Biomechanical interpretation of clinical data for cervical injury rehabilitation assessment

D.F. Sim\textsuperscript{1,2}, T.M. Barker\textsuperscript{1,2}, M.J. Pearcy\textsuperscript{1,2}, J.H. Evans\textsuperscript{2}

\textsuperscript{1}Mechanical, Manufacturing & Medical Engineering, Queensland University of Technology, Australia
\textsuperscript{2}Centre for Rehabilitation Science and Engineering, Queensland University of Technology, Australia

Introduction
The diagnosis and rehabilitation of cervical spine dysfunction has posed significant challenges which has attracted a multi-disciplinary research effort. In a recent review of minor cervical injuries, Bogduk & Yoganandan (2001) emphasised again that current imaging technology often fails to highlight any abnormality even though substantial dysfunction is exhibited. Following the collection of objective clinical data in a non-invasive manner, rational interpretation of the data is vital for effective rehabilitation delivery. Given the disproportionately high share of resources allocated to whiplash associated disorders, as a consequence of imprecise diagnosis, there is a clear call for the application of biomechanical modelling and analysis to cervical injury diagnosis and rehabilitation management. The modelling and analysis discussed in this paper is drawn from current research aimed at correlating clinically measurable range of motion and muscle control with particular dysfunction of the muscles, joints and ligaments in the neck.

Methods
The detailed biomechanical model of the cervical region developed by M. de Jager and refined by M. van der Horst et al using MADYMO (Mathematical Dynamic Modelling) software provided the basis for the current research. MADYMO provides a combined rigid body and finite element platform and was primarily developed by TNO (Netherlands) for crash test simulation. The response of this musculoskeletal model had been validated for impact, but adaptation of the model was required to simulate responses to clinical procedures and range of motion testing.

A surface was defined which surrounded the vertebrae and musculature to allow the simulation of manual manipulation and the deep neck flexor test that is used for muscle control evaluation. The surface – external force contact interactions are defined by either linear or non-linear force-displacement functions. In the case of the deep neck flexor test and deep muscle rehabilitation exercises, this external force is applied by a pressure biofeedback unit (airbag) that is placed under the posterior surface of the neck.

Passive range of motion testing was simulated by applying a torque in the direction of the desired motion at the body coordinate origin of the head. The body rotation output from the model was compared with data collected using a 3D Polhemus Fastrak electromagnetic goniometer at the Whiplash Diagnostic Clinic, University of Queensland.

Muscle spasm is often manifested clinically leading to restricted range of motion and aberrant motion patterns. Spasm was modelled by inverting the force-length function specified for the Hill muscle model and setting the reference length (at which the maximum force is generated) equal to the muscle resting length. Figure 1 shows the difference between the active and spasm force-length functions. The width of the curve models the severity of the spasm (resistance to stretch).

![Force length curves](image)

**Figure 1.** Normalised force-length functions showing the difference between active and spasm characteristics.
The dysfunction caused by capsular ligament laxity (facet joint hypermobility) or reactive joints that are protected by deep muscle are modelled by altering the joint and ligament stiffness. Ligaments are represented by tension only Kelvin force elements and joints have a lumped resistance to translation and rotation about the joint coordinate axes that are located at the approximate centres of rotation.

**Results**

The model is constructed as a chain of nine rigid bodies representing the spine from T1 to the head, and connected by lumped resistances (representing the intervertebral disc), 122 muscle segments, 62 ligaments and 19 contact points, all of which can be used for force and kinematic output. While in-vivo validation of the absolute results of internal forces and motions is near impossible, it is assumed that if the kinematic response of the bodies is realistic, the forces that effect and constrain the motion should be within range.

Passive range of motion in the primary planes of flexion/extension, lateral flexion and rotation initially revealed instability in the upper cervical spine. The facet joints of the upper cervical spine have a major influence on segmental motion at that level and the constraint of a central ligament, as defined for the lower cervical spine, was insufficient. The capsular ligaments were remodelled using four ligaments on the periphery of the facet to more effectively control the translation of the surfaces as they moved relative to each other. Table 1 shows the comparison between the data collected from asymptomatic subjects at the Whiplash Physical Diagnostic Clinic, University of Queensland, and the output from the model. As an example of the effects of muscle spasm on the range of motion, unilateral spasm of the scalene muscle group is presented. This response is yet to be compared with aberrant motion patterns demonstrated clinically. The most marked effect from the muscle spasm was demonstrated in flexion because rotation about this action causes the greatest relative length change in the scalene muscle group.

<table>
<thead>
<tr>
<th>Clinic Asymptomatic data</th>
<th>Passive ROM (model)</th>
<th>Avg(Std dev.)</th>
<th>Maximum</th>
<th>X</th>
<th>Y</th>
<th>Z</th>
</tr>
</thead>
<tbody>
<tr>
<td>Passive ROM with unilateral spasm</td>
<td>Maximum</td>
<td>-54 (13)</td>
<td>2 (6)</td>
<td>0 (7)</td>
<td>-50</td>
<td>0</td>
</tr>
<tr>
<td>Passive Flexion</td>
<td>47 (8)</td>
<td>0 (7)</td>
<td>2 (6)</td>
<td>-49</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Passive Left Lateral Flexion</td>
<td>5 (10)</td>
<td>36 (14)</td>
<td>8 (12)</td>
<td>11</td>
<td>33</td>
<td>13</td>
</tr>
<tr>
<td>Passive Right Lateral Flexion</td>
<td>3 (9)</td>
<td>-35 (13)</td>
<td>-8 (10)</td>
<td>11</td>
<td>-33</td>
<td>-13</td>
</tr>
<tr>
<td>Passive Left Rotation</td>
<td>2 (13)</td>
<td>1 (10)</td>
<td>70 (23)</td>
<td>21</td>
<td>5</td>
<td>55</td>
</tr>
<tr>
<td>Passive Right Rotation</td>
<td>4 (11)</td>
<td>0 (9)</td>
<td>-65 (15)</td>
<td>21</td>
<td>-5</td>
<td>-55</td>
</tr>
<tr>
<td>Passive ROM with unilateral spasm</td>
<td>Maximum</td>
<td>12</td>
<td>6</td>
<td>51</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 1. Comparison of range of motion data from asymptomatic subjects and the output of the model. The angles of rotation shown are successive Euler angles about the initial X (left) Y (rearward) and Z (upwards) axes as defined by the clinical set-up, with the primary axis being the axis of primary motion. The model output data is representative and will be subject to further validation.

The simulation of postero-anterior manipulation highlighted the sensitivity of the model to the force application rate. The viscous damping component contributed up to 95% of the peak force during rapid manipulation, but significantly lower peak forces were produced for quasi-static loading. The explicit tissue stiffening that had been incorporated in the model to account for the deformation rate during impact was disabled for the clinical assessment model.

Simulation of the deep neck flexor test that evaluates the control and stability functions of the deep neck muscles confirmed that the longus colli / longus capitus group of muscles are the most effective at
reducing the lordosis of the cervical spine which causes the measured pressure increase in the pressure biofeedback unit. The hyoid group of muscles can be recruited at a low level to assist the deep muscles but at higher activation levels they perform like the superficial muscles causing head lift, with a subsequent reduction in airbag pressure as demonstrated in Figure 2. Figure 3 shows the configuration of the airbag during the test. The muscles and skin surface have been removed to show the bodies of the cervical spine model.

![Force on PBU during DNF test](image1)

**Figure 2.** Force time relationship for the activation of flexor muscles showing the ability of the longus colli/capitus group to hold a reduction in cervical lordosis.

![Configuration for the deep neck flexor test with skin surface and muscles removed.](image2)

**Figure 3.**

**Discussion**
The influence of a localised injury or deficit on the segmental response, and the relationship between these changes and their clinical manifestation can be analysed and visualised using a biomechanical model. The analysis of the internal forces and relative motion between segments in the neck in response to physiological movement and during diagnostic procedures will be further directed toward correlating the severity of a particular dysfunction with the clinical symptoms. In terms of rehabilitation assessment, biomechanical modelling of the segmental response to exercise or manipulation can be used to test the effectiveness of the prescribed techniques.

**References**
Whiplash Diagnostic Clinic, Range of motion data, University of Queensland, Australia.