of rotation not only leads to rotation and/or translation in that direction but also motion in the other two planes. There are two main coupled movements in the lower cervical spine that are due to the spatial orientation of the facet joints and the uncovertebral joints[16]. The first is that flexion is coupled with anterior translation and extension is coupled with posterior translation. The second behavior is the
coupled motion between lateral bending and axial rotation. In the upper cervical spine, there is a significant coupling pattern where the axial rotation of C1 is associated with vertical translation.\cite{4}

Motion studies of the cervical spine have been carried out for a long time.\cite{17-19} Because of the limited accuracy of these earlier studies, they are neither described nor their results included for consideration in this document. The following sections report the modern \textit{in vivo} and \textit{in vitro} studies of the cervical spine and coupling characteristics found by various studies in flexion/extension, lateral bending and axial rotation.

\subsection{2.2.1 Flexion-Extension}

Large sagittal plane rotations have been attributed to the cranovertebral junction (Table 2.1). An \textit{in vivo} study done by Penning\cite{20, 21} reported that both levels (C0-C1 and C1-C2) demonstrated 30° of rotation, while C2-C3 has 12° of rotation. Another \textit{in vivo} study Dvorak et al.\cite{22} studied 9 healthy adults and 43 patients with cervical spine injury using functional CT scans. The results are given in Table 2.1. Dvorak and his colleague\cite{23} introduced the concept of \textit{in vivo} passive motion measurements in contrast to the normally used \textit{in vivo} active motion measurements. The cervical spines of 59 adults were examined by means of functional radiographs. They studied 28 healthy adults and 31 patients who had sustained soft tissue injury to the cervical spine. By laying the flexion x-ray on top of the extension x-ray and using simple graphical construction, they measured the rotational ranges of motion in flexion-extension. On average, the passive motion was about 2-3° larger than the active motion. Results of their study are given in Table 2.1.
Panjabi and associates\cite{24} measured the load-displacement curves of the cervical spine specimens. The loads applied were six forces: anterior-posterior shear, medial-lateral shear, and compression-tension. The maximum force used was 50 N. Using a novel technique they measured three dimensional motions of each motion segment of the cervical spine specimens. The resulting rotational motions are given in Table 2.1. Additionally, they also measured the translations as well as neutral zones for both the rotations and translations. Flexibility coefficients were also computed.

Panjabi and co-workers\cite{25}, in another in vitro study, measured motions of the upper cervical spine in three dimensions. Ten fresh cadaver spine specimens were used and subjected to a maximum moment of 1.5 Nm at the occiput and the three dimensional motions of occiput-C1 and C1-C2 were measured, using stereophotogrammetry. In addition to the ranges of motion, which are provided in Table 2.1, the authors also measured the elastic and neutral zones.

Later, White and Panjabi\cite{4} reviewed the literature on kinematics of the spine and realized the inadequacy of the available data concerning the cervical spine. As a result, they produced a set of values ranges of motion that they thought best represented the motions of the spine. This compilation is also given in the Table 2.1, as a reference.
Table 2-1: Ranges of combined Flexion/Extension motion reported in degrees (mean ± standard deviation) from in vivo and in vitro studies at all the three levels in the upper cervical spine.

<table>
<thead>
<tr>
<th>Flexion+Extension</th>
<th>C0-C1</th>
<th>C1-C2</th>
<th>C2-C3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Penning[20]</td>
<td>30.0±5.0</td>
<td>30.0±5.0</td>
<td>12.0±2.8</td>
</tr>
<tr>
<td>In vivo</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dvorak[22]</td>
<td>-</td>
<td>12.0±4.0</td>
<td>10.0±3.0</td>
</tr>
<tr>
<td>In vivo</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dvorak[26]</td>
<td>-</td>
<td>15.0±3.0</td>
<td>12.0±2.0</td>
</tr>
<tr>
<td>In vivo</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Panjabi[24]</td>
<td>-</td>
<td>-</td>
<td>9.9±1.2</td>
</tr>
<tr>
<td>In vitro</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Panjabi[25]</td>
<td>24.5±4.0</td>
<td>22.4±4.7</td>
<td>-</td>
</tr>
<tr>
<td>In vitro</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>White and Panjabi[4]</td>
<td>25.0</td>
<td>20.0</td>
<td>10.0±2.8</td>
</tr>
</tbody>
</table>

2.2.2 Lateral Bending

Lateral flexion in the occipito-atlanto-axial complex can be studied by observing the movement of the atlas with respect to the occiput. Due to its triangular lateral mass geometry, the atlas moves in the same direction of lateral flexion of the head. The same can be said for the occiput, that is if the atlas is pushed laterally, the occiput, and hence the entire head, will follow in the same direction. As is shown in Table 2-2, an in vitro study by Panjabi et.al [25] reported the occipito-atlantal lateral bending to be 5.5° and that of atlanto-axial bending is 6.7°. The lateral bending at C2-C3 level was reported as 4.7° from an in vitro study done by Moroney et.al.[27] From the review of literature by White and Panjabi[4], they reported 5° at each of the C0-C1 and C1-C2 levels and 10° at C2-C3 level.
Table 2-2: Ranges of lateral bending motion reported in degrees (mean ± standard deviation) at all the three levels in the upper cervical spine.

<table>
<thead>
<tr>
<th>Lateral Bending (one side)</th>
<th>Panjabi\textsuperscript{[25]}</th>
<th>Moroney\textsuperscript{[27]}</th>
<th>White &amp; Panjabi\textsuperscript{[4]}</th>
</tr>
</thead>
<tbody>
<tr>
<td>C0-C1</td>
<td>5.5±2.5</td>
<td>-</td>
<td>5.0</td>
</tr>
<tr>
<td>C1-C2</td>
<td>6.7±4.4</td>
<td>-</td>
<td>5.0</td>
</tr>
<tr>
<td>C2-C3</td>
<td>-</td>
<td>4.7±3.0</td>
<td>10.0±2.3</td>
</tr>
</tbody>
</table>

2.2.3 Axial Rotation

Almost all of the occipito-atlanto-axial contribution to axial rotation occurs in the atlanto-axial region. Atlanto-axial rotation occurs about an axis that passes vertically through the center of the odontoid process. In maximal rotation there is minimal joint surface contact, and sudden over-rotation of the head can lead to interlocking of the C1-C2 facets, making it impossible to rotate the head back to neutral. Table 2-3 lists the amount of rotation found at C0-C1, C1-C2 and C2-C3 joints by various researchers both \textit{in vitro} and \textit{in vivo}.

The findings in Table 2-3 demonstrate that there is a relatively small contribution from the C0-C1 joint, with researchers finding between 5° and 7.2° of rotation. One interesting anatomical note concerning axial rotation is the behavior of the vertebral artery during rotation. The vertebral artery possesses a loop between the atlas and axis, thus affording it over-length. Upon atlanto-axial rotation the slack is taken up in the loop.
and its straightens, thus preventing over-stretching and possible rupture during maximal rotation.

Table 2-3: Ranges of axial rotation motion reported in degrees (mean ± standard deviation) at all the three levels in the upper cervical spine.

<table>
<thead>
<tr>
<th>Axial Rotation (one side)</th>
<th>Dvorak[^{26}] In vivo</th>
<th>Penning[^{20}] In vivo</th>
<th>Panjabi[^{25}] In vitro</th>
<th>Moroney[^{27}] In vitro</th>
<th>White &amp; Panjabi[^{4}]</th>
</tr>
</thead>
<tbody>
<tr>
<td>C0-C1</td>
<td>4.0±1.6</td>
<td>1.0±1.8</td>
<td>7.3±2.2</td>
<td>-</td>
<td>5.0</td>
</tr>
<tr>
<td>C1-C2</td>
<td>43.1±5.5</td>
<td>40.5±4.3</td>
<td>38.9±5.4</td>
<td>-</td>
<td>40.0</td>
</tr>
<tr>
<td>C2-C3</td>
<td>-</td>
<td>3.0±2.5</td>
<td>-</td>
<td>1.9±0.7</td>
<td>3.0±2.5</td>
</tr>
</tbody>
</table>

The preceding has demonstrated the degree of motion afforded by the occipito-atlanto-axial complex. Description of this motion is important to the current investigation and will be used to show that modeling of this region falls within these reported ranges, thus validating the model for future predictions.

2.3 Instability of Atlanto-Axial Complex

The atlanto-axial complex is a complicated structure composed of the upper two vertebrae of the cervical spine, their articulating surfaces, and several crucial ligaments. Because of this intricate relationship, the atlanto-axial instability can occur when any part of the components are damaged by trauma, inflammation, neoplasm, or congenital defects.\[^{28}\] Several techniques have been described to address this problem, ranging from
external immobilization to ventral and/or dorsal surgical fusion and internal fixation procedures. All of these have their different advantages, risks, and success or failure rates.

2.3.1 Indications for C1-2 Stabilization

2.3.1.1 Traumatic Fracture

Trauma is among the most frequent indications for posterior C1-C2 stabilization. Traumatic injuries that are amendable to posterior C1-C2 fixation include certain subsets of Type II and Type III odontoid fractures. Anderson and D’Alonzo \[29\] divided the dens fracture into three groups (figure 2.7). Type 1 fractures involve fractures through the upper dens, and, although they are generally considered stable, a recent report suggested that this stability is not absolute.\[30\] Type 2 fractures occur at the junction of the dens and the body of C2, and Type 3 fractures extend into the C2 body.

Although most Type II odontoid fractures can be treated either with immobilization or with anterior odontoid screw fixation,\[31\] there are several subsets of this fracture pattern which are not amendable to these treatment measures. These include Type II odontoid fractures associated with fractures of the atlantoaxial joint, Type II odontoid fractures with oblique fractures in the frontal plane that preclude odontoid screw placement, Type II odontoid fractures with significant displacement which may not heal in immobilization (and are too displaced to place an odontoid screw), Type II odontoid fractures with an associated Jefferson fracture, and Type II odontoid fractures with a ruptured transverse ligament.\[31\]
In addition, patients with a very large thoracic kyphosis or a very large barrel chest preclude the appropriate angle for anterior odontoid screw placement, and must be treated with a posterior C1 and C2 stabilization procedure.\textsuperscript{[31]}

Even when there is a Type II odontoid fracture that might heal with immobilization, there are certain cases where immobilization is not practical. Elderly patients in particular do not heal well with immobilization. They have a higher rate of nonunion due to osteoporosis and have increased respiratory morbidity when placed in halo vests.\textsuperscript{[32]} In addition, all patients initially treated with immobilization who develop a pseudoarthrosis are not good candidates for subsequent attempts at anterior odontoid screw fixation because of the pseudoarthrotic material occupying the fracture line which prevents contact of the decorticated fracture surfaces.\textsuperscript{[31]}

For patients who have failed immobilization and are no longer good candidates for anterior odontoid screw fixation, C1 and C2 fixation is the only remaining treatment option.

Type III odontoid fractures with atlantoaxial joint fracture combinations and Type III odontoid fractures with associated Jefferson fracture are also unstable and are often best treated with a posterior C1 and C2 stabilization procedure.\textsuperscript{[31]}
Figure 2-7: The three types of odontoid process fracture patterns described by Anderson and D’Allozono. Type II is the most commonly occurring (taken from Anderson and D’Alonzo[29]).

2.3.1.2 Traumatic Ligamentous Laxity

Flexion of the upper cervical spine is limited by the tectorial membrane and cruciform ligament, including the transverse ligament, and the alar ligaments.[23, 33] If the atlantodental interval (normal value, 2–4 mm) has a value greater or equal to 5 mm, ligamentous laxity should be suspected; and an interval of more than 10 to 12 mm indicates complete destruction of the ligamentous complex.[34] Axial rotation of the upper cervical spine is limited by the alar ligaments and damage to these increases rotation in
the contralateral side by 30%.\textsuperscript{[22]} Failure of any of the components of the atlantoaxial ligament complex requires dorsal surgical fusion.

2.3.1.3 Rheumatoid Arthritis

Rheumatoid arthritis (RA) is a systemic disease that often affects the cervical spine, both the atlantoaxial junction and the subaxial spine. In as many as 49% of patients with RA, symptomatic atlantoaxial subluxation is encountered,\textsuperscript{[35]} with 20% of these becoming myelopathic.\textsuperscript{[36]} As many as 88% of RA patients have radiographical evidence of C1–C2 involvement, and postmortem studies have shown atlantoaxial dislocation to be the cause of death in as many as 10% of patients with RA.\textsuperscript{[35, 37]} RA progression in the cervical spine causes pain and symptoms of cord compression, and multiple authors advocate surgery, regardless of the presence of neurological symptoms, if the posterior atlantodental interval is equal or less than 14 mm.\textsuperscript{[38]} Dorsal fusion has been more successful than ventral approaches for stabilizing the rheumatic atlantoaxial complex.\textsuperscript{[39, 40]}

2.3.1.4 Congenital Disorders

A failure of the tip of the dens to fuse with the main odontoid process results in persistent ossiculum terminale. Although it is often confused with a Type 2 dens fracture, this process is stable, and the height of the dens is unaffected. Os odontoideum is the failure of the dens to fuse with the body of the axis. Both os odontoideum and odontoid agenesis may lead to incompetence of the cruciate ligament and subsequent atlantoaxial instability.\textsuperscript{[40]}
2.3.2 Techniques of C1-2 Fixation

Until the past few decades, the treatment for atlanto-axial instability was external immobilization. The non-union rate and morbidity associated with this technique limited its use in some instances. In 1910 Mixter and Osgood, [41] presented the first technique of dorsal atlantoaxial fusion which involved variations of wiring together of the posterior elements of the axis and atlas. These techniques are technically simple and require no special intraoperative equipment, such as fluoroscopy or surgical navigation. These techniques all require rigid postoperative immobilization for successful fusion.

2.3.2.1 Wiring Fixations of C1-C2

In 1939, Gallie [42] first described the stabilization of subluxed vertebrae with “fine steel wire passed around the laminae or spines . . . and . . . by bone grafts laid in the spines of the laminae and articular facets.” The Gallie-type fusion (Figure 2.8 (a)) involves a median bone graft notched over the spinous process of C2, with a sublaminar wire placed around the posterior arch of C1 and looped around the spinous process of C2 to hold the graft in place. Although this is the simplest dorsal fusion with minimal technical hazards, it remains the poorest biomechanical construct. The Gallie-type fusion offers minimal stabilization in rotation, with comparable anterior-posterior translation in response to flexion with other techniques.[43] This contributes to its rate on nonunion as high as 25%, and patients require longer periods of external immobilization postoperatively.[44] The placement of an onlay graft also reduces the fusion rate compared with grafts placed under compression. Because it requires an intact posterior arch of C1, the Gallie-type
fusion cannot be used if there is an associated Jefferson fracture or rheumatic involvement of the atlas. Posterior rachischisis, with an incidence of 4%,\textsuperscript{[45]} also prevents stabilization by this approach. The Gallie-type fusion is often used to supplement other techniques.

In the Brooks-Jenkins fusion (Figure 2.8 (b)), the posterior arch of the atlas and the laminae of the axis are exposed, and doubled 20-gauge wires are passed under the laminae of the atlas and axis bilaterally. Two posterolateral autologous iliac crest bone grafts are beveled to fit both interlaminar spaces and held in place by the overlying wire.\textsuperscript{[46]} Although this construct provides more rotational stability than the Gallie-like technique, the simultaneous passage of two-segment sublaminar wires results in a wider curvature and potentially compresses the cord in a “clothesline” fashion.\textsuperscript{[47]} Successful fusion rates of 93% are described.\textsuperscript{[46]} The Brooks-Jenkins fusion also requires an intact posterior arch of C1 and has the same contraindications as described for the Gallie method.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure2-8.png}
\caption{(a) Gallie-type fusion \hspace{1cm} (b) Brooks-Jenkins fusion}
\end{figure}

\textbf{Figure 2-8:} (From, Wright NM et al\textsuperscript{[59]})

In Sonntag’s Modified Gallie fusion (Figure 2.9 (a)), a single bicortical bone graft is fit into the interlaminar space between the atlas and axis and notched to
accommodate the spinous processes of the axis. This technique provides increased stability without the disadvantages of the two-level sublaminar wires in the Brooks-Jenkins technique\textsuperscript{[48]}. Patients are kept in a halo preoperatively and intraoperatively for optimal anatomic realignment. The C1–C2 interlaminar space is widened with a high-speed drill, the spinous process and laminae of the axis are decorticated, and the inferior aspect of the spinous process is notched to seat the wire. A 4-cm-long iliac crest graft is shaped to fit the interlaminar space, with the concave cortical surface facing the dura. The inferior aspect of the graft is notched to lie over the spinous process of the atlas, and two strands of #24 wires are passed around the posterior arch of the atlas, over the bone graft, and around the notched spinous process of the axis. The wires are tightened to three turns per centimeter. Postoperatively, the patients are recommended to stay in a halo for 3 months, followed by a hard collar for 4 to 6 weeks. A 97% fusion rate is described.

2.3.2.2 Interlaminar Clamps

The interlaminar clamp technique (Figure 2.9 (b)) provides similar fusion to that of the Brooks-Jenkins method, but without the disadvantage of sublaminar wires. In the Halifax technique, a double hook and screw construct stabilizes the laminae of C1 and C2 bilaterally and secures bilateral interlaminar bone grafts.\textsuperscript{[47]} Initially, this technique was described on a single side only and without the addition of a bone graft with acceptable results, but when used to stabilize the C1–C2 complex, bilateral clamps with bone grafts have proven to be superior.\textsuperscript{[49]} Biomechanical experiments have shown this technique to provide excellent anteroposterior stability. However, the rotational movement has been less successful than either the Brooks-Jenkins or the Magerl techniques.\textsuperscript{[43,50]} Because it
also requires an intact arch of C1, it has the same contraindications as the methods described previously. Immobilization after surgery only requires a cervical collar, allowing early mobilization.

(a) Sonntag’s modified Gallie fusion                        (b) interlaminar clamp fusion

**Figure 2-9:** *(From, Wright NM et. al[51]*)

2.3.2.3 C1-2 Screw Fixation

Because of the limitations of stabilization of rotation observed with all dorsal wiring techniques, newer techniques of dorsal atlantoaxial fixation have used rigid screw fixation of the atlas and axis. These rigid screw techniques provide significantly higher rates of fusion and allow less rigid postoperative immobilization but are technically more demanding, requiring intraoperative fluoroscopy and/or surgical navigation tools.

2.3.2.3.1 Transarticular Screw Fixation

The Magerl transarticular screw method[52] (Figure 2.10) is more technically demanding than techniques of dorsal wiring, but it has two advantages. First, it does not rely on intact posterior elements and can, therefore, be used in patients in whom the previously
described techniques are contraindicated. Secondly, the transarticular screw greatly reduces rotatory movement, increasing the stability and fusion rates of the construct.

Transarticular instrumentation of Magner is considered the gold standard technique for posterior fusions of C1-C2. The screws are inserted at the inferior aspect of the laminae approximately 2 mm cranial and lateral of the medial border of the caudal articular process of C2. A drill guide and fluoroscopic imaging are used as the screw is passed across the facet joint into the lateral mass of C1. Transarticular screw fixation requires considerable experience.

Several clinical and cadaveric studies have shown the reliability in the strength and stability of the transarticular screw constructs. Successful fusion rates have been described as between 86.9 and 100%, and biomechanical cadaveric studies show that the construct is stable not only in flexion and extension, but also in rotation. A recent study by Reilly et al. comparing C1-C2 wiring with transarticular screw fixation, documented fusion rates of 71% and 93%, respectively. Fuji et al., however, described a 95.5% rate of adequately positioned screws with a 69.4% rate of the screws perforating the anterior cortex of the anterior arch of the atlas, which, in most cases, was clinically nonsignificant. A review of 75 patients performed by Haid et al. documented a fusion rate of 96%.

Several biomechanical studies have compared the transarticular screws technique with wiring techniques (the Gallie type and the Brooks type). Hanley et al. found that the Brooks technique was twice as stiff in extension and flexion and five times as stiff in rotation compared with the Gallie technique and simple midline wiring. Naderi et al. studied biomechanically in vitro combinations of cable-graft-screw at C1-C2. Spinal
stiffness increases after spinal instrumentation with two transarticular screws plus a posterior wire-graft compared with a wire-graft alone. The transarticular screws prevented lateral bending and axial rotation better than the posterior wire-graft. The wire-graft prevented flexion and extension better than the screws. Increasing the number of fixation points often significantly decreased the rotation and translation. Axes of rotation shifted from their normal location toward the hardware. Naderi et al. [69] studied one-point fixation technique compared to three-point fixations using transarticular screws and concluded that it is mechanically advantageous to include as many fixation points as possible for the treatment of atlantoaxial instability. Montesano [70] and Grob [43] compared in vitro different posterior atlantoaxial fusions: the two wire techniques (Gallie type and Brooks type) to the transarticular screw method (Magerl technique). Each fixation technique decreased motion in all directions significantly when compared to the intact and injured spines. They found that the Gallie technique was associated with significantly more motion in flexion, extension, axial rotation, and lateral bending than Brooks and Magerl technique. The Magerl technique tended to allow the least rotation. Magerl et al. [52] and Grob et al. [43] did not report any vertebral artery injury.

The main limitation of the transarticular technique relates to anatomic variations precluding safe screw placement in some patients. A cadaveric study by Madawi et al. [6] and Madawi et al. [5] demonstrated that bilateral screws could not be placed in up to 20% of specimens because of anatomic variations in the location of the foramen transversarium that placed the vertebral artery at risk during screw placement.
2.3.2.3.2 C1-2 Screw-Rod Fixation

Fixation by transarticular screws combined with posterior wiring and structural bone graft leads to excellent fusion rates but the technique is very demanding and poses a great risk of vertebral artery injury. Because of the narrow restrictions on screw trajectory necessary to traverse the C2 pedicle and C1–C2 joint without endangering the vertebral artery, up to 20% of patients cannot have safe placement of bilateral screws. In response to these technical limitations, several authors reported on a atlantoaxial fixation using independently placed screws into the C1 lateral mass and C2 pedicles, connected with posterior rods (Figure 2.11 (a)), or plates. Because C1 and C2 are instrumented individually, fewer patients are precluded from rigid fixation because of anatomic variations.

Harms and Melcher reported satisfactory screw placement with no vertebral artery injuries. Integrity of the posterior arch of C1 is not necessary for stable fixation. This technique seems reliable and effective and should be considered as an efficient alternative to the traditional fixations of the atlantoaxial complex.
Although technically simpler than the transarticular screw technique of Magerl, C2 pars screw placement remains technically demanding and requires intraoperative fluoroscopy and/or surgical navigation. Although more widely applicable than transarticular screws, some patients have a narrow C2 pars or medially located foramen transversarium, precluding safe C2 pars screw placement.\[75\]

Leonard and Wright\[76\] and Wright\[51,77\] recently proposed a new technique which involves the insertion of polyaxial screws into the laminae of C2 in a bilateral crossing fashion and then connected to C1 lateral mass screws using rods (Figure 2.11 (b)). This technique does not have the risk of vertebral artery injury providing safe fixation of C2 since the C2 screws are not placed near the vertebral artery. However, this technique requires intact posterior elements of C2. A larger series of 20 patients with a follow-up period of at least 1 year has recently been reported,\[77\] with 100% fusion rates and no complications. Biomechanical studies showed that the C2 laminar screws have same flexibility as that of transarticular screw fixation.

![Figure 2-11:](From, Wright NM et. al\[62\])
2.4 Mathematical Modeling Techniques

A computer model is a set of mathematical equations that incorporate the geometry and physical properties of a structure that they represent. The most common mathematical model of the cervical spine is a finite element model (FEM). The FEMs incorporate realistic geometry of the vertebrae, taken from CT scans of real spines, and physical properties of the soft tissue connecting the vertebrae. The kinematics (intervertebral motions), kinetics (motions in response to applied loads), internal strains and stresses are possible subjects for study. Although kinematics and kinetics can be studied by *in vivo* and *in vitro* experiments, internal stresses and strains cannot be studied by any experiment. Knowing the stresses and strains within the vertebrae in response to loads applied to the specimen, it is possible to obtain insight in phenomena such as bone adaptation, osteopenia, and osteoporosis. Puttilitz et al.[78] determined the alterations in the joint loading, kinematics, and instrumentation stresses in the craniovertebral junction after application of a novel instrumentation system that can be used with transarticular screws or with C2 pedicle screws unilaterally or bilaterally. From the finite element predictions they concluded that the C2 pedicle screw fixation provided the same relative stability and instrumentation stresses as C1-2 transarticular screws when bilateral instrumentation is applied.

2.5 Summary

As has been demonstrated in the preceding review of literature, there were many biomechanical studies which discussed and compared different techniques to stabilize the
atlanto-axial segment. None of the studies available so far compared the alternatives that are present to fix the contralateral side in case of complications arising due to the vertebral artery injury when unilateral transarticular screw fixation is employed. Therefore, this study focused on the current options available to fix the contralateral side: C1 Lateral Mass Screw and short C2 Pedicle Screw (C1LMS+C2PS), or C1 Lateral Mass Screw and C2 Intralaminar Screw (C1LMS+C2 ILS), or Sublaminar Wire (WIRE). The objective of the present study is to biomechanically compare the three atlantoaxial transarticular salvaging fixation techniques in a type II odontoid fracture model. Our working hypothesis is that transarticular screw with C1LMS+C2PS is equivalent to transarticular screw with C1LMS+C2ILS and that sublaminar wire technique is not as biomechanically stable as the other two techniques.

Biomechanical studies performed involving cadaver models have a significant anatomic variation that could affect the results. This is the main limitation of the cadaver model. Furthermore, changes in the loading profile caused by the application of instrumentation to the craniovertebral junction have not been reported. One of the main advantages of using the finite element method is its time efficiency and computational exactness. That is, parametric studies, whereby one variable at a time is altered, can be readily performed. Interspecimen and anatomic variabilities are eliminated. Investigation of fixation aspects which demonstrates the minimum degree of anatomic involvement while maintaining adequate fixation in order to promote fusion, could be extremely useful to both the designer and the clinician. This investigation examines which techniques afford the greatest mechanical stability. From this, it may be possible to identify which aspects of certain fixations are critical to their success and to implement
these aspects in current and future designs. Therefore, the following sections describe a mechanical investigation of the atlanto-axial junction using a cadaver model and a FE model with respect to the stabilization techniques.
Chapter 3

Materials and Methods

Finite Element (FE) Models have been widely used for simulating the kinematics and biomechanics of the human cervical spine at physiological, injured, or stabilized states. The practical difficulties, restrictions, and cost involved in experimental in vitro and in vivo studies besides ethical considerations have promoted the needs for using the computational models as complementary tools for studies in orthopedic biomechanics. The finite element (FE) technique allows the researcher to extract parameters which are experimentally unattainable for example, stresses and strains within the bone, ligamentous stress and strain, and load sharing among spinal elements. An additional limitation encountered in experimental studies is interspecimen variability. The FE method is characterized by absolute reproducibility, thus allowing model parameters to be modified.

3.1 Background

The first numerical spinal models were developed in the late 1950’s, to understand the biomechanics behind spinal injury due to pilot ejection. Since then, research with numerical techniques, such as the Finite Element Method (FEM), has made
progress towards a better understanding of spinal behavior and injury mechanics. The FEM makes it possible to incorporate realistic geometry of the spinal tissues and their physical properties, to evaluate spinal kinematics, kinetics, internal stresses and strains of the different spinal components. Lumped parameter models can also be used to study kinematics, but does not give the possibility to evaluate tissue stresses and strains since they are built up of rigid bodies. Knowing the stresses and strains in response to given load it is possible to predict the outcome, which is important for injury prevention. The anatomical details give a more reliable kinematical response than spherical representations of vertebrae and joints used in the lumped parameter models. The first Finite Element (FE) spinal models were of the lumbar and thoracic regions [79-81].

3.1.1 Cervical Motion Segment Models

FE models of cervical motion segments have been developed [55, 82-85] to study load distribution between the vertebrae and biomechanical effects secondary to impact. All the mentioned cervical models had linear elastic material characteristics. Kumaresan et.al [86] developed an anatomically detailed C4-C6 motion segment of the adult human and three age-specific one, three, and six year old models. The vertebrae were represented with shell elements for the cortical bone and solid elements for the trabecular bone. The intervertebral disc was modeled with annulus fibers and ground substance in combination with linear elastic solid elements for the nucleus pulposus, which is the first model to include anisotropic behavior of the disc. Non-linear spring elements were used to model the ligaments. In 1998, Goel et.al [87] implemented non-linear material properties
for the ligaments in an FE model of the C5-C6 complex. Validation was conducted for compression, flexion, extension, axial rotation, and lateral bending.

### 3.1.2 Models of Cervical Injury

Cervical spinal models were developed by several authors to study fracture fixation methods in the cervical spine.\(^{[78, 86, 88]}\)

### 3.1.3 The Upper Cervical Spine

Significant improvement has been made in the development of FE cervical spine models. While earlier FE cervical spine models consisted only one vertebra or up to two motion segments as mentioned earlier. Current FE models usually include complete anatomically detailed cervical spine. The advancement in computer and medical imaging technology has enabled increasingly more accurate geometric representation of the cervical spine, especially for the vertebrae and intervertebral discs. The new and improved material characterization data of the osseous and soft tissues allows the FE analysis to make more realistic predictions. FE models of the craniovertebral junction\(^{[78, 89, 90]}\) resulted in variations in the reported results. This might be because of the variation in the material properties assigned to the model or inclusion of the anatomical structures (e.g. ligament attachment sites) accurately. Till the last decade, the most anatomically detailed cervical model was developed by Kimpara et al.\(^{[91]}\) in order to study the anteroposterior head-neck responses during severe frontal impacts. A Head-Neck FE model with Brain-Spinal Cord Complex was developed with 49,579 elements and 12,586
elements defined, respectively, for the head and the neck. Musculature was modeled as tension-only bar element based on Hill-type muscle equation without activation.

The following section details the generation of the baseline craniovertebral junction model. Subsequent sections detail how this model was altered or varied to study the biomechanical aspects of stabilization of the atlantoaxial junction.

3.2 Model Geometry

A three-dimensional non-linear FE model of an intact C0-C3 spine (Figure 3.2) was created from CT scans of a female subject with no prior spinal disorders. Radiographs and dual-energy x-ray absorptiometry (DEXA) were used to confirm the integrity of the bony structure. The region of C0-C3 was scanned using CT at 1 mm thickness, producing transverse plane CT images. Further image processing was performed through the software ImageJ. For each CT image, the bony region was delineated and then divided into several quadrilateral shaped elements. The four nodes forming a particular element were digitized to obtain their X and Y coordinates with respect to the global axes system (Figure 3.1). The Z coordinate was computed based on the sequence of the corresponding transverse slice in the CT image data set and the slice thickness. The transverse cross sectional shape of the intact spine is symmetric about the mid-sagittal plane. Hence, only one-half of the bony regions in each image were digitized to create one half of the FE mesh, and the other half was reflected symmetrically in the finite element software. The coordinate data of the elements from adjacent cross sections were then combined to generate eight-node brick elements, thus, producing the three-dimensional meshes of the bony structure.
The intact model contained 121,721 nodes and 91,936 elements. The global coordinate system of the model was defined as follows: x was oriented positive from left to right in the lateral direction, y was oriented positive from anterior to posterior and z was positive downward axially along the spine (Figure 3.2).

Figure 3-1: Generation of the finite element mesh from CT data. Serial CT scans (a) were digitized for nodal coordinates. (b) The data were assigned connectivities thus producing three-dimensional elements. (c) Stacking of the elements resulted in the final FE mesh. The drawings above represent creation of a slice (shading in (c)) for the odontoid process. Note in (b) that the cancellous and cortical regions are delineated such that different material properties can be assigned for each area.
Figure 3-2: FE model mesh of the occiput (C0), atlas (C1) and axis (C2) shown from anterior (top) and lateral (bottom) viewpoints. The lateral view also depicts some of the ligaments.
3.2.1 Vertebral Body and Posterior Element

The vertebral body consists of a hard cortical shell which encloses the soft cancellous (porous) bone. The thickness of the cortical shell is taken as 0.5 mm. The posterior bony elements are joined to the vertebral body posterolaterally and all of them were given one value for elasticity modulus. The thickness of the endplate is taken as 0.5 mm. All the bone in this model was defined using three-dimensional eight-node brick elements.

3.2.2 Facet Joints

The facet surfaces were oriented at twenty five degrees from the transverse plane at C 1-2 level and forty five degrees at C 2-3 level. All the facet articulations were modeled using the ABAQUS ® (version 6.7). Six sliding articulations were incorporated into upper cervical (C0-C1-C2) spine: right and left occipito-atlantal joints, right and left atlanto-axial joints, odont o-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). The effect of cartilage on load transfer was taken into account by using the “softened” contact option available in ABAQUS. Also, the effect of synovial joint lubrication was incorporated in the model by assigning a coefficient of friction of 0.005 to each joint.
3.2.3 Intervertebral Discs

The cervical disc incorporated at C 2-3 level consists of anulus fibrosus and nucleus pulposus. The anulus fibrosus was modeled as a composite material with ground substance and fibrous components. Anulus fibers are oriented at +40° or -40° with respect to the transverse plane and occupy approximately 20% of the volume. The ground substance was modeled using eight-node brick elements. An ABAQUS ‘rebar’ option was used to simulate the fibers in the annulus such that the desired fiber volume fraction and direction was created. Further, a ‘no compression’ option was used which allows the fibers to transmit only tension. A total of 270 fiber-reinforced elements were used to simulate the annulus at each level. This allowed the generation of four layers of eight-node brick elements in the radial direction, each layer embedded with four layers of annular fibers in the radial direction, to fully simulate the intervertebral disc. The nucleus was modeled as an incompressible fluid using fluid elements to enclose the nucleus space.

3.2.4 Ligaments

The ligament attachment sites, area and orientations were determined from literature\cite{82, 85} and a previous model of the upper cervical spine generated by Puttlitz et al.\cite{78} Ligaments modeled in upper cervical spine were cruciform ligament, alar ligament, apical ligament, accessory ligament, nuchal ligament, tectorial membrane, anterior longitudinal ligament, occipito-atlantal capsular ligament and atlanto-axial capsular ligaments. The cruciform ligament was modeled as having two components: the transverse ligament and the superior/inferior fascicles.
All the ligaments except the transverse ligament were modeled using three-dimensional two-node truss elements allowing a “neutral zone” to be incorporated. The transverse ligament was modeled using eight-node brick elements such that the modeling of odontoid process-transverse ligament articulation could be represented accurately. In addition, the transverse ligament was modeled using a single elastic modulus, based on the work of Fielding et al.,[34] where they described this ligament as “a very firm, inelastic band of tissue.”

All the seven major ligaments were modeled in the lower level (C2-C3), including Anterior Longitudinal Ligament (ALL), Posterior Longitudinal Ligament (PLL), Interspinous Ligament (ISL), Capsular Ligament (CL), Ligamentum Flavum (LF).

### 3.3 Material Properties

The material properties for the lower cervical ligaments were taken from the literature and a previous model of a cervical spine functional unit generated by Clausen et al.[82] (Table 3.1). Cortical and cancellous bones were assigned Young’s moduli of 450MPa and 10,000 MPa, respectively. The nucleus pulposus was modeled as an incompressible fluid using fluid elements. The annulus fibrosus was modeled as a composite with the ground matrix having a modulus of 4.2 MPa and the fibers having a modulus of 450 MPa.

The Young’s modulus for all the ligaments (except transverse ligament in upper cervical spine) was defined over a range of strains to simulate their nonlinear stress-strain curve characteristics.
Table 3-1: Upper cervical spine finite element model characteristics, with element types and number of elements indicated. C3D8\(^*\) = linear, isotropic 8-node element (“brick”); T3D2\(^**\) = linear two-node element (no compression transmission, “cable”). ^Surface elements were internally generated from the existing mesh, thus they represent opposing faces of brick elements.

<table>
<thead>
<tr>
<th>Element Group Name</th>
<th>Number of Elements</th>
<th>Element Type</th>
<th>Young’s Modulus (MPa)</th>
<th>Poisson’s Ratio</th>
<th>Cross-Sectional Area (mm(^2))</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Bony Regions in Skull</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cortical Bone</td>
<td>~26,906</td>
<td>C3D8(^*)</td>
<td>15,000</td>
<td>0.3</td>
<td>-</td>
</tr>
<tr>
<td>Cancellous Bone</td>
<td>~13,453</td>
<td>C3D8(^*)</td>
<td>500</td>
<td>0.3</td>
<td>-</td>
</tr>
<tr>
<td>Teeth</td>
<td>32</td>
<td>C3D8(^*)</td>
<td>20,000</td>
<td>0.3</td>
<td>-</td>
</tr>
<tr>
<td><strong>Bony Regions in Spine</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Vertebral Cortical</td>
<td>18,664</td>
<td>C3D8(^*)</td>
<td>10,000</td>
<td>0.3</td>
<td>-</td>
</tr>
<tr>
<td>Vertebral</td>
<td>18,564</td>
<td>C3D8(^*)</td>
<td>450</td>
<td>0.25</td>
<td>-</td>
</tr>
<tr>
<td>Posterior Bone</td>
<td>784</td>
<td>C3D8(^*)</td>
<td>3,500</td>
<td>0.25</td>
<td>-</td>
</tr>
<tr>
<td><strong>Intervertebral Disc</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Annulus (Ground)</td>
<td>8,542</td>
<td>C3D8(^*)</td>
<td>4.2</td>
<td>0.45</td>
<td>-</td>
</tr>
<tr>
<td>Annulus Fibers</td>
<td>8,576</td>
<td>REBAR</td>
<td>450</td>
<td>0.3</td>
<td>-</td>
</tr>
<tr>
<td>Nucleus Pulposus</td>
<td>2,816</td>
<td>C3D8(^*)</td>
<td>-</td>
<td></td>
<td>-</td>
</tr>
<tr>
<td><strong>Joints</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Facet Joints</td>
<td>-</td>
<td>SURFACE</td>
<td>Soft contact</td>
<td></td>
<td>-</td>
</tr>
<tr>
<td><strong>Ligaments</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Upper Cervical</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Transverse</td>
<td>350</td>
<td>C3D8(^*)</td>
<td>20</td>
<td>0.3</td>
<td>18</td>
</tr>
<tr>
<td>Sup./inf. cruciform</td>
<td>27</td>
<td>T3D2(^**)</td>
<td>6.0(&lt;17%), 10.0(&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Alar</td>
<td>22</td>
<td>T3D2(^**)</td>
<td>3.0(&lt;17%), 8.5(&gt;17%)</td>
<td>0.3</td>
<td>22</td>
</tr>
<tr>
<td>Apical</td>
<td>4</td>
<td>T3D2(^**)</td>
<td>6.0(&lt;17%), 10.0 (&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Accessory</td>
<td>14</td>
<td>T3D2(^**)</td>
<td>6.0(&lt;17%), 10.0 (&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Nuchal</td>
<td>24</td>
<td>T3D2(^**)</td>
<td>12.0(&lt;17%),</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Tectorial Membrane</td>
<td>68</td>
<td>T3D2(^**)</td>
<td>6.3</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Anterior</td>
<td>31</td>
<td>T3D2(^**)</td>
<td>12.0(&lt;17%), 20.</td>
<td>0.3</td>
<td>6</td>
</tr>
<tr>
<td>Post. C0 /C1</td>
<td>11</td>
<td>T3D2(^**)</td>
<td>6.0(&lt;17%), 10.0 (&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Ant. C0 /C1</td>
<td>13</td>
<td>T3D2(^**)</td>
<td>6.0(&lt;17%), 10.0 (&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Post. C1 /C2</td>
<td>12</td>
<td>T3D2(^**)</td>
<td>6.0(&lt;17%), 10.0 (&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td>Ant. C1 /C2</td>
<td>22</td>
<td>T3D2(^**)</td>
<td>0.2(&lt;17%), 1.25 (&gt;17%)</td>
<td>0.3</td>
<td>5</td>
</tr>
<tr>
<td><strong>Lower Cervical (C2-C3)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Anterior</td>
<td>1,120</td>
<td>T3D2(^**)</td>
<td>18(&lt;17%), 30(&gt;17%)</td>
<td>0.3</td>
<td>33</td>
</tr>
<tr>
<td>Posterior</td>
<td>800</td>
<td>T3D2(^**)</td>
<td>12(&lt;17%), 20(&gt;17%)</td>
<td>0.3</td>
<td>33</td>
</tr>
<tr>
<td>Ligamentum</td>
<td>35</td>
<td>T3D2(^**)</td>
<td>3.5(&lt;30%), 10(&gt;30%)</td>
<td>0.3</td>
<td>50.1</td>
</tr>
<tr>
<td>Interspinous</td>
<td>28</td>
<td>T3D2(^**)</td>
<td>2.8(&lt;45%), 8(&gt;45%)</td>
<td>0.3</td>
<td>13.0</td>
</tr>
<tr>
<td>Capsular Ligament</td>
<td>166</td>
<td>T3D2(^**)</td>
<td>8.5(&lt;17%), 30(&gt;17%)</td>
<td>0.3</td>
<td>46.6</td>
</tr>
</tbody>
</table>
3.4 Intact Finite Element Model Validation

First, the model was compared with the experimental data obtained in the ECORE lab at The University of Toledo. The model was kinematically constrained for all nodes lying on the inferior surface of the C3 vertebra, the same way as in the *in vitro* experiments. Occipital moments of 2N-m were applied to the model, the same location and load magnitude used in the cadaveric investigation. The resultant in-plane motions for C0-C1, C1-C2 as well as C1-C3 were compared in extension/flexion, lateral bending and axial rotation. Second, the model was validated against experiments by Panjabi et.al.[24, 25] The model was subjected to moments of 1.5 Nm with C3 rigidly fixed and the loads were applied at the occiput for axial rotation, flexion, extension, and lateral bending. The predicted kinematic data under different loading configurations were analyzed and compared against different published data.

3.5 Investigation of the Stabilization Techniques

The investigation was based on:

1) Results from the cadaver testing comparing the stability of the three C1-2 transarticular screw salvaging fixation techniques which were used to validate the finite element predictions.
2) Finite element modeling of the fixation techniques to investigate the kinematics and load sharing properties of the stabilization techniques used in cadaver testing.

The following sections describe the biomechanical evaluation of the stabilization techniques using the FE approach and cadaver study.

3.5.1 Cadaver Testing of the Stabilization Techniques

In vitro cadaver testing was used to study the stability afforded by the stabilization techniques. The testing was carried out on nine cadaver upper cervical spine specimens.

3.5.1.1 Specimen Acquisition and Preparation

Nine fresh-frozen C0-C4 spine segments with ligaments intact were obtained. The specimens were sealed in double plastic bags and kept frozen at -20° until dissection and testing. The specimen was evaluated for bony deformity and alignment using anteroposterior and sagittal plane radiographs. Dual energy x-ray absorptometry (DEXA, Hologic QDR -1000 Bone Densimeter, Hologic, Inc. Waltham, MA.) was performed (settings: 140/70 kVp, 2.0 mA average, 60 Hz) on the specimen to assess bone quality and density. Both diagnostic techniques indicated that the specimen was devoid of gross deformity and possessed excellent bone density in the occiput and the cervical spine. Specimens were denuded of all paravertebral musculature leaving the ligaments structures and disc intact (Figure 3.3).
The C4 vertebral body of the specimen was potted in order to immobilize this part of the spine. The potting process involved passing three 2 in. wood screws through the vertebral body anteriorly into the body of C4 oriented in a manner such that they were directed slightly cephalad and angled approximately 30° from the midsagittal plane. These screws served to stabilize C4 in the resin like outriggers. The specimen was placed into an acrylic container (6 in. x 6 in. x 1.25 in.) that held four eyebolts (1/4 x 3 inches) protruding through the bottom of the container. These bolts were used to attach the potted specimen to the testing fixture in latter portions of the experimental testing. The potting material, Bondo® (Bondo/Mar-Hyde Corp., Atlanta, GA), was prepared by mixing liquid resin, solid body filler polymer, and cream hardener. This mixture was introduced into the container holding the specimen and allowed to polymerize for at least 20 minutes. Once the cement had hardened, the container was removed, leaving only the specimen coupled to the potted base and the threaded portion of the eyebolt screws protruding from the inferior aspect of the base (Figure 3.3).
Figure 3-3: Photographs showing the potted specimen in the testing frame (top) and the position of the occipital loading bars (bottom).
A loading apparatus was attached to the occiput. This was accomplished by passing threaded rods (1/4 inch diameter) through the occiput: one in the medial-lateral direction and one in the anteroposterior direction. The point of entry and exit for both rods is demonstrated in Figure 3.3. Four additional threaded rods (with internal threads) were then attached to the extended portions of the two rods drilled through the Occiput. Therefore, the total length of each rod measured approximately ten inches from the center of occiput body. These rods were later used as the point of load application to supply the cervical specimens with pure torsional loads. L-shaped brackets, each having three infrared light-emitting diodes (IRED’s) were attached to C0, C1, C2 and C3 vertebrae with wood screws. A special set of IREDs was placed on the immovable base to provide a global coordinate system of the experiment.

3.5.1.2 Kinematic Data Apparatus

The three-dimensional motion of the upper cervical spine was performed using a motion measurement system (Optotrak™; Northern Digital, Waterloo, Ontario, Canada). This system incorporates high resolution sensors that detect the three-dimensional position of a light emitting diode through triangulation. The IRED's are turned on and off in sequence by a strober unit permitting the system control unit (SCU) to identify which IRED is active at any point in time.
3.5.1.3 Specimen Testing

All the specimens were tested without compression. The prepared specimens were attached to the rigid base in a testing frame. Using a system of loading arms, pulleys and weights, quasi-static loads were applied to the skull leading to sequential pure moments of 0.5, 1.0, 1.5, 2.0 N m. Moments were applied to generate the following six loading modes: extension (EXT), flexion (FLEX), left and right lateral bending (LB, RB) and left and right axial rotation (LR, RR). To overcome the spine’s viscoelastic effects, the specimens were ranged maximally in all directions at least three times prior to initial testing. Additionally, after each load application, the system was allowed to stabilize for 30 seconds to minimize the creep before data collection. The positions of the LEDs were recorded six times for each of the aforementioned loading modalities: (1) initially when the specimen was unloaded in a neutral position; (2-5) thirty seconds after each of the four load increments creating moments of 0.5 N m each; and (6) when the loads were removed from the specimen. The specimens were frequently sprayed with normal saline to prevent drying of the tissue during testing. The specimens were tested in the following sequence:

1. Intact Spine. The intact spines were tested for load displacement behavior. This step generated baseline values which were used for later comparisons.

2. Destabilization and stabilization. The specimens were destabilized by creating type II odontoid fracture by first drilling multiple holes at the base of the odontoid and then connecting them with quarter inch osteotome. Due to severe instability
caused by the fracture it was difficult to perform motion measurements and was tested at the end.

After destabilization, each of the specimens was stabilized in the following order:

i. Transarticular C1-2 screw alone. (C1-C2 TS) (figure 3.4(a))

ii. Transarticular C1-C2 screw on one side in combination with C1 lateral mass and C2 pedicle screws connected with 3.5 mm titanium rod. (TS+C1LMS+C2PS) (figure 3.4(b))

iii. Transarticular C1-C2 screw in combination with C1 lateral mass and C2 intralaminar screws connected with 3.5 mm titanium rod. (TS+C1LMS+C2ILS) (figure 3.4(c))

iv. Transarticular C1-C2 screw supplemented by sublaminar wire. 1.0 mm diameter cables a total of two were used in each specimen. A modified Gallie’s technique of C1-2 wiring was used. Each cable was passed under C1 posterior arch then through a hole that was drilled in C2 spinous process. All the cables were tightened up to 15 inch-pound using a torque limiting handle. (TS+WIRE) (figure 3.4(d))
3.5.1.4 Reported Data

The spatial locations of the LEDs fixed to the vertebral bodies were recorded by the Optotrak™ motion measurement system. These spatial locations were transformed into angular rotations referenced to the base plate, using the rigid body transformation. The rotations of the vertebrae were plotted against the load applied to characterize the load-deformation behavior of the motion segment. The intersegmental rotation across C1-C2 segment for each of the intact and stabilized specimens was calculated. The stabilized
specimens were compared to the intact state for efficacies of the three techniques for stabilizing the atlanto-axial complex. Since the objective of the study was to determine the degree of stability afforded by the application of the stabilization techniques, all the stabilization data was normalized to the intact data. The percentage change in motion (change) due to the implantation of the hardware was calculated using the following formula:

\[
\text{Change} = \frac{\text{ROM}(\text{inst}) - \text{ROM}(\text{int})}{\text{ROM}(\text{int})}
\]

where ROM (\text{inst}) is the relative rotation between C1 and C2 in the stabilized case and ROM (\text{int}) is the relative rotation between C1 and C2 in the intact case. Thus, reductions and increases in motion due to stabilization are denoted by negative and positive values, respectively.

Statistical analysis was performed to find out any statistical differences between intact and the three instrumented specimens (Intact, TS+C1LMS+C2 PS, TS+C1LMS+C2ILS, TS+WIRE). Firstly, one-way repeated-measures analysis of variance (ANOVA) was performed on the data to see if there was any significant difference among the four sets at confidence level of 95%. Statistical Analysis was also done to see the effectiveness of TS alone technique when compared to intact specimen. Comparison among the three fixation techniques was done by performing ANOVA at confidence interval of 95%. Pair-wise comparison was performed to evaluate the differences between the three instrumented sets. For this purpose, two-tailed unpaired \(t\)-test was performed on the data.
3.5.2 Finite Element Investigation of the Stabilization Techniques

Model de stabilization was achieved by removing a horizontal layer of the elements from the region in which the odontoid process meets the body of the axis (C2), simulating a Type II odontoid fracture. Finite element models were developed to evaluate the effects of unilateral fixation (transarticular screw on the right side) and bilateral fixation techniques using three different options for fixing the contra-lateral atlanto-axial joint with transarticular screw fixed on one side (right side) on the destabilized model.

The destabilized C0-C3 model was modified to simulate the following instrumented models:

1) Transarticular Screw in the right atlanto-axial joint. (C1-C2 TS) (figure 3.5)

2) C1-2 Transarticular screw placed in the right atlanto-axial joint and C1 Lateral Mass Screw + C2 Pedicle Screw connected with 3.5 mm titanium rod on the other side. (TS+C1LMS+C2PS) (figure 3.6)

3) C1-2 Transarticular screw placed in the right atlanto-axial joint and C1 Lateral Mass Screw + C2 Intralaminar Screw connected with 3.5 mm titanium rod on the other side. (TS+C1LMS+C2ILS) (figure 3.7)

4) C1-2 Transarticular screw placed in the right atlanto-axial joint and C1-2 Sublaminar Wire. (TS+WIRE) (figure 3.8)

All the screws for atlanto-axial fixation (C1-C2 Transarticular Screw, C2 Pedicle Screw, C2 Intralaminar screw, C2 lateral Mass Screw), C1-C2 Sublaminar Wire and the connecting rods were modeled and meshed using ABAQUS 6.7. The screws were given a
constant cross section; therefore the threads were not modeled. All the four screws used in different models were coupled with the spine using “coupling” command in ABAQUS. This allowed simulation of “no screw loosening” scenario. The C1 lateral mass screws were connected to C2 pedicle and C2 intralaminar screws by curved rods of 3.5mm diameter (hexahedral mesh, 1700 elements for each rod). The rods were tied (“tie” command) to the screws to simulate a tight grip. All the screws and rods were assigned the material properties of titanium (young’s modulus=115GPa, Poisson’s ratio = 0.34. A modified Gallie’ technique of C1-2 wiring was used in this study. This technique involves passing of two 1mm diameter cables under C1 posterior arch, through a hole that was drilled in C2 spinous process on either side. The cables are tightened until the symmetric tension is achieved. This technique is simulated in the model by locating four nodes, two each on C1 arch and C2 spinous process which were in the area where the wires would be passed. The vertebrae were wired together by placing a cable element with the material property of 20-gauge stainless steel wire (E = 200 GPa, Poisson’s = 0.28, diameter = 1 mm) between the C1 and C2 nodes on the left and right sides.

3.5.2.1 Boundary and loading conditions for instrumented models

The applied boundary conditions and loadings were consistent for all the models, including the intact. The model was constrained at all nodes located on the inferior surface of C3 vertebra. All the models were run in flexion (FLEX), extension (EXT), left/right bending (LB/RB) and left/right axial rotation (LR/RR) by applying pure moments of 2.0 Nm to the occiput.
Figure 3-5: Simulation of the unilateral Transarticular Screw (C1-C2 TS) at C1-2 in the C0-C3 cervical spine model.
Figure 3-6: Simulation of the Transarticular Screw on one side and a combination of C1 Lateral Mass Screw and C2 Pedicle Screw (TS+C1LMS+C2PS) on the other side in the C0-C3 cervical spine model.
Figure 3-7: Simulation of the Transarticular Screw on one side and a combination of C1 Lateral Mass Screw and C2 Intralaminar Screw (TS+C1LMS+C2ILS) on the other side in the C0-C3 cervical spine model.
Figure 3-8: Simulation of the Transarticular Screw on one side and C1-C2 Sublaminar Wire (TS+WIRE) on the other side in the C0-C3 cervical spine model.
3.5.2.2 Reported Data

Kinematic predictions were reported for all the intact and instrumented cases. A range of motion (ROM) at a particular level is defined as the relative angle change between two adjacent vertebrae. Changes from intact rotations were normalized to intact case by following:

\[
\text{Change} = \frac{\text{ROM}(\text{inst}) - \text{ROM}(\text{int})}{\text{ROM}(\text{int})}
\]

where \( \text{ROM} \) (\( \text{inst} \)) is the relative rotation between C1 and C2 in the stabilized case and \( \text{ROM} \) (\( \text{int} \)) is the relative rotation between C1 and C2 in the intact case. Thus, reductions and increases in motion due to stabilization are denoted by negative and positive values, respectively.

Joint contact force transmissions for the facet articulations (right and left C0–C1 facets, right and left C1–C2 facets) were determined to assess the degree of load transfer caused by the various models. Maximal Von Mises stresses in the instrumentation for each loading and model variant were also reported. The joint forces and the Von Mises stresses calculated were compared between the intact and the instrumented cases.
Chapter 4

Results

4.1 Kinematic Validation of Intact Finite Element Model

The model was first validated by comparing its kinematic predictions to cadaveric data obtained previously at E-CORE lab under 2.0 N-m of pure moment loading. Table 4.1 shows the model predictions and associated cadaver data. All FE model motion prediction data, due to loading in the sagittal (flexion and extension) frontal planes (lateral bending) and axial rotation fell within the standard deviation of the cadaveric data. Also, the FE model predicted ROM by applying 1.5 N m moments to the occiput was compared with previously published data (Tables 4.2-4.6). The ROMs correlated well with the experimental studies by Panjabi et al\textsuperscript{[25]}, Moroney et al\textsuperscript{[27]} and also some of the in-vivo studies for all the three levels. Apart from slight difference in rotations of axial rotation at the C1-C2 level, the model also correlated well with the recommended data by Panjabi et al\textsuperscript{[25]}. 
Table 4-1: Validation of model kinematic predictions compared with cadaveric data (mean ± standard deviation) for 2.0 N-m occipital moment loading. Units for all data reported in degrees.

<table>
<thead>
<tr>
<th>Applied Moment</th>
<th>C0-C1</th>
<th>C1-C2</th>
<th>C2-C3</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Cadaver Data</td>
<td>Model Prediction</td>
<td>Cadaver Data</td>
</tr>
<tr>
<td>EXT</td>
<td>19.4 ± 6.6</td>
<td>19.3</td>
<td>11.4 ± 8.6</td>
</tr>
<tr>
<td>FLEX</td>
<td>3.4 ± 2.8</td>
<td>5.4</td>
<td>11.7 ± 3.2</td>
</tr>
<tr>
<td>LB</td>
<td>4.6 ± 2.1</td>
<td>3.4</td>
<td>5.7 ± 3.9</td>
</tr>
<tr>
<td>RB</td>
<td>7.0± 4.2</td>
<td>3.3</td>
<td>5.6± 2.5</td>
</tr>
<tr>
<td>LR</td>
<td>6.1± 2.5</td>
<td>8.2</td>
<td>24.8± 8.5</td>
</tr>
<tr>
<td>RR</td>
<td>6.3± 2.9</td>
<td>6.8</td>
<td>23.3± 9.6</td>
</tr>
</tbody>
</table>

The range-of-motion predicted by the FE model (Table 4.1), obtained by applying 1.5 N-m moments on the occiput, was comparable to in vivo and in vitro published data (Tables 4.2- 4.6). Panjabi et al.\cite{92} have recommended ranges of motion at the levels of the upper cervical spine (Tables 4.4 - 4.6). The C0-C1 and C1-C2 motion for all three motion types, is taken from Panjabi et al.\cite{25} The C2-C3 flexion-extension ranges of motion are taken from Dvorak et al.\cite{23} Using this data, they have computed the upper limits (mean value plus 2 standard deviations) and lower limits (mean value minus 2 standard deviations) for all spinal levels and motion types (Tables 4.4 - 4.6). From the viewpoint of the injury causing motions, they interpreted the lower and upper limits in the following manner. The lower limit of a range of motion represents the motion at a specific intervertebral level that would not cause injury in most people. The upper limit is the threshold motion if the spine is bent or twisted beyond this threshold, there will be injury.
in most persons. Therefore, the model predicted motions were also compared with those ranges of motions reported by Panjabi et.al\textsuperscript{[92]} (Tables 4.4 - 4.6).

### Table 4-2: Range-of-motion (degrees) comparison of the data reported in the literature and the FE Model predictions (shown in bold) for C0-C1, C1-C2 and C2-C3 in Extension under 1.5Nm

<table>
<thead>
<tr>
<th>Extension</th>
<th>Panjabi\textsuperscript{[25]}</th>
<th>Panjabi\textsuperscript{[24]}</th>
<th>Moroney\textsuperscript{[27]}</th>
<th>Present Study</th>
</tr>
</thead>
<tbody>
<tr>
<td>C0-C1</td>
<td>21.0±6.0</td>
<td>-</td>
<td>-</td>
<td>18.13</td>
</tr>
<tr>
<td>C1-C2</td>
<td>10.9±3.5</td>
<td>-</td>
<td>-</td>
<td>10.05</td>
</tr>
<tr>
<td>C2-C3</td>
<td>-</td>
<td>3.6±1.2</td>
<td>3.5±1.9</td>
<td>3.82</td>
</tr>
</tbody>
</table>

### Table 4-3: Range-of-motion (degrees) comparison of the data reported in the literature and the FE Model predictions (shown in bold) for C0-C1, C1-C2 and C2-C3 in Flexion under 1.5Nm.

<table>
<thead>
<tr>
<th>Flexion</th>
<th>Panjabi\textsuperscript{[25]}</th>
<th>Panjabi\textsuperscript{[24]}</th>
<th>Moroney\textsuperscript{[27]}</th>
<th>Present Study</th>
</tr>
</thead>
<tbody>
<tr>
<td>C0-C1</td>
<td>3.5±1.9</td>
<td>-</td>
<td>-</td>
<td>3.79</td>
</tr>
<tr>
<td>C1-C2</td>
<td>11.5±6.3</td>
<td>-</td>
<td>-</td>
<td>10.83</td>
</tr>
<tr>
<td>C2-C3</td>
<td>-</td>
<td>6.3±1.2</td>
<td>5.6±1.8</td>
<td>7.38</td>
</tr>
</tbody>
</table>
Table 4-4: Comparison of flexion-extension combined range-of-motion (degrees) data reported in the literature and the FE Model predictions (shown in bold) for C0-C1, C1-C2 and C2-C3 under 1.5Nm.

<table>
<thead>
<tr>
<th>Flexion+Extension</th>
<th>C0-C1</th>
<th>C1-C2</th>
<th>C2-C3</th>
</tr>
</thead>
<tbody>
<tr>
<td>penning[20]</td>
<td>30.0±5.0</td>
<td>30.0±5.0</td>
<td>12.0±2.8</td>
</tr>
<tr>
<td>Dvorak[22]</td>
<td></td>
<td>12.0±4.0</td>
<td>10.0±3.0</td>
</tr>
<tr>
<td>Dvorak[26]</td>
<td>-</td>
<td>15.0±3.0</td>
<td>12.0±2.0</td>
</tr>
<tr>
<td>Panjabi[24]</td>
<td>-</td>
<td>-</td>
<td>9.9±1.2</td>
</tr>
<tr>
<td>Panjabi[25]</td>
<td>24.5±4.0</td>
<td>22.4±4.7</td>
<td>-</td>
</tr>
<tr>
<td>White and panjabi[4]</td>
<td>25.0</td>
<td>20.0</td>
<td>10.0±2.8</td>
</tr>
<tr>
<td>Recommended by Panjabi et.al [92]</td>
<td>24.5±4.0</td>
<td>22.4±4.7</td>
<td>12.0±2.0</td>
</tr>
<tr>
<td>Lower limit/upper limit[92]</td>
<td>16.5/32.5</td>
<td>13.0/31.8</td>
<td>8.0/16.0</td>
</tr>
<tr>
<td>Present Study</td>
<td>21.92</td>
<td>20.88</td>
<td>11.2</td>
</tr>
</tbody>
</table>

Table 4-5: Range-of-motion (degrees) comparison of the data reported in the literature and the FE Model predictions (shown in bold) for C0-C1, C1-C2 and C2-C3 in lateral bending under 1.5Nm.

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>C0-C1</td>
<td>5.5±2.5</td>
<td>-</td>
<td>5.0</td>
<td>5.5±2.5</td>
<td>0.5/10.5</td>
<td>5.26</td>
</tr>
<tr>
<td>C1-C2</td>
<td>6.7±4.4</td>
<td>-</td>
<td>5.0</td>
<td>6.7±4.4</td>
<td>-2.1/15.5</td>
<td>5.35</td>
</tr>
<tr>
<td>C2-C3</td>
<td>-</td>
<td>4.7±3.0</td>
<td>10.0±2.3</td>
<td>4.0±2.9</td>
<td>-1.9/9.8</td>
<td>3.82</td>
</tr>
</tbody>
</table>
Table 4-6: Range-of-motion (degrees) comparison of the data reported in the literature and the FE Model predictions (shown in bold) for C0-C1, C1-C2 and C2-C3 in axial rotation under 1.5Nm.

| Axial Rotation (one side) | Panjabi\(^{[25]}\) | Moroney\(^{[27]}\) & white & Panjabi\(^{[4]}\) | Recommended ranges\(^{[92]}\) | Lower limit/Upper limit\(^{[92]}\) | Present Study |
|--------------------------|-------------------|-------------------|-------------------|-------------------|-------------------|----------------|
| C0-C1 7.3±2.2 5.0 7.3±2.2 | 2.9/11.7 | 8.22 |
| C1-C2 38.9±5.4 - 38.9±5.4 | 28.1/49.7 | 30.3 |
| C2-C3 - 1.9±0.7 3.0±2.5 | -2.0/8.0 | 2.72 |

From the comparisons showed above the FE model reasonably represents, within anatomic variation, “normal” range-of-motion in flexion, extension, lateral bending and axial rotation modes.

4.2 Analyses of Stabilization Techniques

Investigation of the stabilization techniques was evaluated using a cadaver model and by application to the FE model. The following sections describe the results found using these two techniques.

4.2.1 Cadaver Experiment Results

The following is a list of scenarios that were tested with the cadaver model:

(1) Intact
(2) Injured (osteotome cut of the base of the odontoid process replicating a Type II fracture) and stabilized with the three constructs

(3) Transarticular Screw in the right atlanto-axial joint. (C1-C2 TS)

(4) C1-2 Transarticular screw placed in the right atlanto-axial joint and C1 Lateral Mass Screw + C2 Pedicle Screw connected with 3.5 mm titanium rod on the other side (TS+C1LMS+C2PS)

(5) C1-2 Transarticular screw placed in the right atlanto-axial joint and C1 Lateral Mass Screw + C2 Intralaminar Screw connected with 3.5 mm titanium rod on the other side (TS+C1LMS+C2ILS)

(6) C1-2 Transarticular screw placed in the right atlanto-axial joint and C1-2 Sublaminar Wire (TS+WIRE)

Each scenario was evaluated in flexion (FLEX), extension (EXT), right lateral bending (RB), left lateral bending (LB), right axial rotation (RR), and left axial rotation (LR). The data was outputted in the form of moment-rotation relations. An example of the pure data is shown in Figure 4.1. The results of the cadaver experiment are shown in Figure 4.2 for C1-C2 level.
Figure 4-1: An example of the load-displacement behavior of the intact specimen recorded during the cadaver model testing under the influence of extension/flexion, lateral bending and axial rotation moments. This data illustrates the application of four equal rotational moments of 0.5 Nm, resulting in a maximal load of 2.0 Nm.
Motion at the C1-C2 level is shown in Figure 4.1, with data normalized to the intact motion shown in Table 4.7. The data indicated that the unilateral stabilization of the spine (C1-C2 TS) produced significant reductions in lateral bending and axial rotation modes except an increase of 7.6% was seen in extension and a decrease of 39% in flexion which is not as significant as compared to the lateral bending and axial rotation modes. Right and left lateral bending demonstrated motion reductions of 82.3% and 70.5% respectively where as right and left axial rotations showed motion reductions of 50.1% and 66.8% respectively as compared to the intact case.

Statistical Analysis was also performed as mentioned in the previous chapter to find out the efficiency of unilateral transarticular screw when compared to intact. Data
(Table 4.8) showed that the unilateral transarticular screw restricted motion effectively in right axial rotation ($p=0.005$), left axial rotation ($p=0.03$), left lateral bending ($p=0.016$) and right lateral bending ($p=0.049$) when compared to intact. However, in extension ($p=0.54$) and flexion ($p=0.08$), unilateral procedure did not result in higher stiffness than the intact spine.

Stabilization of the spine bilaterally with all the three instrumentation techniques (TS+C1LMS+C2PS, TS+C1LMS+C2ILS, TS+WIRE) resulted in greater motion reductions in extension, flexion and axial rotation when compared to unilateral procedure alone (C1-C2 TS). Lateral bending motion was moderately reduced when compared to unilateral fixation. When comparing the three bilateral stabilization techniques, TS+C1LMS+C2PS and TS+C1LMS+C2ILS have similar motion reductions across C1-C2 level in extension (53.9% and 50.8%), flexion (73.1% and 73.3%), left lateral bending (89.3% and 83.4%), right lateral bending (84.4% and 80.1%), left axial rotation (92.7% and 94.7%) and right axial rotation (89.7% and 90.7%). TS+WIRE resulted in comparable reduction of motion in extension (30.3%), flexion (76.6%), left lateral bending (80.3%), right lateral bending (81.1%) and a minimal decrease in left axial rotation (74.7%) and right axial rotation (73.6%) motions when compared to the other bilateral fixation techniques (Table 4.7).
Table 4-7: Results of the in vitro cadaver test normalized to the intact, uninjured spine for C1-C2. Negative values represent reductions in motion, positive values represent increases in motion.

<table>
<thead>
<tr>
<th>Rotation</th>
<th>C1-C2 TS</th>
<th>TS+C1LMS+C2PS</th>
<th>TS+C1LMS+C2ILS</th>
<th>TS+WIRE</th>
</tr>
</thead>
<tbody>
<tr>
<td>EXTENSION (EXT)</td>
<td>7.60</td>
<td>-53.99</td>
<td>-50.82</td>
<td>-30.35</td>
</tr>
<tr>
<td>FLEXION (FLEX)</td>
<td>-39.19</td>
<td>-73.17</td>
<td>-73.37</td>
<td>-76.68</td>
</tr>
<tr>
<td>LEFT BENDING (LB)</td>
<td>-70.58</td>
<td>-89.38</td>
<td>-83.43</td>
<td>-80.33</td>
</tr>
<tr>
<td>RIGHT BENDING (RB)</td>
<td>-82.37</td>
<td>-84.43</td>
<td>-80.15</td>
<td>-81.12</td>
</tr>
<tr>
<td>LEFT ROTATION (LR)</td>
<td>-66.81</td>
<td>-92.72</td>
<td>-94.77</td>
<td>-74.73</td>
</tr>
<tr>
<td>RIGHT ROTATION (RR)</td>
<td>-50.12</td>
<td>-89.71</td>
<td>-90.73</td>
<td>-73.67</td>
</tr>
</tbody>
</table>
Table 4-8: *P* values calculated from comparison (One-Way Repeated-Measures analysis of variance (ANOVA), at 95% confidence interval) of the unilateral instrumented case with intact and the three bilateral instrumented cases for C1-C2 motion under pure moment of 2Nm. Numbers in bold (*p*<0.05) indicate a statistically significant difference among the sets compared.

<table>
<thead>
<tr>
<th>Loading condition</th>
<th>Intact Vs. C1-C2 TS</th>
<th>C1-C2 TS Vs. Bilateral techniques</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>(TS+C1LMS+CPS, TS+CLMS+C2ILS, TS+WIRE)</td>
</tr>
<tr>
<td>EXT</td>
<td>0.54</td>
<td><strong>0.05</strong></td>
</tr>
<tr>
<td>FLEX</td>
<td>0.08</td>
<td><strong>0.001</strong></td>
</tr>
<tr>
<td>LB</td>
<td>0.01</td>
<td>0.27</td>
</tr>
<tr>
<td>RB</td>
<td>0.04</td>
<td>0.89</td>
</tr>
<tr>
<td>LR</td>
<td>0.03</td>
<td><strong>0.003</strong></td>
</tr>
<tr>
<td>RR</td>
<td><strong>0.005</strong></td>
<td><strong>0.001</strong></td>
</tr>
</tbody>
</table>

Data from statistical analysis (Table 4.8) showed that the combination of TS with other techniques (C1LMS+C2PS, C1LMS+C2ILS, Wire) provided significant stability in extension / flexion (*p*=0.05/*p*=0.001) and left/right axial rotation (*p*=0.003/*p*=0.01) and equally stable in left/right lateral bending (*p*=0.27/*p*=0.89) when compared to unilateral procedure. When pairwise comparisons among the three stabilization techniques were done, results (Table 4.9) demonstrated that TS+C1LMS+C2PS provided the same stability as that of TS+C1LMS+C2ILS techniques in extension (*p*=0.8), flexion (*p*=0.6), left lateral bending (0.27), right lateral bending (0.44), left axial rotation (0.49) and right axial rotation (0.61). TS+C1LMS+C2PS provided the same stability as that of TS+WIRE
in extension (p=0.75), flexion (0.51), left lateral bending (p=0.22), right lateral bending (p=0.58) and showed significant stability in left axial rotation (p=0.009) and right axial rotation (p=0.01) motions. TS+C1ILMS+C2ILS resulted in same stability as that of TS+WIRE in extension (p=0.19), flexion (p=0.93), left lateral bending (p=0.69), right lateral bending (p=0.84) and showed statistically significant stability in left axial rotation (p=0.03) and right axial rotation (p=0.01).

**Table 4-9:** P values calculated from pair wise comparison (two-tailed unpaired t-test) of the instrumented cases for C1-C2 motion under pure moment of 2Nm. Numbers in bold (p<0.05) indicate a statistically significant difference among the sets compared.

<table>
<thead>
<tr>
<th>Loading condition</th>
<th>TS+C1ILMS+C2PS Vs. TS+C1ILMS+C2ILS</th>
<th>TS+C1ILMS+C2PS Vs. TS+WIRE</th>
<th>TS+C1ILMS+C2ILS Vs. TS+WIRE</th>
</tr>
</thead>
<tbody>
<tr>
<td>EXT</td>
<td>0.81</td>
<td>0.75</td>
<td>0.916</td>
</tr>
<tr>
<td>FLEX</td>
<td>0.592</td>
<td>0.513</td>
<td>0.938</td>
</tr>
<tr>
<td>LB</td>
<td>0.273</td>
<td>0.224</td>
<td>0.69</td>
</tr>
<tr>
<td>RB</td>
<td>0.442</td>
<td>0.586</td>
<td>0.843</td>
</tr>
<tr>
<td>LR</td>
<td>0.499</td>
<td><strong>0.009</strong></td>
<td><strong>0.038</strong></td>
</tr>
<tr>
<td>RR</td>
<td>0.611</td>
<td><strong>0.017</strong></td>
<td><strong>0.01</strong></td>
</tr>
</tbody>
</table>
4.2.2 FE Model Contact Force Transmission and Kinematic Predictions

The model was used to investigate various fixation techniques: unilateral (C1-C2 transarticular screw) vs. bilateral (combination of transarticular screw with either of C1 lateral mass + C2 pedicle, C1 lateral mass + C2 intralaminar or C1-2 sublaminar wire) and also three different bilateral cervical spine fixation techniques. The FE prediction of C1-C2 motion data is shown in Figure 4.3 and the normalized data with respect to the intact is given in Table 4.10.

**Figure 4-3:** Finite element model predicted C1-C2 Range of motion for under the moment of 1.5 Nm in all the loading modes.
Table 4-10: Finite element results of the C1-C2 range of motion normalized to the intact, uninjured spine. Negative values represent reductions in motion.

<table>
<thead>
<tr>
<th>Rotation</th>
<th>C1-C2 TS</th>
<th>TS+C1LMS+C2PS</th>
<th>TS+C1LMS+C2ILS</th>
<th>TS+WIRE</th>
</tr>
</thead>
<tbody>
<tr>
<td>extension (EXT)</td>
<td>-13.14</td>
<td>-41.51</td>
<td>-47.32</td>
<td>-43.67</td>
</tr>
<tr>
<td>flexion (FLEX)</td>
<td>-47.35</td>
<td>-82.96</td>
<td>-82.82</td>
<td>-76.72</td>
</tr>
<tr>
<td>left bending (LB)</td>
<td>-70.41</td>
<td>-98.28</td>
<td>-97.89</td>
<td>-98.37</td>
</tr>
<tr>
<td>right bending (RB)</td>
<td>-73.35</td>
<td>-97.94</td>
<td>-97.50</td>
<td>-95.93</td>
</tr>
<tr>
<td>left rotation (LR)</td>
<td>-63.34</td>
<td>-96.45</td>
<td>-96.46</td>
<td>-77.96</td>
</tr>
<tr>
<td>right rotation (RR)</td>
<td>-64.21</td>
<td>-96.45</td>
<td>-96.65</td>
<td>-76.15</td>
</tr>
</tbody>
</table>

The contact force transmission data were reported for all the facet articulations in extension (Figure 4.4), flexion (Figure 4.5), left lateral bending (Figure 4.6), right lateral bending (Figure 4.7), left axial rotation (Figure 4.8) and right axial rotation (Figure 4.10). Both unilateral transarticular C1-C2 screws and the three bilateral stabilization techniques reduced all contact forces in flexion at both C0-C1 (3%-30%) and C1-C2 levels (40%-90%) when compared to intact. In extension, unilateral instrumentation (right joint) increased both right (22%) and left C0-C1 (3%) contact forces while maintaining the left C1-C2 contact force nearly constant when compared to the intact. The right C1-C2 facet experienced a decrease (44% of intact) in the contact force with unilateral instrumentation due to the load transmission through the transarticular screw. All the bilateral techniques decreased the contact forces through C0-C1 (ranges from 9%-28%...
for different techniques) and C1-C2 (ranging from 65%-89% for different techniques) joints when compared to intact.

Right/left lateral bending contact force transmission results indicated that bilateral instrumentation is capable of maintaining contact at the contralateral left/right C1-C2 articulation. This contact transmission was not demonstrated for the unilateral or intact cases, indicating loss of articulation contact, or "lift off" of the left C1 inferior facet from the left C2 superior facet. However, when bilateral instrumentation was used, the contact forces at C0-C1 facet joints decreased about 5-25% of the intact in all the loading modes except for an increase seen in axial rotation case which went up to 50%.

All the three bilateral fixation techniques behaved similarly with respect to the force transmission in extension, flexion and lateral bending motions. However, the transmission of forces through the C1-C2 right/left facet in left/right axial rotations in TS+WIRE technique is more when compared to the other two techniques confirming the kinematic FE prediction of markedly less reduction of motion in axial rotation at C1-C2.
Figure 4-4: Contact force transmission effect for unilateral vs. three bilateral fixations. Rotation moment of 1.5 N-m applied.

Figure 4-5: Contact force transmission effect for unilateral vs. three bilateral fixations. Rotation moment of 1.5 N-m applied.
**Figure 4-6:** Contact force transmission effect for unilateral vs. three bilateral fixations. Rotation moment of 1.5 N-m applied.

**Figure 4-7:** Contact force transmission effect for unilateral vs. three bilateral fixations. Rotation moment of 1.5 N-m applied.
Figure 4-8: Contact force transmission effect for unilateral vs. three bilateral fixations. Rotation moment of 1.5 N-m applied.

Figure 4-9: Contact force transmission effect for unilateral vs. three bilateral fixations. Rotation moment of 1.5 N-m applied.
4.2.2.1 Instrumentation Stress Analysis

The magnitudes of the maximum von Mises stresses for all the stabilization techniques (C1-C2 TS, TS+1LMS+C2PS, TS+1LMS+C2ILS and TS+WIRE) are given in (Table 4.11). Data shows that the maximum stress in bilateral instrumentation was less than the maximum stress in the unilateral instrumentation in all the loading conditions. Also, left axial rotation produced the highest stresses in both the unilateral and bilateral hardware systems. The hardware experienced same maximal stresses when intralaminar screws (TS+1LMS+C2ILS) and pedicle screws (TS+1LMS+C2PS) techniques were used as opposed to the wire (TS+WIRE).

Table 4-11: Finite element model predicted values of Maximum Von Mises stresses (MPa) for the different instrumented models in all the loading modes.

<table>
<thead>
<tr>
<th>Rotation</th>
<th>C1-C2 TS</th>
<th>TS+1LMS+C2PS</th>
<th>TS+1LMS+C2ILS</th>
<th>TS+WIRE</th>
</tr>
</thead>
<tbody>
<tr>
<td>EXT</td>
<td>140.60</td>
<td>60.47</td>
<td>56.60</td>
<td>92.10</td>
</tr>
<tr>
<td>FLEX</td>
<td>148.27</td>
<td>53.20</td>
<td>49.33</td>
<td>92.10</td>
</tr>
<tr>
<td>LB</td>
<td>102.80</td>
<td>40.13</td>
<td>43.03</td>
<td>66.20</td>
</tr>
<tr>
<td>RB</td>
<td>135.67</td>
<td>53.53</td>
<td>56.07</td>
<td>88.45</td>
</tr>
<tr>
<td>LR</td>
<td>185.67</td>
<td>91.47</td>
<td>86.80</td>
<td>115.97</td>
</tr>
<tr>
<td>RR</td>
<td>196.13</td>
<td>69.40</td>
<td>72.93</td>
<td>120.40</td>
</tr>
</tbody>
</table>
Chapter 5

Discussion

The findings of the various investigations contained in this thesis represent some novel data with respect to atlanto-axial biomechanics. Some of the findings confirm earlier reports found in the literature, while others contradict findings of reported studies. The following serves to discuss the findings contained herein and to compare these results to those reported in the literature.

5.1 Goals of this study

The goals of this research project were enumerated in the introduction. A reiteration of these goals seems appropriate such that the following discussion of the results center on whether these findings attained the stated goals and/or raised further research questions. The goals of this study were to perform: (1) Biomechanical investigation of three atlanto-axial transarticular salvaging fixation techniques (2) Development of a finite element model of an upper cervical spine to study the effect of three atlanto-axial transarticular salvaging fixation techniques on the kinematics of upper cervical spine.
5.2 Assumptions and Limitations of the Methods Employed

Modeling of the body is very difficult due to the complex, dynamic nature of human anatomy and physiology. Thus, modeling of any human system inherently results in shortcomings that are not evident in the true physiological system. These shortcomings are usually presented in the form of modeling assumptions and limitations. Such factors need to be pinpointed and the results of any study should be kept within their context. Both in vitro and modeling methods employed in this study contain limitations and assumptions that are unique and/or common to either method. The following sections describe these shortcomings.

5.2.1 Limitations of the Cadaver Model

The cadaver results from our own in vitro study (a total of nine specimens) do not include the effect of muscle forces and compressive preload. Some of the more prominent roles of the muscles in the craniovertebral junction are to initiate and provide motion and stabilize the spine. Certainly, these roles are not viable in a cadaver specimen. However, application of occipital moments serves to grossly simulate the motion contribution of the muscles. Also, by testing the spine in the absence of muscular stabilization, the evaluation of the stabilization techniques can be thought of as occurring under less than ideal conditions. That is, the spine was evaluated as to the isolated contribution of the stabilization hardware and the associated ligaments. Finally, all the stabilization data was compared to the intact case. The intact data reflected motion of the spine in the absence of musculature contributions as well. Thus, a truly comparative study as to the role of the stabilization hardware was achieved.
5.2.2 Limitations of the Finite Element Model

The geometry was obtained from a single subject. The fear was that the subject did not reflect an anatomical “norm” (if one exists). The process of CT scanning, determination and tracing of osseous borders and digitization of the CT data could have potentially led to model inaccuracies, particularly since there existed noticeable blurring of the osseous CT images. In addition, the high degree of refinement (0.5 mm thick sections) used for the generation of the model produced blurring of the outputted CT images. Although there is likely more of a transition in material property from cortical to cancellous bone, this transition is not easily mimicked in the finite element model. Thus, the differing sections of the model were limited to one modulus definition (i.e. cortical bone was given a modulus of 10,000 MPa and a Poisson’s ratio of 0.3). Also, the endplate was specified to be 0.5 mm thick throughout the model, although realistically this thickness probably varied from segment to segment in the cadaver specimen. The vertebral cortical shell was also approximated to range from 0.3-0.5 mm in thickness throughout the model, based on literature values.\[^{93}\] Ironically, while the visual image of the model may reflect a lack of refinement, it was the relatively high degree of CT section refinement that indirectly produced this effect. The net effect, with regards to model stress and load predictions, was minor. In fact, during the course of this investigation, not once did a review of the von Mises stress contour plots indicate discontinuity due to what might be perceived as a lack of a smooth surface.

Probably the greatest simplification and assumption made during the modeling process involved the cross-sectional and material property assignment of the ligaments.
While the elastic moduli of the alar and transverse ligaments were calculated from reported data, the remaining ligaments were assigned properties from an earlier model of the lower cervical spine. The inherent assumption of this process is that there exists a continuity of material properties between the ligature of the upper and subaxial cervical spines. In addition, the cross-sectional properties were also taken from the pre-existing lower cervical FE model. Agreement between kinematic FE predictions and cadaveric in vitro data (Table 4.1) affirmed that the material property magnitudes and strain excursion profiles closely resembled the physiological values. The agreement existed for occipital loading at 1.5 Nm.

The transverse ligament was modeled using three-dimensional “brick” elements. All other elements were modeled using three-dimensional cable elements. This transverse ligament modeling deviation was requisite for inclusion of dens – transverse ligament contact modeling. Observation of the transverse ligament in dissection reveals that it represents a broad band of tissue and thus the use of brick elements does not deviate significantly from what is found in vivo. The transverse ligament was modeled in such a manner as to not support compressive loads, a property that the cable elements inherently display. The validation data seems to support the opinion that the modeling of the transverse ligament using brick elements did not significantly affect the motion characteristics of the model. In addition to geometric and material property assumptions, boundary and loading conditions for the model influence the mechanical behavior of the spine. Translations, rotations, stresses, and strains are all dictated by the loads and boundary conditions applied to the model. Precisely mimicking the diverse in vivo loading scenarios in a finite element model which does not simulate musculature
effects is impossible. The cadaver results from our own *in vitro* studies (a total of nine specimens) do not include the effect of muscle forces and compressive load. The finite element cases carried out in this study were under “no preload” and no musculature conditions (under pure moment of up to 2.0 Nm) to mimic the cadaver experiments. Thus, the instrumentation stresses and the contact forces reported in this study might differ under the presence of compressive load and musculature forces.

5.3 Intact Finite Element Model Behavior

The intact model kinematic behavior was evaluated. This was done to determine if the model could be considered valid with respect to predicting normal physiologic motion. Also, the model range-of-motion values were compared to those found in the literature.

5.3.1 Validation of the Finite Element Model

Validation of the model was accomplished by comparison of model kinematic predictions with cadaveric data at 2.0 Nm pure moment loading. In addition, the model range of motion for flexion, extension, and lateral bending was obtained by the application of 1.5 N-m loading. The ranges of motion predicted by the model fell within the physiologic range of motion determined by earlier published studies using both *in vivo* and *in vitro* methods. These findings are also in agreement with the work of Panjabi et al. in which they determined that 1.5 N-m occipital moments produced physiologic range of motion.
5.4 Investigation of Stabilization Hardware

The evaluation of the instrumentation techniques employed in this study was performed by using two investigational models: a cadaver model and a FE model. The finite element investigation was used to determine the effects of numerous variants of the instrumentation (for example: unilateral vs. bilateral) while the cadaver model allowed for actual testing of the constructs. The proceeding discusses the results of both models with a comparison of the results obtained using the two different models included in the following section.

5.4.1 Comparison of Cadaver and Finite Element Model Stabilization Results

Overall, the agreement found between the measured cadaver motion and the FE model predictions for both unilateral and bilateral instrumentation fixation techniques was reasonable (Table 5.1). All predictions for flexion, extension, lateral bending and axial rotation at C1-C2 fell within 10% of the data obtained from the cadaver model.
Table 5-1: Comparison of FE model predictions with cadaver model results. The data represents the ratio of cadaver motion to FE model prediction motion at C1-C2. Values greater than one correspond to larger cadaver motion, values less than one correspond to larger FE model prediction motion (than was measured in the cadaver model).

<table>
<thead>
<tr>
<th>Rotation</th>
<th>C1-C2 TS</th>
<th>TS+C1LMS+C2PS</th>
<th>TS+C1LMS+C2ILS</th>
<th>TS+WIRE</th>
</tr>
</thead>
<tbody>
<tr>
<td>EXT</td>
<td>1.18</td>
<td>1.07</td>
<td>0.89</td>
<td>1.18</td>
</tr>
<tr>
<td>FLEX</td>
<td>1.08</td>
<td>1.48</td>
<td>1.45</td>
<td>0.94</td>
</tr>
<tr>
<td>LB</td>
<td>1.01</td>
<td>0.83</td>
<td>2.02</td>
<td>3.62</td>
</tr>
<tr>
<td>RB</td>
<td>1.19</td>
<td>0.89</td>
<td>4.29</td>
<td>1.74</td>
</tr>
<tr>
<td>LR</td>
<td>0.99</td>
<td>1.68</td>
<td>1.21</td>
<td>0.94</td>
</tr>
<tr>
<td>RR</td>
<td>1.07</td>
<td>2.22</td>
<td>2.11</td>
<td>0.84</td>
</tr>
</tbody>
</table>

5.4.2 Stabilization of the Upper Cervical Cadaver Model

Overall, the unilateral fixation technique (C1-C2 TS) restricted motion at C1-C2 level effectively in left axial rotation (-66.8%), right axial rotation (-50.1%), left lateral bending (-70.6%), right lateral bending (-82.4%) when compared to intact. However in flexion (-39.2%) and extension (7.59%), TS alone did not result in higher stiffness than in the intact spine which is most likely due to the screw being placed near the center of rotation of C1 and C2 (Table 4.7). Moreover the motion increased in extension when compared to intact. In unilateral fixation, the fixed right atlanto-axial joint acted as a pivot in right axial rotation thus leading to an increase in motion when compared to...
bilateral fixation. This is in acceptance with the previous study done by Kuroki et al.\cite{94} The results from using the unilateral fixation technique in this study were similar to those reported by Kuroki et al.,\cite{94} while using a unilateral transarticular screw except for a slight difference. The present study showed a significant stability in both lateral bending and axial rotation modes with respect to intact when using the C1-C2 TS as opposed to only in axial rotation mode reported by Kuroki et al.\cite{94} The combination of C1-C2 TS with other techniques provided increased stability at C1-C2 in extension/flexion, left/right axial rotation and equally stable in left/right lateral bending when compared to unilateral procedure. The motion reductions at C1-C2 level caused by TS+C1LMS+C2PS and TS+C1LMS+C2ILS and TS+WIRE techniques in extension/flexion and left/right lateral bending were similar when compared to intact. TS+WIRE (LR: -74.7%, RR: -73.6%) on the other hand, failed to restrict the motion at C1-C2 level as effectively as the other two techniques (TS+C1LMS+C2PS (LR: -92.7%, RR: -89.7%)) and TS+C1LMS+C2ILS (LR: -94.7%, RR: -90.7%) in both left and right axial rotations when compared to intact (Table 4.7).

5.4.3 Stabilization of the Upper Cervical Finite Element Model

The FE modeling of upper cervical stabilization hardware is extremely difficult. The application of a relatively stiff hardware system to the mobile craniovertebral junction can, and did result in numerous numerical complications. For instance, placement of C2-C1 transarticular screws across a mobile joint resulted in computational difficulties that were resolved by refinement of the FE mesh.
The results of the FE model support those found by the cadaver model. That is, the FE results demonstrated the ability of all the instrumentation techniques to reduce motion at C1-C2 level. Additionally, the effects of TS+C1LMS+C2PS versus TS+C1LMS+C2ILS versus TS+WIRE and unilateral (C1-C2 TS) versus bilateral instrumentation were evaluated by the FE model. FE model predictions showed that all the three bilateral instrumentation techniques applied to the intact spine (TS+C1LMS+C2PS, TS+C1LMS+C2ILS, TS+WIRE) resulted in motion reductions of greater than 75% for all rotations except extension. The unilateral instrumentation with C1-C2 transarticular screw was only able to provide 13.1% and 47.3% reductions in motion for flexion and extension at C1-C2. Extension was minimally reduced (13.1%) at C1-C2 with unilateral instrumentation (Table 4.11).

The stress analysis of the hardware indicated that the bilateral instrumentation experienced significantly lower maximum von Mises stresses than found in the unilateral case. The largest von Mises stress, 294.2 MPa, was reported for the unilateral instrumentation case with C2 pedicle fixation in right axial rotation. Mechanical properties of titanium alloy (specifically Ti-6Al-4V) indicate that the yield and ultimate strengths of this material are in the range of 795-827 MPa and 860-896 MPa, respectively. Among the bilateral fixation techniques, TS+C1LMS+C2ILS (86.8 MPa) experienced same maximum von Mises stresses when compared to TS+C1LMS+C2PS (91.4 MPa) and TS+WIRE (120.4 MPa) technique showed greater stresses when compared to the other two techniques. The difference in the maximum stresses between the TS+C1LMS+C2PS and TS+C1LMS+C2ILS techniques in all the loading modes was in the area of 3-7 MPa, which is almost insignificant.
The results from the joint contact forces demonstrated that all the three bilateral stabilization techniques decreased the contact forces at the C1-C2 level by 60-95%. Also, the bilateral instrumentation techniques were able to maintain C1-C2 contact due to lateral bending while unilateral instrumentation (right side) was unable to maintain the contact indicated by the loss of contact force at C1-C2 articulation. This finding is in acceptance with previous study done by Puttlitz et.al. [78]

The finite element cases carried out in this study were under “no preload” and no musculature conditions (under pure moment of up to 2.0N m) to mimic the cadaver experiments. Thus, the instrumentation stresses and the contact forces reported in this study might differ under the presence of compressive load and musculature forces.

It is very difficult to place the results of the current study within the framework of the literature database. This is largely due to the fact that biomechanical reports of upper cervical fixation comparing different options for transarticular screw salvaging fixation techniques do not exist. The data provided herein demonstrated that the bilateral fixation provides superior fixation to unilateral fixation. Also, in bilateral fixation scenario, TS+C1LMS+C2PS and TS+C1LMS+C2ILS techniques found to be equally stable and provided equivalent instrumentation stresses at C1-C2 level, whereas TS+WIRE failed to afford the same amount of stability and relatively higher stresses as that of the other two techniques in axial rotation. Therefore this study supports the use of transarticular screw supplemented with C1 lateral mass screw and C2 intralaminar screw in case of serious complications that follow vertebral artery injury by placement of C2 pedicle screws or if the C2 pedicle is not large enough to accommodate the instrumentation.
5.5 Future Work

With regard to the destabilization procedure, there are three typical methods: sectioning of intact ligaments,\cite{43,60} odontoid fracture,\cite{9,96} and odontoidectomy.\cite{97} Crawford et al.\cite{55} dealt with three different types of injuries for testing and indicated that odontoidectomy was predicted to be a much more severe injury than ligament transection and odontoid fracture. In the present study, the atlantoaxial complex was destabilized by odontoid fracture to simulate bone injury. Therefore, there is a need to perform additional studies using the other two destabilization procedures both by testing \textit{in vitro} and simulating the in a finite element model which helps to explore the effect of other pathologies like ligamentous instabilities or tumors on the biomechanics of spine. The anatomic arrangement of the craniovertebral structures is very complex. This complexity makes FE modeling of the region extremely difficult. The model described in this thesis contains only some of these structures, namely the osteoligamentous components. The anatomic variability of the upper cervical spine among the general population could be investigated by varying the dimensions and orientations of certain structures. For example, the role of relative C1-C2 facet orientations could be investigated via a parametric study that changes their sagittal and frontal plane orientations. Also, neural and vascular element contributions to the biomechanical function of the upper cervical spine should not be ignored. The contribution of muscle loading was not explicitly modeled in the current investigations. The effect of a preload on the reported stresses, facet loads and the motion data reported
here needs to be studied. Crawford et al\cite{55} applied multidirectional cyclic loads through 6,000 cycles of 1.5 Nm moments at 2 Hz to the specimens that were fixated using a standard posterior cable and graft system, and reported the increment of C1-C2 motion after fatigue testing. It would be interesting to see if the stabilization techniques employed in this study provide enough mechanical stability to withstand fatigue. Due to time constraint no attempts were made to study the effect of cycling loads (simulating daily living activities) on the results. Overall, a study such as this may provide important information as to the magnitudes of various biomechanical parameters among the three stabilization techniques, such as structural rigidity. This information may further help the \textit{in vivo} promotion of fusion in the atlanto-axial junction.
References


