INTRODUCTION

THE whiplash INJURIES include a variety of symptoms and their definitions are undetermined and controversial. The aim of this study is to reconstruct the cervical response and predict the neck soft tissue injuries during rear impacts for better seat design. A head-neck unit model was developed based on the finite element human model, called "THUMS (Total HUMAN Model for Safety)”, to better understand the head-neck kinematics. The cervical model (Hasegawa, 2003) was validated against the test data (Cholewicki, 1998) with human cadavers. Since the facet joints in the cervical spine are thought to be one of the causes of pain, the study focused on their kinematics during rear impacts.

WHOLE BODY MODEL VALIDATION

For the evaluation of whiplash injury occurring in rear-end collisions, a detailed head-neck model was developed and incorporated into THUMS model. Then the head and neck kinematics was validated against that of living human subjects sitting on a rigid seat without muscular tension (Ono, 1999). The tests were carried out at delta V of 8 km/h. A finite element code, LS-DYNA3D, was used for the analysis. The material property data defined for the human tissues were referred to the author’s previous report (Hasegawa, 2003). Figure 1 shows the calculated global kinematics. Comparison of the angles in head-neck rotation between the test result and calculated result is shown in Fig. 2. A good correlation was found for the cervical vertebrae up to 100 ms while a slight difference was noted for the head. A possible reason for the discrepancy after 100 ms in two results is the activation of neck muscles, which is thought to be generating around 100 ms after impact (Ono, 1999). Because the model included only the passive effects of muscles, it was difficult to accurately reproduce the human-like response over a long duration.

KINEMATICS AND INJURY PREDICTION OF FACET JOINTS

Figure 3 and 4 show the kinematics of the C4-C5 facet joint and the relative displacements respectively. Displacement of the C4 was measured at a point A based on a reference point O on the C5. It was noted that the C4 slid posteriorly relative to the C5. Due to the relative rotation between two vertebrae, compressive force was generated in the posterior region of each facet joint. Figure 5 shows the resultant contact force between adjacent vertebrae. As there was little information about the mechanical properties of spinal ligaments, which is thought to produce pain in the range of nerve ending firing, a failure strain of joint capsule (JC) proposed by Joganandan (1998) was used instead, as a first step for the injury prediction. Figure 6 shows the time history of the strain in the lateral part of each JC, with the failure strain level shown as a dotted line. The JC strains at C4-C5 and C6-C7 exceeded the failure value in the calculate result, while there is no cervical injury in the volunteer test. This is possibly due to the lack of muscle activation in the model. Figure 7 shows the relative rotational
angle between adjacent vertebrae. A similar trend was observed for the history curves in Fig. 6 and 7. It suggests that the rotational angle between the vertebrae could be related to the JC strain. In this study, it was assumed that the nerve could be damaged when JC is ruptured at failure strain. And the damage can start when the rotational angle of each cervical vertebra exceeds the physiological range of motion (ROM). The physiological ROM of the cervical spine in extension is shown in Table 1 (Marko de Jager, 1996). The black circular markers in Fig. 7 indicate the points when the rotational angle reaches the ROM. For the C4-C5 and C6-C7, the points were close to the failure value of 143%. As for the C7-T1, however, the strain level was lower; around 100% at 5 deg. calculated in the model. Extensional rotation at the T1 was smaller than that of the other lower cervical vertebrae, because of the constraint by the ribs. The C6 and C7 vertebrae started rotating simultaneously but the C7 reached its physiological ROM earlier, as shown in Fig 6. This is why CL strain at the C7-T1 facet joint was smaller than that of the C6-C7. Repo et al., (1977) reported that the chondrocytes of the human cartilage specimens could be damaged under impact when pressure level exceeds 25 MPa. In the area pf 100-200 ms of the simulation, it was found that the contact force on facet joint increased with time, as shown in Fig. 5. The peak contact force at the C4-C5 and C7-T1 facet joints is in the range of 300-400 N. Winkelstein (2001) reported that the average length of JC boundary at the C4-C5/C5-C6 level was about 20 mm. By assuming that the contact region is circular and its area is about 30 mm², the average contact stress is estimated to be 13 MPa for the C4-C5 facet joint. If the contact area decreases with time, the estimated stress increases, as shown in Fig. 3. If the pressure level on the cartilage surface is high enough, there may be a risk of injury.

**SUMMARY**

The calculated cervical spine kinematics showed a good correlation with the volunteer's test data. Although further improvement will be necessary for more accuracy, the model was capable of estimating facet joint kinematics. It is recognized that the human FE model could be an efficient tool to better understand injury mechanisms.

**REFERENCES**


<table>
<thead>
<tr>
<th>Table 1</th>
<th>physiological ranges of motion of cervical spines in extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Angle (deg.)</td>
<td>C2-C3</td>
</tr>
<tr>
<td>4.0</td>
<td>6.7</td>
</tr>
</tbody>
</table>