Creation and Evaluation of a Dynamic, EMG-Driven Cervical Spine Model

THESIS

Presented in Partial Fulfillment of the Requirements for the Degree Master of Science in the Graduate School of The Ohio State University

By

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2013

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Abstract

Biomechanical modeling is one of the ways to assess the risk of pain and injury to an individual. Injuries and pain are becoming more common in the cervical spine. There are many cervical spine models available, but almost all of them were made to study impact loading (whiplash), which makes them unsuitable for studying activities of daily living. A personalized, dynamic, EMG-driven model is very important, because everybody activates their muscles differently, and that different activation will cause different loading patterns. The goal of this study was to create and evaluate a dynamic, EMG-driven cervical spine model that tracks motion and calculates spinal loads in the cardinal planes.

The cervical spine model was created in MSC Adams™ and included vertebrae, intervertebral discs, ligaments, and muscles. The vertebrae (C1-C7) were harvested from a cadaver, and using a white light scanner, were turned into 3D models. The discs were modeled as bushing forces. The ligaments were modeled as single vector forces. Each ligament consisted of three vectors, which enables the model to calculate shear as well as force on the ligament. The muscles were modeled as straight line force vectors between the origin and insertion points. The loads on the discs were calculated using moment arms and force vectors from the ligaments and muscles, as well as velocity and acceleration of the vertebrae and skull.
Three subjects were used to evaluate the model. Subjects completed sagittal flexion and lateral flexion trials. Electromyography data, optical data, and force plate data were collected. The internal load, calculated using the EMG and optical data, was compared to the external load, calculated with the force plate.

The trials in the sagittal plane performed well ($r^2 = .70$, AAE = 1.65 N), while some of the trials in the lateral plane worked well and some did not perform as well ($r^2 = .48$, AAE = 1.74 N). There was significant co-activation in the muscles during the trials. The loads on the discs were reasonable. The sagittal trials had an average compression load of 130.94 N on the discs, an average lateral shear load of 70.52 N on the discs, and an average anterior-posterior shear load of 52.72 N on the discs. The lateral trials had an average compression load of 234.21 N on the discs, an average lateral shear load of 136.78 N on the discs, and an average anterior-posterior shear load of 84.61 N on the discs.

This model met its goals reasonably well, it was able to track motion, predict moments on the cervical spine, and calculate loads on the discs. The co-activation in the muscles was expected, as all of the muscles were needed to maintain balance of the head and complete the motion. The loads on the discs were plausible; they were well below established failure tolerances, as well as below the loads on the discs in the lumbar spine during similar trials. With improvements, this model will be able to accurately predict the loads on the discs during many different tasks.
To my wife, Elise,
And my parents, Rob and Leslie
Acknowledgments

I would like to thank my advisor, Dr. William Marras, for his help and patience during my thesis work. He has taught me to think critically about everything, and I am better off for being a part of his lab.

I would also like to thank the students and staff in the Biodynamics Laboratory for their help, especially to Greg Knapik and Jon Dufour for their help with the modeling. I could not have gotten this far without their help.
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Chapter 1: Introduction

Neck pain is increasing in prevalence as more and more jobs in the United States involve working with computers in an office setting. A study in Great Britain found that 19.5% of people had reported some form of neck pain in the last week alone (Palmer, et al. 1995). In the last 6 months, 47.2% of people have reported neck pain at least once (Cote, Cassidy and Carroll 1998), while 75.1% of people reported neck pain in the last year (Rajala, et al. 2000). Up to 50% of all soldiers leaving their units now leave for neck pain, (S Cohen 2012) and only 14% of the soldiers ever return to their unit (S. Cohen, et al. 2010). One of the strong predictive factors for neck pain is also having low back pain (S. Cohen, et al. 2010) (Croft, et al. 2001). More severe neck pain is also becoming a problem. Thirteen point eight percent of people reported an incident of neck pain lasting 6 months or more (Bovim, Schrader and Sand 1994). Neck pain has a very high remission rate, with between 50-85% of people who experience an episode of neck pain reporting neck pain of the same severity or greater 1-5 years later (Carroll, et al. 2008). Forty eight percent of patients who present with neck pain in the last month have pain of at least the same frequency one year later (Hill, et al. 2004). This shows that not only does neck pain occur frequently; it is difficult to eliminate once it does occur.

While we know that neck pain is prevalent, it is very difficult to predict the risk of injury. The loading on the disc is often the trigger for pain, and biomechanical models allow us to predict what the loading will be. A personalized model is very important,
because every subject is different, and activates their muscles differently to produce a given motion. The muscle activation of each individual person is very important, as the muscle force is a major contributor to tissue loading. Biomechanical models provide us with a way to test a person on a personalized level. This model will incorporate subject specific motion and EMG activity to calculate the load on their cervical spine.

Currently, there are several inverse dynamic models available for the analysis of the motion and loading of the cervical spine. These models are mostly used for studying how the cervical spine responds to whiplash and impacts (car crashes, slips/falls, etc.) Several inverse dynamic models are reported in the literature that include muscles with passive force only (M. de Jager 1994) (R. Jost 2000) (KH. Yang 1998). In addition, there have also been models that have incorporated the active force component of the muscles (K Brolin 2005)(M de Jager 1998) (Horst 2002) (CA Van Ee 2000). While inverse dynamic models work well for impact testing, where muscles do not respond quickly enough to alter tissue loading, they do not work as well for quantifying activities of daily living, when muscle activity is the major loading mechanism. The inverse dynamic models optimize the muscle force, which will eliminate co-activation. It has been shown that muscle force estimates were as much as 218% higher when using an EMG-driven model than when using an inverse dynamic model (J Cholewicki 1995). Attempts to create an EMG-driven cervical spine model (K Netto 2008), found that more work was needed in order for the model to be valid. There is a significant amount of co-activation of the low back musculature and this co-activation can significantly increase the spinal loading (KP Granata 1995). Similarly, it is expected that there will be as much or more
co-activation in the cervical spine that can greatly impact tissue loading. Such a model will allow researchers to study the motion and loading of the cervical spine on a subject specific level.

Therefore, the specific aims of this research are to create a model that:

- Accurately tracks motion of subject
- Accurately predicts moments in the cardinal planes.
- Assess the validity of the predicted moments with external measurements
- Investigate amount of co-activation during trials
- Predict 3D tissue loads
Chapter 2: Model Development

The model was built in the MSC ADAMSTM (Automatic Dynamic Analysis of Mechanical Systems) software package. ADAMSTM is a powerful modeling package that is used broadly in the automotive industry for testing and analyzing new designs. It can simulate the reaction of modeled bodies to motions, forces, and other test data while accounting for momentum and gravitational effects. Rapidform™ was used to create the models of the vertebrae. Rapidform™ is a 3D CAD software package. It was used extensively to turn the scan data of the vertebrae into CAD models that could be imported into ADAMSTM and manipulated.

Model Overview

The model incorporated kinematic, kinetic, EMG, and anthropometric measures to evaluate a subject's loading pattern. Fig. 1 shows the flow of information through the model.
MSC Adams™ tracks modeled bodies based on their position in a Cartesian coordinate system, and their orientation with Euler angles. The position and orientation of the bodies was then used to calculate the longitudinal and angular velocities of the bodies. The inertia was then calculated using the position, orientation, velocity, acceleration, and mass of the bodies. Using the kinetic and kinematic data, Adams then generated a set of differential equations, one equation for each force, moment, or distance measured by the model, that provide the information necessary to solve the dynamic
These equations were then solved with an iterative Newton-Raphson algorithm, which attempts to minimize the potential energy of the system. Once the potential energy was minimized, Adams calculated each of the differential equations, 5 of which give tissue loads.

### Ligaments

The ligaments were modeled as single force vectors located between 2 points. Four ligaments were included in the model; the anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL), the ligamentum flavum (LF), the interspinous ligament (ISL) (Fig. 2).

![Figure 2: Close up of ligaments in the lateral plane. Locations of different ligaments highlighted.](image)

The properties of the SSL (supraspinous ligament) were included in the ISL as per the dissection of Panjabi et al. (Panjabi 1991). It was missing in half of the subjects and
falling apart in the other half of subjects. The ligament properties used in the model are displayed below (Table 1).

Table 1: Ligament physiological cross-sectional areas (F Pintar 1992)

<table>
<thead>
<tr>
<th>Ligament</th>
<th>CSA (m$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior Longitudinal (ALL)</td>
<td>3.24E-05</td>
</tr>
<tr>
<td>Posterior Longitudinal (PLL)</td>
<td>5.20E-06</td>
</tr>
<tr>
<td>Ligamentum Flavum (LF)</td>
<td>8.42E-05</td>
</tr>
<tr>
<td>Interspinous Ligament (ISL)</td>
<td>3.51E-05</td>
</tr>
</tbody>
</table>

The force on the ligaments was calculated using (Eqn. 1).

$$ F = \dot{\epsilon} \times \dot{A} \times \text{Modulus} $$

(1)

where:

$\dot{\epsilon}$ = current ligament strain

$\dot{A}$ = physiological cross-sectional area

Modulus = modulus of elasticity (M Panjabi 1982)

Each ligament was modeled as 3 component force vectors (Fig. 3), and the resultant force vector was calculated. This was done to ensure that the force could be obtained for the physiological width of the ligament, as well as shear loading on the ligaments.
Figure 3: Close up of the ligaments, showing 3 force vectors for ALL.

Vertebrae

The vertebrae used were harvested from a subject who was 28 years old (178 cm, 87.3 kg) with no history of neck or back pain, and no degenerative bone disease. The vertebrae were carefully cleaned by trained anatomists. They were then scanned with a Faro™ arm device (Faro Arm Technologies, Lake Mary FL). The C7 vertebra was locked to the torso of the skeleton within the model, which assumes that there is no motion between T1-C7. The facet joints were modeling using contact forces in MSC Adams.

Muscles

Each muscle was represented as a single force vector located between two points (defined by the origin and insertion), located on the plane at the level of C7 and either C1 or the skull respectively. The muscles were chosen based on their moment arm.
mechanical advantage, and their accessibility via surface electromyography (EMG). The muscles in the neck are deeply layered, which makes EMG collection difficult. However, the three muscle pairs chosen were the sternocleidomastoid (SCM), the trapezius (TRAP), and the semispinalis capitis (SCAP) (Figs. 4, 5, and 6).

Figure 4: Lateral view of the muscles modeled.
Figure 5: Frontal view of the muscles modeled (red=scm, blue=scap, green=trap).

Figure 6: Posterior view of the muscles modeled (red=scm, blue=scap, green=trap).
Table 2 shows the cross sectional areas as well as the origin and insertion points of the muscles used in the creation of the model.

Table 2: Insertion and origin locations and physiological cross section of the muscles.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Origin</th>
<th>Insertion</th>
<th>CSA (cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sternocleidomastoid</td>
<td>Medial portion of clavicle</td>
<td>Mastoid process</td>
<td>2.901</td>
</tr>
<tr>
<td>Semispinalis Capitis</td>
<td>Transverse processes of T6-C7</td>
<td>Superior nuchal line</td>
<td>4.267</td>
</tr>
<tr>
<td>Trapezius</td>
<td>Spinous process of C2</td>
<td>Spine of scapula</td>
<td>4.95</td>
</tr>
</tbody>
</table>

The model calculated spinal loads based on body segment kinematics, externally applied loads, and muscle-generated moments (M). Three-dimensional muscle-generated moments predictions were made by dynamically summing muscle force vectors (F) and moment arm vectors (r) for each of the 6 model muscles (j) (Eqn. 2).

Dynamic muscle-generated moments (Eqn. 2)

\[ M = \sum_{j=1}^{6} r_j \times F_j \]  

(2)
The muscle force was calculated using (Eqn. 3).

\[
Force = (EMG \times GR \times \dot{\alpha} \times f_{\text{len}(act)}) + (\dot{\alpha} \times \dot{\varepsilon} \times f_{\text{len}(pass)})
\]

(3)

where:

- \(EMG\) = muscle activity
- \(GR\) = gain-max ratio (JS Dufour n.d.)
- \(\dot{\alpha}\) = physiological cross-sectional area
- \(\dot{\varepsilon}\) = muscle strain

\(f_{\text{len}(act)}\) = active portion of force-length equation (A. Hill 1938)

\(f_{\text{len}(pass)}\) = passive portion of force-length equation (A. Hill 1938)
Intervertebral Discs

The intervertebral discs have stiffness properties that were adapted from a lumbar spine model (GG Knapik 2009). The stiffness properties used in the current model are lower than those reported in (GG Knapik 2009). This is because the stiffness properties in (GG Knapik 2009) were taken from a study (MG Gardner-Morse 2004) which calculated disc stiffness properties with the ligaments included with the disc in the motion segment. The current model calculates ligament force separately from the disc. The disc stiffness values used in this study are displayed in Table 3.

Table 3: Translational and rotational stiffness values of cervical discs.

<table>
<thead>
<tr>
<th></th>
<th>X</th>
<th>Y</th>
<th>Z</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Translational Stiffness (MN/m)</strong></td>
<td>0.7303</td>
<td>0.5289</td>
<td>0.5604</td>
</tr>
<tr>
<td><strong>Rotational Stiffness (Nm/deg)</strong></td>
<td>0.8635</td>
<td>0.7844</td>
<td>0.6262</td>
</tr>
</tbody>
</table>
Chapter 3: Evaluation

Approach

The goal of this research was to create a dynamic, EMG-driven biomechanical model of the cervical spine to be evaluated as it moves in cardinal planes. The model was evaluated by comparing predicted dynamic moments to measured dynamic moments in each plane.

Subjects

Three subjects were tested to initially evaluate the model. The subjects were recruited from the university population. Age, stature, and mass are displayed below (Table 4). All subjects were healthy and had no previous history of neck pain that limited daily activities for 7 days in a row. Approval from the Institutional Review Board was obtained and each subject gave informed consent prior to participation.

Table 4: Mean, standard deviation, and range of age, stature, and mass of subjects

<table>
<thead>
<tr>
<th></th>
<th>Mean</th>
<th>Standard Deviation</th>
<th>Minimum</th>
<th>Maximum</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Age</strong></td>
<td>26.3</td>
<td>1.247219</td>
<td>25</td>
<td>28</td>
</tr>
<tr>
<td><strong>Stature (cm)</strong></td>
<td>1.79E+02</td>
<td>4.640697</td>
<td>172.72</td>
<td>183.3</td>
</tr>
<tr>
<td><strong>Mass (kg)</strong></td>
<td>8.15E+01</td>
<td>13.14682</td>
<td>65.32</td>
<td>97.52</td>
</tr>
</tbody>
</table>
Experimental Design

The subjects completed 12 trials during this experiment. The subject completed 6 sagittal flexion trials and 6 lateral flexion trials. The trials were performed at 2 different speeds, natural and fast. The lateral trials were all bends to the right. The order of the trials was randomized prior to the collection. The starting position was a seated position, with the head facing forward. The data collected was used to evaluate the model.

Procedure

Initially, the experimental procedure was explained to the subject, a consent form was signed, and background information and anthropometry were collected. The following anthropometric measures were recorded: weight, stature, spine length, shoulder width, xyphoid trunk depth and breadth, head length, head height, head breadth, skull circumference, neck circumference, trunk circumference, and subscapular skinfold. Age, gender, and ethnicity were also recorded.

An OptiTrack™ optical motion capture system (NaturalPoint, Corvallis, Oregon, USA) was used to capture the motion of the subject. The system was calibrated to an accuracy of 0.4 mm. The system used 24 cameras to triangulate the positions of reflective markers (B&L Engineering, Santa Ana, California, USA) that were placed on the body. The markers were 6.4 mm in diameter and attached to the subject with double sided electrode tape.

34 markers were used to create a skeleton (Figs. 7 and 8), using the pre-set locations given by the Arena Software™ (NaturalPoint, Corvallis, Oregon, USA) which
was used to collect the data. The skeleton allowed the camera system to track the location of all of the body segments based on the location of the individual markers. The Arena Software outputs the location of the body segments as well as the angle between them. The segments of interest in this study were the shoulders, neck, and head.

Figure 7: Subject in T-Pose before beginning trial

Figure 8: Skeleton created in OptiTrack
A Grass™ EMG system (Grass Instruments, Warwick, Rhode Island, USA) was used to collect the EMG data. 3 bilateral muscle pairs were used. The 3 muscle pairs that were used in the creation of the model were the semispinalis capitis, the sternocleidomastoid, and the trapezius.

The EMG sensors on the semispinalis capitis were placed on the back of the neck approximately 2 cm below the occipital bone and 2 cm lateral to midline (E Keshner 1989). (Fig. 9, pair 2)

Figure 9: EMG sensor placement, semispinalis capitis.
The EMG sensors on the sternocleidomastoid were placed over the muscle belly approximately 1/3 of the way down from the mastoid process (E Keshner 1989). (Fig 10, pair 1)

The EMG sensors placed on the trapezius were located at the base of the neck, at the C6-C7 level (E Keshner 1989). (Fig 10, pair 3)

After the EMG sensors and optical markers were in place the subject was allowed to move around to get used to the sensors and ensure that their motion was not restricted.

To begin each trial, the subject would stand in a T-pose (Fig. 7). The T-pose is the neutral position for the OptiTrack system, and this allows all of the markers to be located at the beginning of each trial. After the T-pose the subjects sat directly on a Bertec™ force plate (Bertec Corporation, Columbus, Ohio, USA). The force plate had a rigid structure attached to it with a low profile back to ensure that the optical markers
could be located. The subject was secured to the structure so that the only motion picked up by the force plate would be from the motion of the head (Fig 11).

Figure 11: Experimental set-up. Subject is sitting on the force plate, secured to the rigid structure.

Once the subject was secured to the force plate, it was hardware and software zeroed, to ensure that any change in the force plate data reflected a movement by the subject. The subject would then perform the required motion. Optical data was collected using Arena™ software. EMG data was collected using proprietary software developed in the Biodynamics Laboratory at The Ohio State University.
Data Analysis

Data were exported from Arena and the OSU custom data acquisition software to MATLAB where the analysis was performed. The EMG data were first high-pass filtered at 30 Hz, low-pass filtered at 450 Hz, and notch filtered at 60 Hz and corresponding aliased frequencies with a sideband of .25 Hz. The optical data was filtered with a moving average with a window of 10 ms. The data were then exported to MSC Adams where it was input into the model. The model was then run and the data were analyzed. The data recorded from the force plate were used to determine the measured moment (MM), which was defined as the moment that was created when the head moved during the trials. The EMG and motion data were used to calculate the predicted moment (PM), which was the moment that the model predicts the motion of the head to produce. To evaluate the model, the PM was compared to the MM over time to determine how well the model could accurately predict the moment that the motion would produce.

Results

The objective of this evaluation was to do a preliminary analysis of the model concept. A descriptive analysis was done instead of a formal statistical validation.

The two types of trials that were collected and analyzed were sagittal flexion and lateral flexion. The model tracked the motion of the subject very well; there was very little delay (< 5 ms) between when the subject started the motion and the model output began. Overall, the model was able to generally track the subject’s motion in all 3
planes, and predict the moment in the sagittal and lateral planes. $R^2$ and average absolute error (AAE) were calculated and reported. $R^2$ is describes how well the PM matches the MM. AAE is defined as the average difference (error) between the MM and the PM.

Sagittal Flexion

Table 5 shows the overall $r^2$ and (AAE) for the sagittal flexion trials.

Table 5: Comparison of the natural and fast trials in the sagittal direction.

<table>
<thead>
<tr>
<th></th>
<th>r2</th>
<th>Standard Deviation</th>
<th>AAE (Nm)</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Natural</td>
<td>0.717</td>
<td>0.109</td>
<td>1.565</td>
<td>0.664</td>
</tr>
<tr>
<td>Fast</td>
<td>0.672</td>
<td>0.221</td>
<td>1.686</td>
<td>0.780</td>
</tr>
</tbody>
</table>

Of the 18 sagittal trials run, 1 was discarded as the data from Arena was corrupted, and the motion could not be modeled.
Fig. 12 shows a sagittal flexion trial that worked very well.

Figure 12: Good sagittal flexion trial (r² = 0.902 AAE = 0.866 Nm).

The trials that worked well had the PM match the MM accurately from start to finish, with no large offsets. This type of trial constituted 2 of the 17 flexion trials.
Fig. 13 shows an average sagittal flexion trial.

![Average Sagittal Flexion Trial](image)

Figure 13: Average sagittal flexion trial ($r^2 = 0.764$ AAE = 1.06 Nm).

The average sagittal flexion trials matched the peak moment well, but had areas where the PM was different from the MM (between .5 and 1 sec. in above plot). This type of trial made up a significant portion of the flexion trials, 13 of the 17 were similar to Fig. 3.2.
Fig. 14 shows a sagittal flexion trial that did not work well.

![Graph showing sagittal flexion trial](image)

Figure 14: Bad sagittal flexion trial ($r^2 = 0.31$, AAE = 1.58 Nm).

The sagittal flexion trials that did not work showed a PM that did not match the peak of the MM, and a large time shift. This type of trial constituted 2 of the 17 flexion trials.
Lateral Flexion

Table 6 shows the overall r2 and average absolute error (AAE) for the lateral flexion trials.

Table 6: Comparison of natural and fast trials in the lateral direction.

<table>
<thead>
<tr>
<th></th>
<th>r2</th>
<th>Standard Deviation</th>
<th>AAE (Nm)</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Natural</td>
<td>0.497</td>
<td>0.143</td>
<td>1.462</td>
<td>0.430</td>
</tr>
<tr>
<td>Fast</td>
<td>0.468</td>
<td>0.233</td>
<td>2.014</td>
<td>1.346</td>
</tr>
</tbody>
</table>

Of the 18 lateral trials run, 2 were discarded due to problems with the data coming in from Arena. This did not allow the model to track the correctly track the motion.
Fig. 15 shows a lateral flexion trial that worked well.

Figure 15: Good lateral flexion trial ($r^2 = 0.78$, $AAE = 0.67$ Nm).

The lateral flexion trials that worked well had a PM that matched the timing of the MM very well. The peak values were very close, and it missed only small peaks. This type of trial constituted 3 of the 16 lateral flexion trials.
Fig. 16 shows an average lateral flexion trial.

![Graph showing average lateral flexion trial](image)

Figure 16: Average lateral flexion trial (r^2 = .45 AAE = 1.7 Nm).

The average lateral flexion trials had the PM match the overall shape of the MM fairly well, but was farther off on the peak moment than the good trials. This type of trials constituted 10 of the 16 lateral bend trials.
Fig. 17 shows a bad lateral flexion trial.

Figure 17: Bad lateral flexion moment matching curve (r2 = 0.1 AAE = 5.1 Nm)

The bad lateral flexion trials had a large time shift, which caused problems. There were 4 bad lateral flexion trials out of the 16 run.
EMG Activity

Co-activation was seen in every trial, in both sagittal and lateral trials. Fig. 18 shows the filtered EMG data from a sagittal flexion trial. The semispinalis capitis and sternocleidomastoid muscles were both firing at a high rate during the motion, and the trapezius muscle was firing to stabilize the head and keep it upright.

Figure 18: Filtered EMG data for a flexion trial
Fig. 19 shows the filtered EMG data from a lateral trial. This trial was a lateral flexion to the right. The EMG activity for the muscles on the right side of the body was higher than that of the left side of the body, but the muscles on the left side of the body were firing the entire time to keep the head upright and on track.

Figure 19: Filtered EMG data for a lateral flexion trial.
Disc Loads

The load on the disc at each level was calculated for the sagittal flexion and lateral flexion trials.

Table 7: Compression, lateral, and AP shear loads on the discs during sagittal flexion trials.

<table>
<thead>
<tr>
<th>Sagittal</th>
<th>Comp (N) (St. Dev.)</th>
<th>Lat (N) (St. Dev.)</th>
<th>AP (N) (St. Dev.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C6-C7</td>
<td>123.32 (78.425)</td>
<td>68.94 (24.63)</td>
<td>45.01 (13.34)</td>
</tr>
<tr>
<td>C5-C6</td>
<td>127.35 (80.14)</td>
<td>68.38 (24.95)</td>
<td>40.79 (10.46)</td>
</tr>
<tr>
<td>C4-C5</td>
<td>148.5 (74.01)</td>
<td>73.07 (22.77)</td>
<td>51.09 (17.42)</td>
</tr>
<tr>
<td>C3-C4</td>
<td>137.19 (65.61)</td>
<td>73.34 (23.87)</td>
<td>50.45 (20.5)</td>
</tr>
<tr>
<td>C2-C3</td>
<td>118.34 (70.41)</td>
<td>68.89 (23.71)</td>
<td>76.25 (23.48)</td>
</tr>
</tbody>
</table>

Table 8: Compression, lateral, and AP shear loads on the discs during lateral flexion trials.

<table>
<thead>
<tr>
<th>Lateral</th>
<th>Comp (N) (St. Dev.)</th>
<th>Lat (N) (St. Dev.)</th>
<th>AP (N) (St. Dev.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>C6-C7</td>
<td>229.62 (158.8)</td>
<td>80.06 (25.95)</td>
<td>115.88 (52.76)</td>
</tr>
<tr>
<td>C5-C6</td>
<td>221.21 (160.19)</td>
<td>88.7 (47.53)</td>
<td>98.74 (49.41)</td>
</tr>
<tr>
<td>C4-C5</td>
<td>277.53 (122.64)</td>
<td>103.42 (93.9)</td>
<td>151.52 (48.9)</td>
</tr>
<tr>
<td>C3-C4</td>
<td>245.76 (120.79)</td>
<td>67.58 (17.69)</td>
<td>198.04 (55.16)</td>
</tr>
<tr>
<td>C2-C3</td>
<td>196.96 (121.51)</td>
<td>83.28 (43.41)</td>
<td>165.42 (85.56)</td>
</tr>
</tbody>
</table>
Figure 20: Plot of compression force for a sagittal flexion trial
Chapter 4: Discussion

The model was designed to predict the loading of a subject’s cervical spine in the cardinal planes. While it accomplished the overall goals, there were still some areas in which improvements could be made.

The model was able to reasonably predict the moment measured on the force plate when the motion stayed in the cardinal planes. The trials in the sagittal direction worked especially well. This was because the subjects were able to keep the motion in the sagittal plane with little motion in the lateral plane. When the motion stays in the sagittal plane the facet joints disengage and reengage at the same time. The lateral direction worked well but not as well as the sagittal plane. One of the reasons that the lateral direction did not work as well is the facet contact forces. MSC Adams handles contact forces as a single force between two bodies, and each vertebral body has two facet joints. The model had some trouble when one of the facets disengages while the other facet has an increased load on it. All of the trials in the lateral direction had sections in the PM that did not match the MM (Figs 15, 16, 17) that coincided with the loading of the right facet and the unloading of the left facet, and then with the reloading of the left facet during the return motion.

The model performed as well at fast speeds as well as normal speeds. All 3 muscles were activated during the motion. The EMG activity was higher during the fast trials, which was expected. The trapezius was activated for the whole trial, and there was
an increase in activity whenever the motion occurred. The semispinalis capitis and sternocleidomastoid both activated when the motion starts (Fig. 18), indicating that co-activation is necessary to balance the head. The co-activity will increase the spinal loads, which will differentiate this model from the many inverse dynamic models that optimize muscle activity. This will allow the model to be individualized to each subject, as everyone has a different pattern of muscle activation.

The loading on the discs was reasonable. Most of the cervical model papers present different results, including angular velocity and acceleration, head angle, head restraint impact forces, and vertebral rotations. However, there are a few models that discuss spinal loads. (V. Goel 2006) wrote a paper on spinal implant testing. When implants are tested in the cervical spine, they recommended a preload between 50-100 N to account for the mass of the head. At the beginning of each trial, before the muscles begin to activate, the load on the discs was approximately 50 N (Fig. 20), which is exactly what they suggest. There was also a study testing cadaver discs to failure (N. Yoganandan 2001). The loads on the discs reported in this paper were far lower than the failure limits across all levels of the spine. The loading also showed a reasonable pattern, as the highest compression values were at the C4-C5 level, which in this model was the vertebral level that was most vertical, where the majority of the force will be in compression. The values were all lower than those presented in (JS Dufour n.d.) for the lumbar spine. The compression was expected to be lower in the cervical spine as there is less weight that needs to be supported (head vs. upper torso and head, respectively). The trials were comparable in the sagittal plane, as the subjects completed flexion trials with
no weight. The study did not have the subjects complete a lateral trial with no weight, so no comparison can be made.

The most important part of each trial is the peak moment. This was where the highest loads on the disc occur, which makes this the most dangerous part of the motion. Even though the MM and PM don’t always match up perfectly as a whole, the peak of the PM was generally the same magnitude as that of the MM. Every trial had a small amount of offset due to lag in the motion tracking system, as well as the electromechanical delay, which is the delay between when a muscle fires and when the EMG sensor receives the signal. The fact that the peak MM and PM match means that the model was able to predict the load on the neck accurately.

There were also parts of the model that didn’t go as planned. One of the things that didn’t go well was predicting the moment when the motion was not in the cardinal planes. This was particularly prevalent in the lateral trials. All of the subjects had at least one trial where there was as much motion in the sagittal plane as in the lateral plane. These trials had a maximum $r^2$ of 0.3. The facet joints were part of this problem. The lateral trials only took into account the facet load in the lateral direction, which ignores the contribution in the sagittal direction of the facet joints. This assumes that the lateral portion of the facet load is the entirety of the facet load.

For the trials that did not work as well (2 sagittal and 3 lateral), we went back and looked at the kinematic data to ensure that the subjects complied with the instructions. We found that for these trials, the subjects displayed motion in the other plane or had motion from body segments that were meant to be stationary. In the sagittal
trial above (Fig. 14) the shoulders began to move at the same time as the MM began to increase, while the head began to move at the same time the PM began to increase. The shoulder motion would cause an increase in MM without any muscle activation, which is needed to calculate the PM. The lateral trials showed that the shoulders move in the lateral direction before the head moves, which caused the offset. The lateral trials also showed two small peaks before and after the main peak. Those were also from the shoulder motion. The harness was designed to ensure that the subject did not move forward, but the shoulders had some room to move laterally. In the future, a strap holding the shoulders down would be helpful.

There were several areas of the model that can be improved in order to make it effective in more cases.

The first area needing improvement was the facet contacts. They were modeled as a contact force in MSC ADAMS, and only one contact force can be applied per solid body. There are two facet joints per vertebral body, with one force measured between them. Fixing the way that facet contact is measured would greatly improve the way the PM is calculated. Also, including the load on the facets in all directions would improve the way the model deals with off-plane motion.

The second area needing improvement was the way that the muscles are modeled. Currently, the muscles are modeled as straight line forces. When the neck moves, the muscles keep their straight line of action; they should bend with the neck. The muscles could be broken into segments, where each vertebral level has a muscle segment. This
will allow the muscle line of action to better follow the motion, which will improve the accuracy of the PM.

The third area needing improvement was the way that the external moment is measured. The external moment is currently measured on a force plate, with the subject sitting on the force plate attached to a rigid structure on the force plate (Fig 11). This was done so that it could be ensured that the only motion detected was from the motion of the head and neck. While this worked very well for the model validation, it makes it difficult for the model to be used in real life applications, as the subject must be seated on and attached to the force plate. A better way to measure the external moment would allow the subject to move around more freely.

The fourth area needing improvement was separating C1 and C2 so that the twisting motion can be realistic. According to (Dugailly 2011) 40% of the twisting motion occurs between C1 and C2, with the rest of the twist being distributed to the rest of the cervical spine. Because the goals of this study were to create and validate a model that worked in the sagittal and lateral planes, C1 and C2 were fused together. The twisting motion that occurred as a result of the sagittal and lateral motion was distributed evenly through the rest of the cervical spine. By evenly distributing the twist about the rest of the cervical spine, each motion segment (2 vertebrae and 1 disc) took approximately 20% of the motion. With C1 and C2 separated, this would go down to approximately 12% per motion segment, with 40% occurring between C1 and C2. Separating C1 and C2 would make the twist motion more realistic.
The fifth area needing improvement was scaling the size of the vertebrae to the subject. The model currently used the vertebrae from a single cadaveric specimen, and the height of the spine, as well as the size of the individual vertebrae, was the same for all 3 subjects even though the subjects were different heights (mean = 179 cm, st dev. = 4.6 cm). In the future, it would be beneficial to scale the vertebrae to the anthropometry of each individual subject. This would allow the motion to be as realistic as possible. Another future improvement would be using CT and MRI scans from each subject to create an exact model of their individual vertebrae and muscle placement. This would allow for very personalized modeling.

The sixth area needing improvement was adjusting the muscle force equation. Currently, the muscle force equation accounts for the amount of force that a muscle can produce at different lengths ($f_{ten}$), but it a general equation (A. Hill 1938). This equation was created using individual muscle fibers from frog skeletal muscle. It could be optimized for the muscles in the human neck. Also, muscles produce different amounts of force depending on how fast they contract (force-velocity), and this is not taken into account in this model. The optimization of the force-length modulation and the addition of force-velocity modulation would increase the accuracy of this model.

There were several limitations with this study that need to be addressed.

The subjects were secured to the rigid structure, which restricted the motion of their upper torso. Being unable to move their upper bodies could potentially have changed the motion pattern of the subject. This was done because small motions by the upper torso could affect the MM, which would change how well the PM matches it.
While the model performed well, only 3 of the 34 muscles in the neck (J. Borst 2011) were modeled. There is some motion that was not picked up by these 3 muscles, especially from the deep muscles that are hard to pick up with surface EMG. While there were only 3 muscles modeled, they represent major power producing muscles in each of the cardinal planes.

The vertebral bodies were taken from a study done on a single cadaver. While the specimen had no history of neck or back problems, it is still a single subject, and people have wide ranges of vertebral body size and shape. Different vertebral body shapes will have different shaped facet joints, which would change the loading due to the facet joints, which would change the overall PM.

The subjects were all young males (26.2 yrs.). Their muscle activity and motion patterns could be different from people of different ages.

With a few improvements, this model could be used in many different circumstances. It could be used for:

- Soldiers: Measure the load on their neck with their helmets on, and help determine the best placement of equipment to decrease the amount of neck pain.

- Surgeons: Work on postural interventions

- Workplace interventions: Monitor height/location

- Many others

Overall this model was a reasonable approximation of the cervical spine and with future improvements can be made even more useful in many different applications.
Chapter 5: Conclusion

In this study, a dynamic, EMG-driven, biomechanical model was created to evaluate the model concept. The model consisted of vertebrae, discs, ligaments, and muscles. The load on the discs was calculated using a combination of muscle force, ligament force, and velocity and acceleration of the head/neck. Three subjects were run to evaluate the credibility of the model. The model performed reasonably well, from tracking the motion of the subject, to calculating the load on the discs, and to matching the measured external moment to the predicted internal moment.

Improvements are needed in several areas.

1. The facet joints need improvement, as they were modeled as a single contact force, which was applied over 2 separate contact areas. Improvement in this area would increase the accuracy of the model.

2. The muscle forces also need to be improved. They are currently straight-line muscle approximations, and curved muscles that follow the path of the neck would greatly improve the ability of the model to match the measured external moment.

3. The way that the measured external moment was calculated needs improvement as well. The subject was secured to the force plate, which restricted their motion. Improvement in this area would allow the model to be used in a wide range of tasks.
4. The way that twisting motion is modeled needs to be improved. C1 and C2 were locked together, so the twisting motion was distributed throughout the rest of the cervical spine. In order to correctly model this, C1 and C2 need to be separated, as in real life that is where 40% of the twisting motion occurs.

5. The size and shape of the vertebrae need to be improved. They were harvested from a single cadaver, and the size was the same for all subjects.

6. The way that the muscle force was calculated also needs to be improved. The current equation accounts for how the change in length affects the ability to produce force, but does not account for how changes in velocity affect the amount of force that can be produced.

With improvements, this model could be used to assess the risk in a wide range of tasks, and help improve the safety of many jobs.
Bibliography


