BIOMECHANICAL ANALYSES OF WHIPLASH INJURY IN THE REAE-END IMPACTS

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CHAPTER 1

GENERAL INTRODUCTION

1.1 Injuries data in car accidents

Recently in Japan, there are more than 1,000,000 people injured in car accidents every year [(1)] as shown in Fig. 1.1, and human loss has amounted about 1,600 billion yen based on Ref. [2] regarding insurance payouts during the year from April 2003 to March 2004. This indicates that traffic accidents involve enormous social loss, and it is very significant issue to prevent traffic accidents. Neck injuries, in particularly, are most suffered in car accidents, there are approximately 600,000 people that are about 46 % of all injuries in the year 2003 (Fig. 1.2), and increase by 16 % compared to 5 years ago. Neck injuries are notably occurred in rear-end collisions, about 360,000 people injured, that is 77 % of all injuries (Fig. 1.3). Therefore, many attempt have been tried to reduce neck injuries and rear-end accidents.

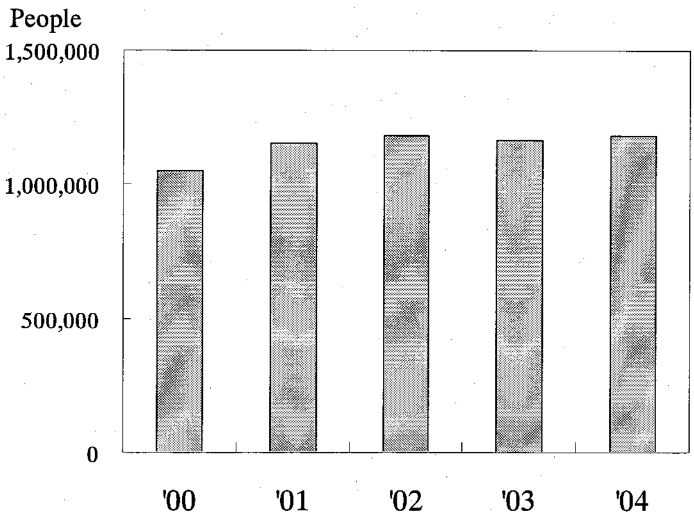


Fig. 1.1 Amount of people injured in car accidents from 2000 to 2004 [(1)]

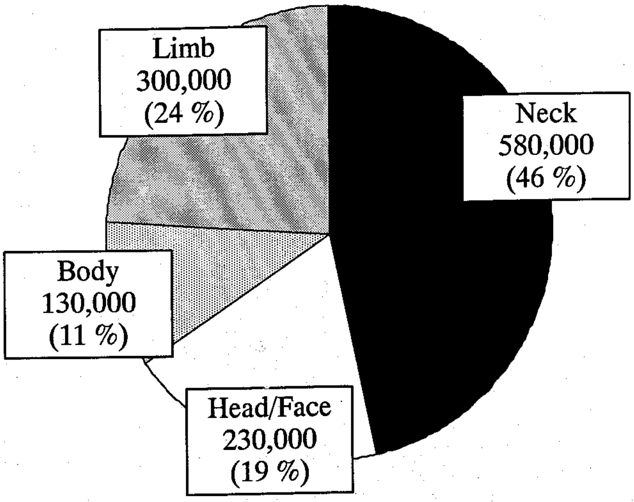


Fig. 1.2 Injured parts and amount of people car accidents

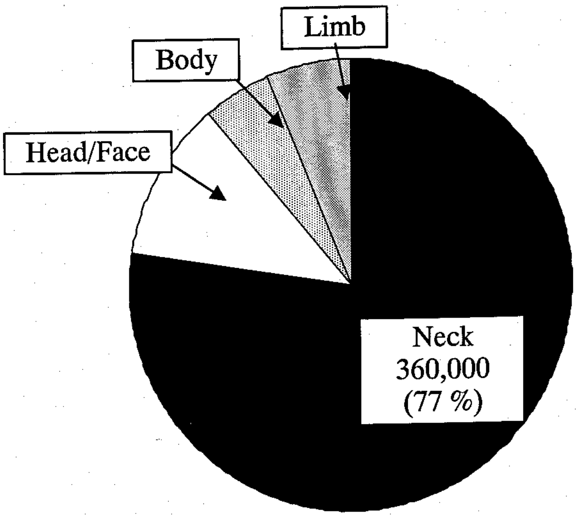


Fig. 1.3 Injured parts in rear-end collisions

1.2 Whiplash injury

"Whiplash injury" is well known as a common injury suffered cervical spine in car accidents. The name Whiplash injury, first described by Crowe in 1928 [(3)], derives from the phenomena that the head and neck suddenly move like a whipping movement at the moment of a traffic accident. Injury symptoms include neck pain, headache, blurred vision, tinnitus, dizziness, nausea, and back pain, and so on. Although many symptoms are reported in clinical cases, the sufficient resolution has not been found to identify these injuries even with presently available imaging methods such as MRI, X-ray and CT-scan. It is difficult to describe clinically the occurrence mechanism, therefore, whiplash injuries have been classified medically depending on typical symptoms [(4)~(6)].

1.3 Studies and mechanisms of whiplash injury

In order to design a prevent system for whiplash injury, it is important to clarify the occurrence mechanism of whiplash injury. Many studies have been conducted to reveal the mechanism of whiplash injury, and various hypotheses were suggested. As the initial theory, Macnab suggested the hyperextension mechanism that the head and neck was extended severely to backward and the anterior injury of cervical spine occurred (Fig. 1.4 (a)) [(7)]. Based on the hypothesis, the head rest was designed to prevent the extension of neck, but the reduction of neck injuries was relatively small. Penning hypothesized the hyper-translation as abnormally large translation that a head moved to forward or backward with respect to the trunk without flexion or extension before supporting of the head rest (Fig. 1.4 (b))[(8), (9)]. He found cranio-vertebral junction (C0-C2) injuries, but this was not a case for the lower cervical spine. Then, Panjabi et al. suggested the S-curve mechanism that the cervical spine was formed the S-shaped curvature, which primarily produces hyper-extension at the lower cervical and flexion at the upper cervical (Fig. 1.4 (c)) [(10), (11)]. They reported that whiplash injuries were resulted in the soft tissues injury caused by the intervertebral extensions in the lower cervical spine beyond their physiological ranges and maximally elongation of capsular ligaments and vertebral artery. There are some studies focused the correlation with the facet joint injury. Kaneoka et al. hypothesized that facet collisions were likely to impinge on and inflame the synovial folds in the zygapophysial joints at C5-C6, causing neck pain (facet synovial fold impingement syndrome) [(12)]. Yoganandan et al. suggested that the pinching mechanism due to compression and sliding of the facet joints caused the injury, which elicited neck pain [(13)]. Tencer et al.

reported the facet shearing as a primary mechanism of whiplash injury [14]. In above studies, vertebral motions were mainly focused as the cause of soft tissues injury. While Sevensson et al. hypothesized that the nerve root region in the cervical spine was injured by the result of transient pressure gradients in the spinal canal during rapid neck bending [15]. In addition, they suggested the injury criterion based on their hypothesis [16].

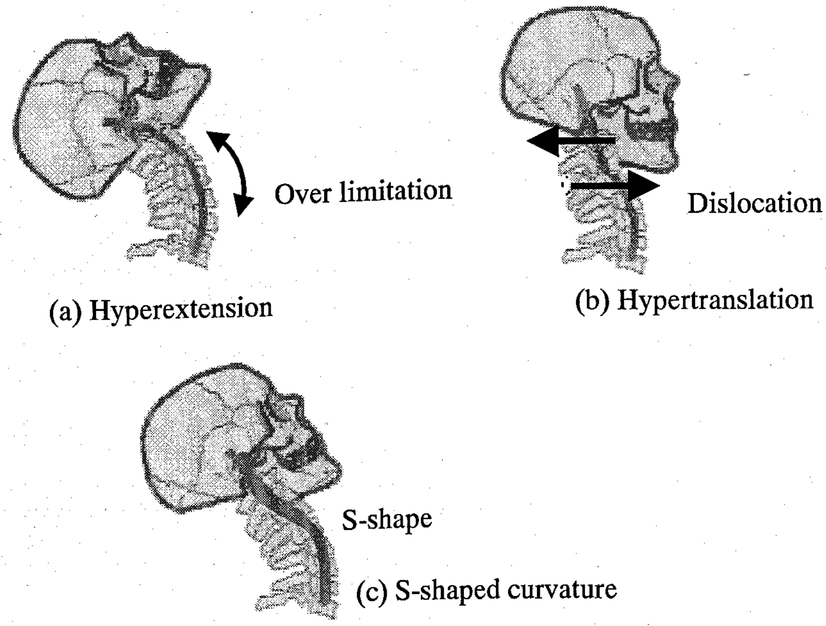


Fig. 1.4 Mechanisms of whiplash injury based on the neck motion

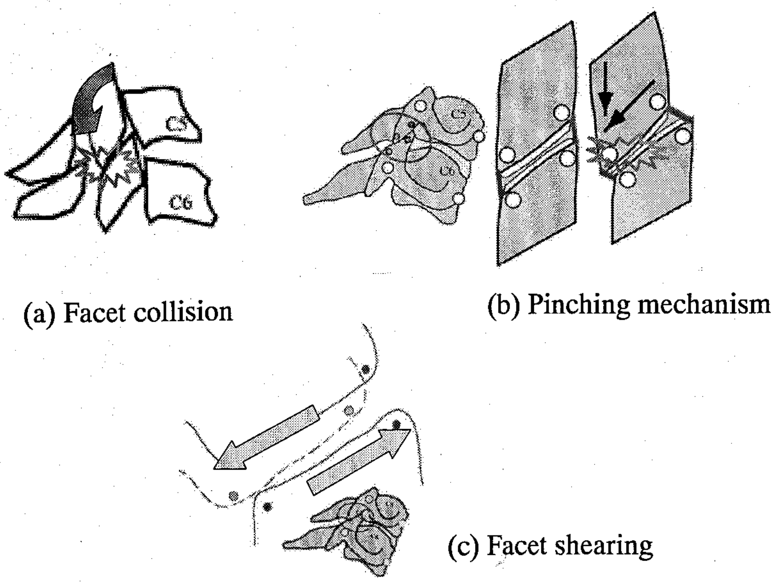


Fig. 1.5 Mechanisms of whiplash injury based on the facet injury

Above hypotheses were suggested based on the investigation of cervical behaviors in a rear-end impact. There are two types methods for reproducing cervical behaviors in a rear-end impact: one is an experimental method using the physical testing such as animals, anthropometric dummies, cadavers and volunteers. Another is a numerical method using the computer simulation. Even in any studies, there are both advantages and disadvantages. In the animal testing [(7) ~ (9), (15)], it could be investigated the change of living tissues caused by an impact, but there were some differences between animals and human, e.g., anatomical shape, mechanical and physiological properties, etc.

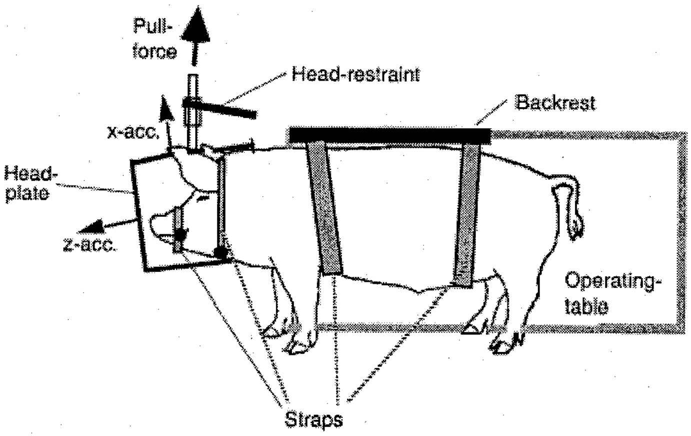
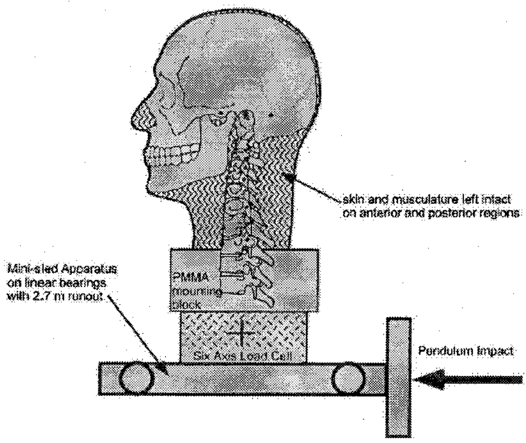
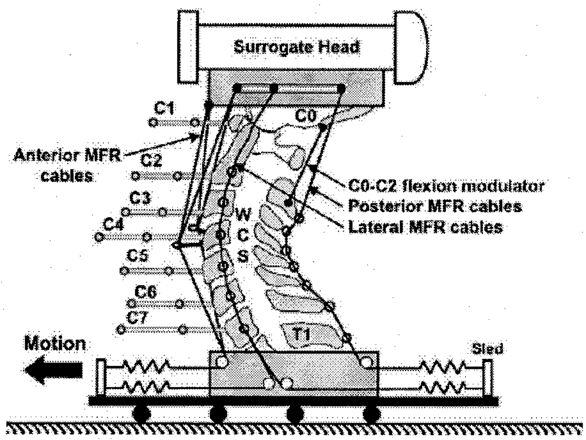


Fig. 1.6 Example of an animal testing method with a pig

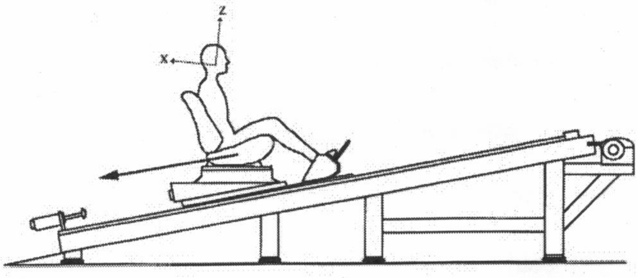


(a) Testing by Yoganandan et al.



(b) Testing by Panjabi et al.

Fig. 1.7 Examples of cadaver testing methods with isolated cervical spine



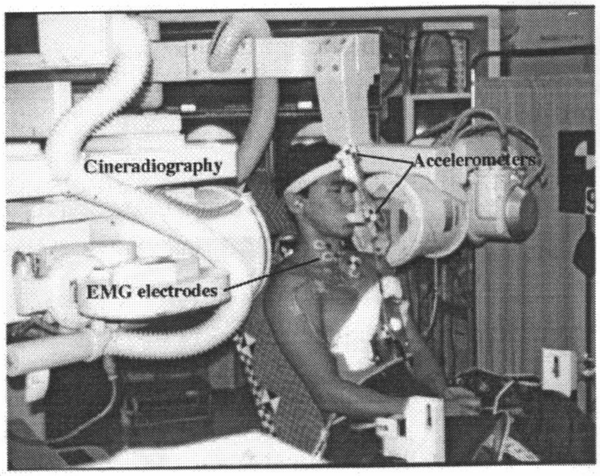
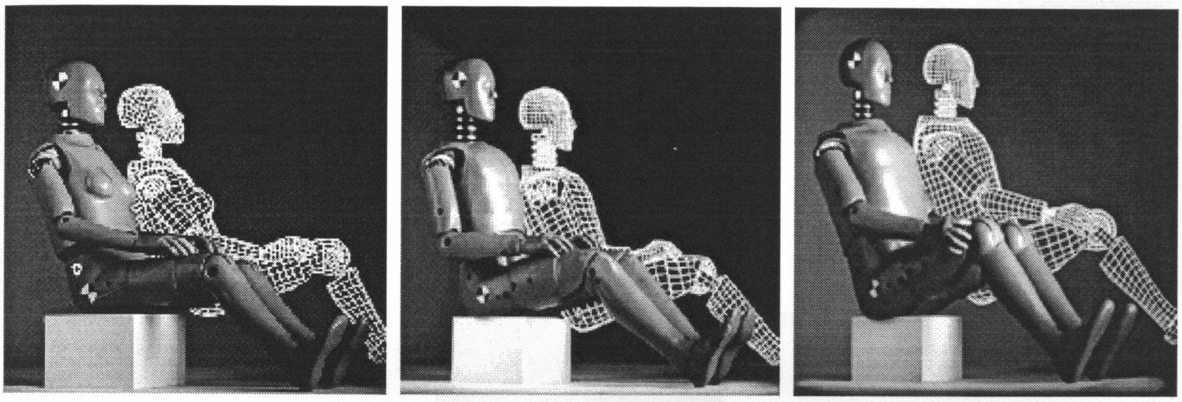


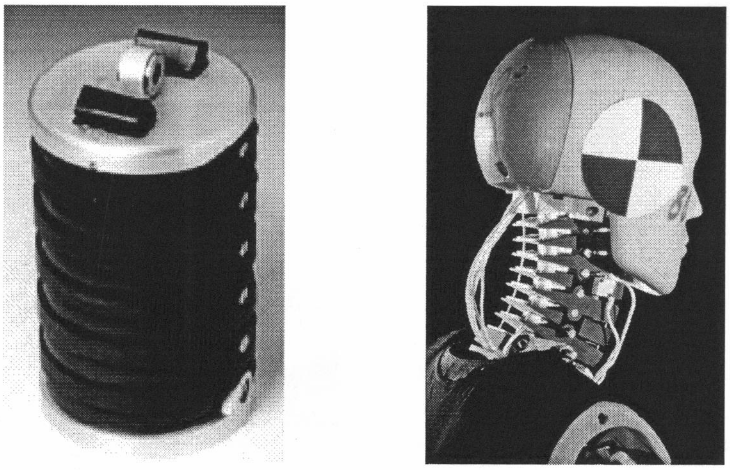
Fig. 1.8 Examples of volunteer testing methods

Anthropometric dummies such as the Hybrid-Ⅲ have been developed to investigate occupant behaviors in car collisions [(17), (18)]. The Hybrid-Ⅲ dummy, however, has conventionally been used, the neck part was much stiffer than the neck of living human because it was made of aluminum and hard rubber to use in the high speed frontal impact test [(19), (20)]. Thus, It was developed that dummy neck models such as the TRID-neck [(21)] and BioRID [(22)] improved more flexibility than the Hybrid-Ⅲ dummy, while they were not anatomically-correct, and connected with hinge joints, it was difficult to validate the bio-fidelity. Then, whole body cadavers or isolated cervical spine specimens were used to identify the injury mechanism [(10), (11), (23), (24)]. There are many advantages compared to use animals or dummies, but there are many ethical, legal and social concerns with using cadavers. Furthermore, cadaver's muscles have no activity and cannot support their posture themselves. While volunteers may contract their neck muscles in anticipation of an impact because they can know the beginning of impact (12), (25) ~ (28). In real accidents, occupants may be shocked involuntarily. Therefore, in cadavers or volunteers testing methods, cervical behaviors may not be reproduced the situation of occupants in real accidents.



(a) 5th female model (b) 50th male model (c) 95th male model

Fig. 1.9 Hybrid-Ⅲ dummies



(a) TRID-neck model (b) Bio-RID model

Fig. 1.10 Dummy neck models for rear-end impact test

From the view points, a new biomechanical neck model called K-D neck was developed to reproduce cervical behaviors of occupants unable to anticipate impending impact (29).

Yoshida et al. evaluated cervical behaviors with the Hybrid-Ⅲ dummy whose neck was replaced by K-D neck, and found that severe shear movement in an anteroposterior direction occurred between the second cervical vertebra (C2) and third cervical vertebra (C3). Based on these findings, new drivers' seat have been developed to prevent whiplash injury [30]. They also conducted numerical analyses with pseudo three-dimensional finite element head-neck model in consideration of muscle activities [31].

Fig. 1.11 K-D neck model

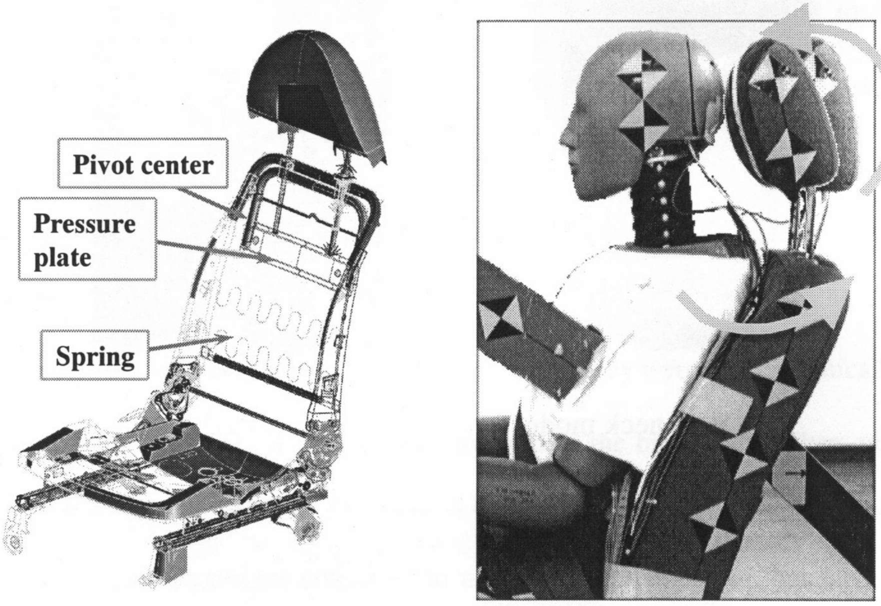


Fig. 1.12 New driver's seat with active head restraint for reduction of whiplash injury

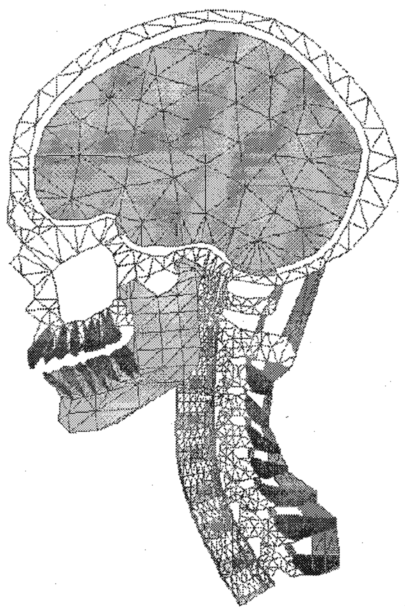


Fig. 1.13 Pseudo 3-D FEM head-neck model

Numerical analysis such as finite element method is an effective method for evaluating the biomechanical change in the cervical spine under impact loadings. There are various studies for whiplash injury in a rear-end impact using the numerical analysis [(32) ~ (41)]. In early studies, a numerical model was represented with rigid bodies, and the shape was very simplified [(32) ~ (34)]. The finite element head-neck model represented a human anatomical shape was used, but the cervical behavior was only investigated two-dimensionally in the sagittal section [(35)]. Although three-dimensional model was developed, the shape was simplified elliptically as a rigid body yet [(36), (37)]. Recently, three-dimensional finite element human model have been developed to investigate biomechanical responses under impact loadings in accidental situations [(38), (39)].

In the method with cadavers or volunteers, it is difficult to directly measure and investigate the internal biomechanical phenomena. However, in order to evaluate the mechanism of whiplash injury, it is important to investigate damages on the soft tissues in the cervical spine. From the view point, finite element models are effective tools for biomechanical analyses of the cervical spine under impact loading conditions.

1.4 Outline of this thesis

The purpose of this study is to evaluate dynamic neck motions and biomechanical changes in the cervical spine under impact loading in rear-end car collisions with 3-D finite element human model, in order to suggest of developing a prevent system for whiplash injury. In many previous studies, two-dimensional cervical behaviors have been mainly observed in the sagittal plane. However, in real accidents, collisions may occur from any direction and occupants may be facing directions other than forward when they happen, cervical behaviors can be more complex and three-dimensional. From the view points, this study was conducted as follows contents:

CHAPTER 1; The general information and background of this research were described about biomechanical studies for whiplash injury in rear-end car collisions.

CHAPTER 2; A FEM analysis was conducted to investigate the cervical behavior in a low-speed rear-end impact with three-dimensional finite element human whole body model-THUMS. It was simulated the rear-end impact on the assumption that the impact occurred from an oblique direction.

CHAPTER 3; It was to be evaluated the effect of impact directions in posterior-oblique impacts on the cervical motion and the soft tissues injury with the THUMS model. The vertebral behaviors were investigated three-dimensionally in the sagittal, lateral and axial plane.

CHAPTER 4; A new injury criterion was proposed to validate a neck injury in rear-end impacts. It was investigated the availability of our criterion to predict the soft tissues injury with the finite element analysis.

CHAPTER 5; It was to be analyzed the effect of head rotation on the injury severity of the soft tissues in the cervical spine during whiplash loading. Furthermore, in order to validate the new criterion, we evaluated the correlation between the criterion and the stresses occurred in the soft tissues.

CHAPTER 6; The FEM analyses were conducted in consideration of the muscle activity in neck muscles. The head-cervical behaviors were investigated in state of muscle tension during a rear-end impact. Furthermore, it was evaluated the effect of the reflexive time until beginning the muscle contraction.

CHAPTER 7; Summaries of this study and discussions of future investigation were presented.

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CHAPTER 2

NUMERICAL ANALYSIS OF THE CERVICAL BEHAVIORS IN A POSTERIOR-OBLIQUE IMPACT WITH 3-D FINITE ELEMENT HUMAN WHOLE BODY MODEL

2.1 Introduction

Neck injuries in rear-end collision occur frequently in automobile accidents. The percentage of rear-end accidents is about 50 % of all automobile accidents in Japan, and the rate of slight injuries is very high, 90 % of rear-end accidents result in neck injuries [(1), (2)]. Neck injuries in rear-end accidents are well known as Whiplash Injury and can occur even at low speed below 5 km/h. Injury symptoms include pain, weakness or abnormal response in various parts of the human body, mainly the neck, shoulder and upper back, which are connected to the central nervous system via the cervical nerve-root. As other symptoms, vision disorders, dizziness, headaches, and neurological symptoms in upper extremities have been reported. However, the mechanism of whiplash injuries is unknown yet.

Many studies have been conducted to reveal the mechanism of whiplash injuries with the experiment using volunteers [(3) ~ (6)], cadavers [(7) ~ (11)], anthropometric dummies [(12) ~ (16)] and the finite element analysis [(17), (18)]. It is considered that whiplash injuries were resulted in the damage of soft tissues of a cervical spine due to the complex cervical behavior during a rear-end collision. Studies of whiplash injuries have been performed with just two-dimensional analysis in the sagittal section, e.g. the observation of vertebral body's behavior with X-ray. However, the shape of a cervical spine is very complex and three-dimensional morphology. Furthermore, collisions were happening in various directions on real accidents like frontal, side, oblique and offset-collision. Therefore, it is not enough to observe the cervical behavior in only two-dimensional sagittal section.

In this study, we investigated cervical behaviors with the three-dimensional finite element human whole body model. Each relative vertebral behavior were observed in more detail considering a rear-end collision from an oblique direction. Using the three-dimensional human whole body model, we could evaluate complex cervical behaviors, which could not observe with two-dimensional analysis.

2.2 Finite element model and methods

2.2.1 Finite element model

Figure 2.1 shows the three-dimensional finite element model used in this study. The FEM model was consisted from four parts: 1) seat, 2) seat belt 3) floor board and 3) human body. The finite element human whole body model, which is called THUMS(Total HUman Model for Safety, automobile occupant model Ver1.52 ß-031117, TOYOTA Central R&D Labs., Inc.), was used as the human body part. THUMS used in this study represents 50 percentile American male with 175 cm height and weighing 77 kg in a seating posture. The model contains about 60,000 nodes and 80,000 elements.

Each bone consists of cancellous bone modeled using solid elements and cortical bone modeled using shell elements. In joints of the THUMS, ligaments that connect to bones are modeled using shell or beam elements and sliding interfaces are defined on contacting surfaces of these bones. Skins and muscles that cover bones are modeled with solid elements. Internal organs and brain are modeled as continuum bodies with solid elements. The material properties of tissues have been taken from literature [(19)]. This model was validated for frontal and/or lateral impacts to the thorax, abdomen hip, internal organs, brain, and pedestrian [(20), (21)]. The cervical parts of THUMS are also constructed to simulate adequately the anatomical shape and

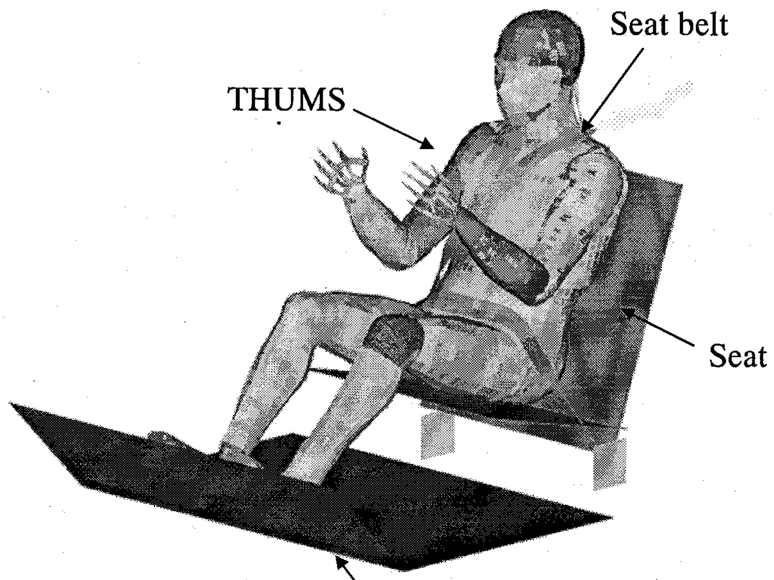


Fig. 2.1 3-D finite element model used in this study

Table. 1 Each component of neck segment in the THUMS

Segment	Young's Modulus [MPa]	Poisson ratio	Element type
Cortical bone	5000	0.3	Solid
Cancellous bone	70	0.3	Solid
Nucleus pulposus	0.198	0.499	Solid
Annulus fibrosus	13.3	0.4	Solid
Facet cartilages	12.6	0.4	Solid
LF	15.07	0.22	Shell
LN	30.16	0.22	Shell
ITL	15.08	0.22	Shell
ALL	3.25	0.22	Shell
PLL	3.25	0.22	Shell
Muscles	Hill-type model (passive state)	—	Beam

properties of the human cervical spine. The modeling details of each component of neck segment in the THUMS are shown in Table. 1.

In order to validate the bio-fidelity of the cervical spine in the THUMS, we compared the global head-neck motion and local cervical vertebral motion of THUMS with the experiment result. To verify the global motion, the horizontal displacement of the head relative to the first thoracic vertebra (T1) of THUMS was compared with that in our previous experimental result [(16)] which were obtained in the rear-end sled test at 8 km/h using the K-D neck. To verify the vertebral motion, the relationship between the rotation angle and bending moment during the flexion-extension motion of adjacent vertebra was compared with the experimental data using cadaveric cervical specimens [(22)].

2.2.2 Finite element analysis condition

The FEM analysis was conducted for the first 200 ms after impact. Impact directions were set as shown in Fig. 2.2. The first case is the conventional rear-end impact at the back (0° impact). The second case is from the right-posterior oblique direction (30° impact), the seat was moved to 30 degrees oblique direction. The velocity curve based on our sled test at 8 km/h was applied to the seat as impact loading as shown in Fig. 2.3. We used LS-DYNA Ver.970 and Hyperworks Ver.6.0 as the FEM solver and pre-post processor, respectively.

2.2.3 Local coordinate system on the vertebral body

In order to investigate cervical behaviors, the local coordinate system was set on each vertebral body (Fig. 2.4). The local coordinate system was positioned as follows: an origin point was determined on the posterior-center of the upper endplate of the vertebral body: X-axis was directed to the anterior-center of the upper endplate of the vertebral body from the origin point: the Z-axis was directed upward in a direction perpendicular to the X-axis: and lastly, the Y-axis was directed laterally perpendicular to both X and Z-axes. We investigated

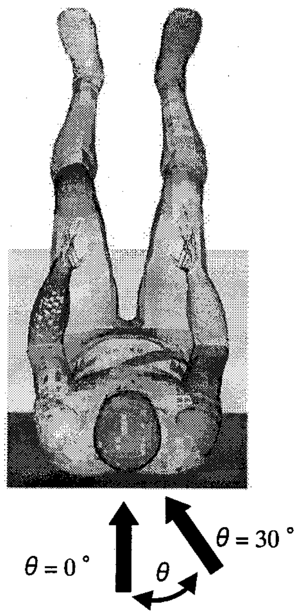


Fig. 2.2 Impact directions

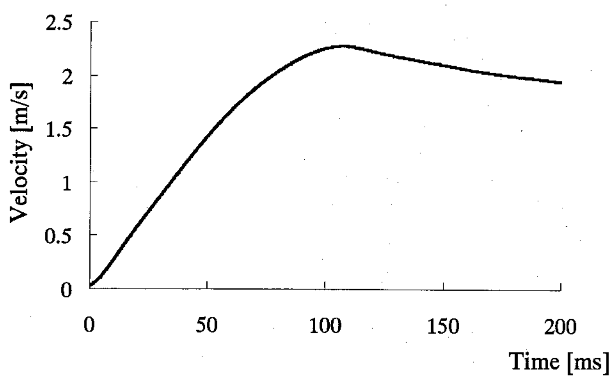


Fig. 2.3 The velocity curve applied the FEM model as the impact loading

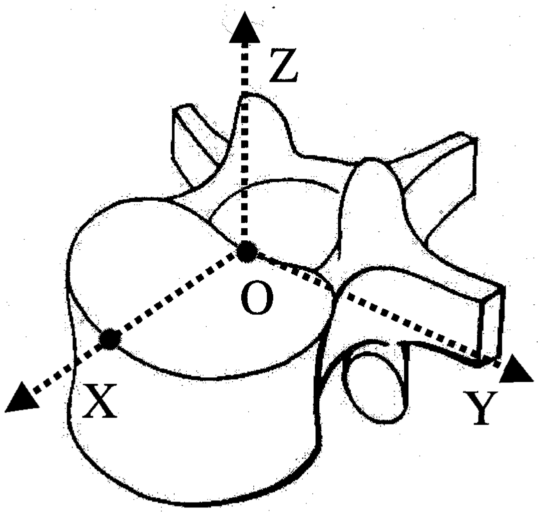
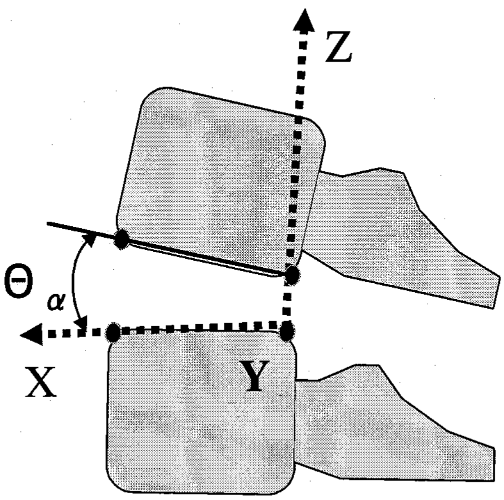
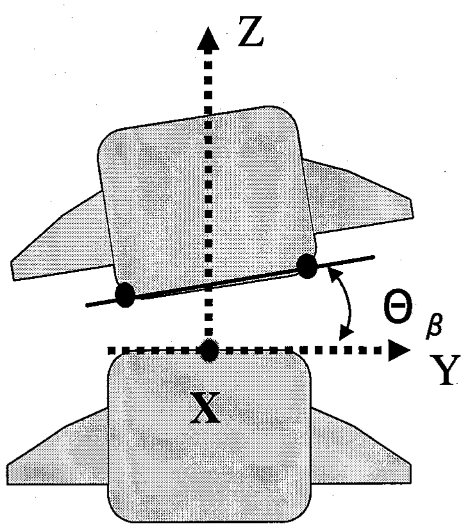


Fig. 2.4 Local coordinate system on the vertebral body

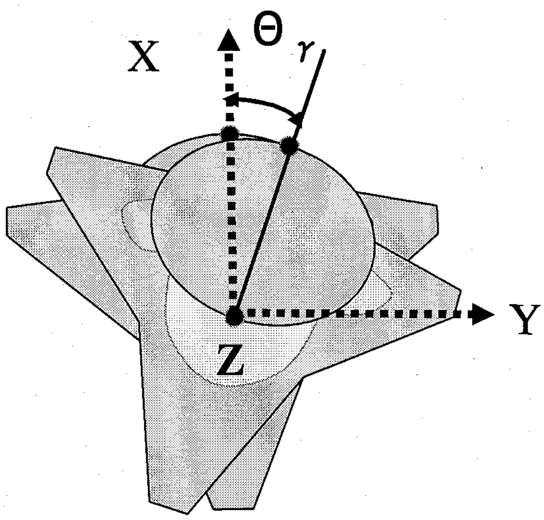
relative rotational behaviors of adjacent superior vertebra against the local coordinate system. Rotation angles around X-, Y- and Z-axis on the local coordinate system were measured in the first 200 ms after an impact, and sampling frequency was 1 kHz. Rotation angles were described (a) θ_a, (b) θ_β, and (c) θ_γ as shown in Fig. 2.5.



(c) Rotation around Y-axis: θ_a
(Flexion-extension)



(b) Rotation around X-axis: $\theta_ß$
(Lateral bending)



(a) Rotation around Z-axis: θ_γ
(Axial torsion)

Fig. 2.5 Rotation angles of vertebral body on the local coordinate system

2.3 Results

2.3.1 FEM model validation

As shown in Fig. 2.6, the head begun to move relative to T1 from 30 ms, and the head displacement peaked at about 120 ms in both the experiment and the FEM analysis. The peak displacement in the FEM analysis was lower than that in experiment. This indicates that the neck stiffness of THUMS was slightly higher than that of K-D neck, because the neck part of THUMS was consisted of elastic elements. However, since the tendency of the displacement was almost same, it seemed that the stiffness of neck part in the THUMS was appropriate.

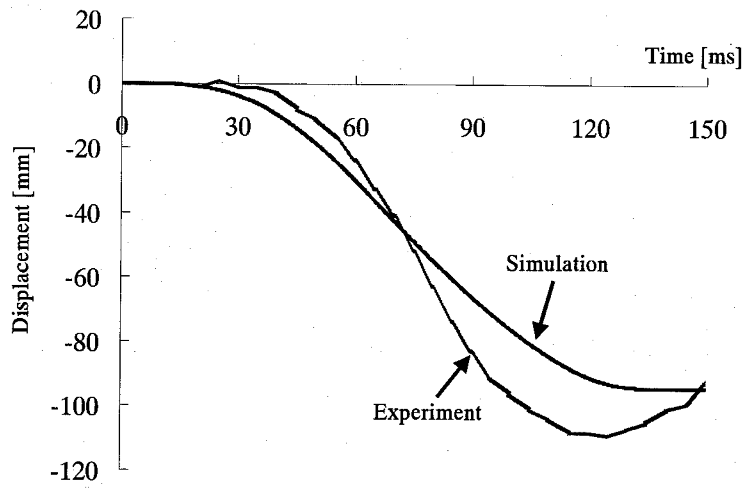


Fig. 2.6 Comparison of the head horizontal displacement relative to the T1 between the simulation and experiment result

In this study, as the vertebral behaviors were evaluated, it was important to verify the bio-fidelity at the vertebra level. The flexion/extension response was verified at C2-C3 and C6-C7 individually because there was the anatomical differences between the upper and lower cervical vertebra. Fig. 2.7 shows the comparison of the rotation angles applied the moment at C2-C3 and C6-C7 between the simulation and experiment. In the extension response, the results of the simulation were in good agreement with the experiment. However, in the flexion response, the rotational angles in the simulation were considerable lower than those in the experiment. It means that the characteristics of the soft tissues around vertebral body in the THUMS were much stiff on the flexion behavior. Therefore, it will need to improve the characteristics of the soft tissues such as ligaments. In the whiplash motions, however, since the backward extension were mainly investigated, FEM analyses were conducted using this model.

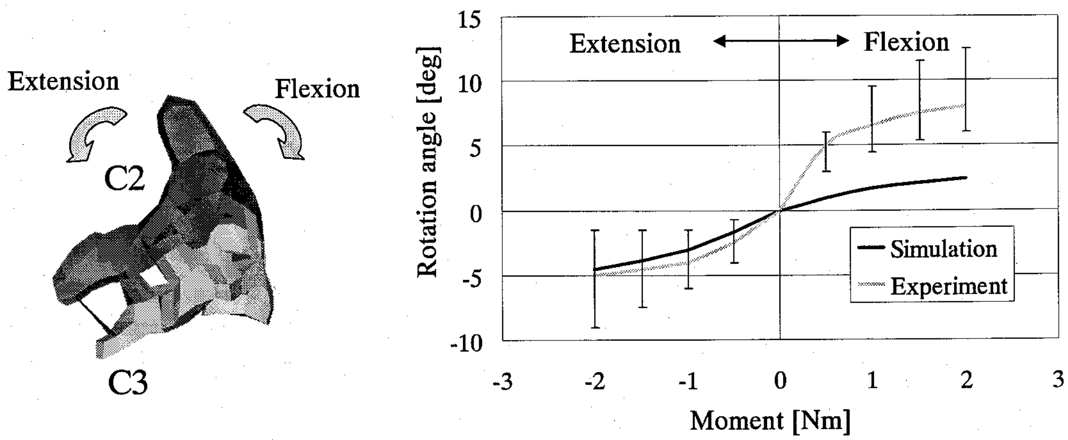


Fig. 2.7(a) Comparison of the flexion / extension responses at C2-C3 between the simulation and experiment result

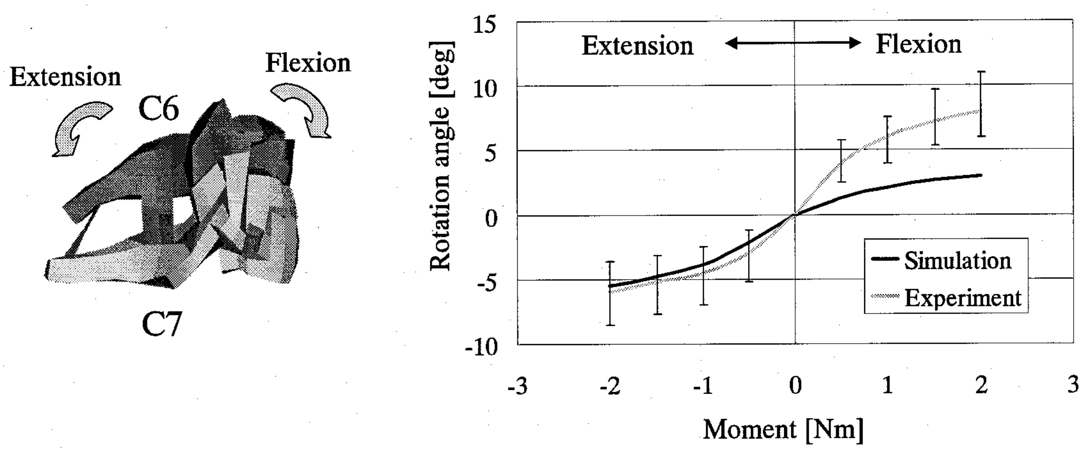


Fig. 2.7(b) Comparison of the flexion / extension responses at C6-C7 between the simulation and experiment result

2.3.2 Whole body motions

Whole body motions after the impact showed differences between 0° and 30° impact. In the 0° impact, the head and neck were extended backward and body motions were almost symmetrical. (Fig. 2.8(a)). In the 30° impact, however, the head and neck were bent to the right with extension backward at 100 ms after the impact, and the whole body moved laterally (Fig.

2.8(b)).

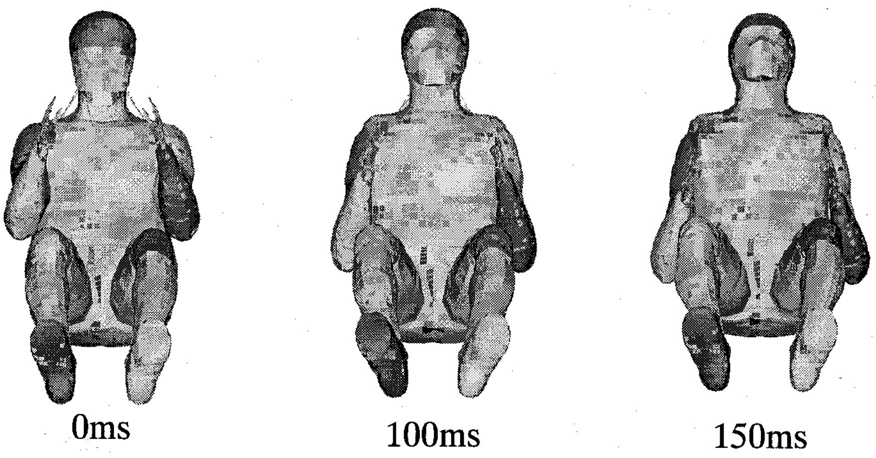


Fig. 2.8(a) The whole body motions after the impact in the 0 ° impact

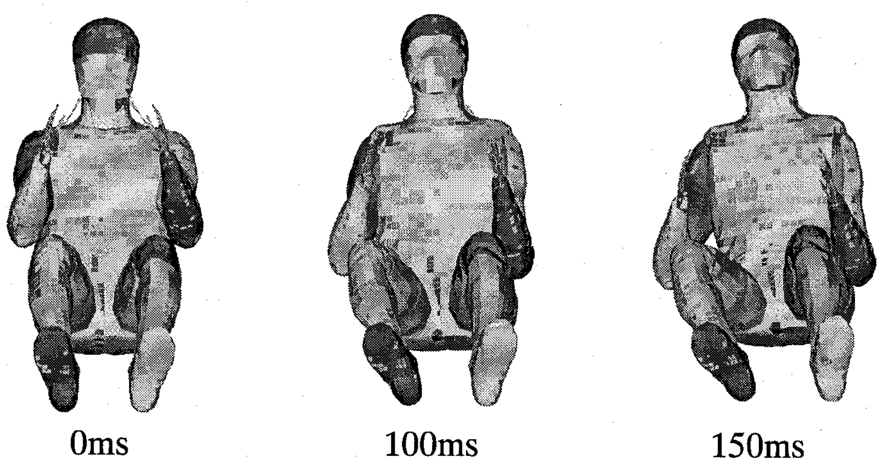


Fig. 2.8(b) The whole body motions after impact in the 30 ° impact

2.3.3 Rotation angles of the vertebral bodies on the local coordinate system

Rotation angles around the Y-axis were almost same between the 0° and 30° impact (Fig. 2.9). As shown in Fig. 2.10 and 2.11, rotation angles around the Z- and X-axis were not occurred. However, in the 30° impact, vertebral bodies were rotated around the X- and Z-axis. It indicates that the cervical spine was twisted and bended in an oblique rear-end impact.

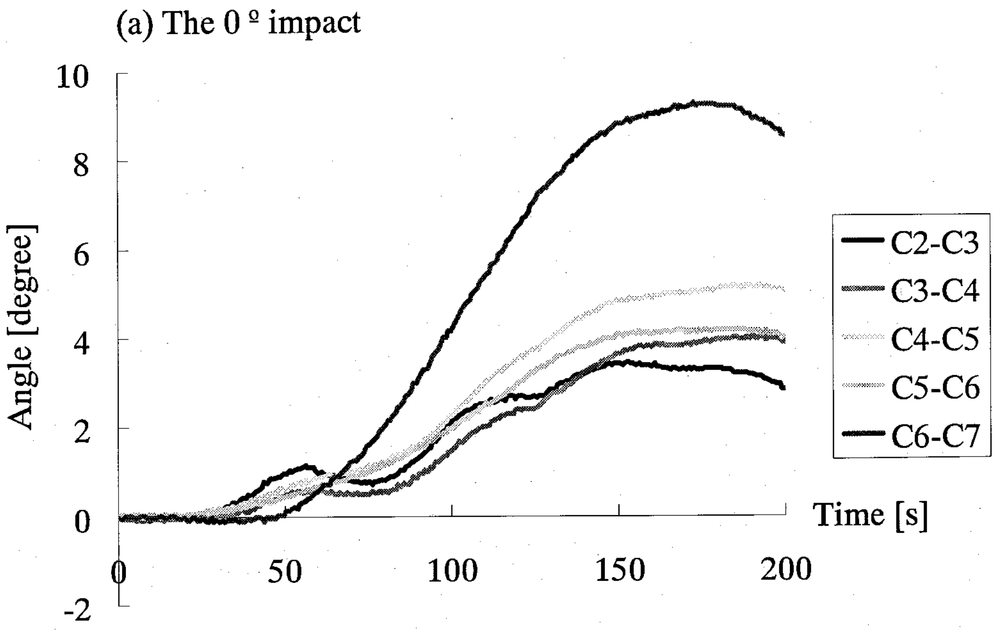


Fig. 2.9(a) Rotation angles of vertebral body around Y-axis in the 0 ° impact

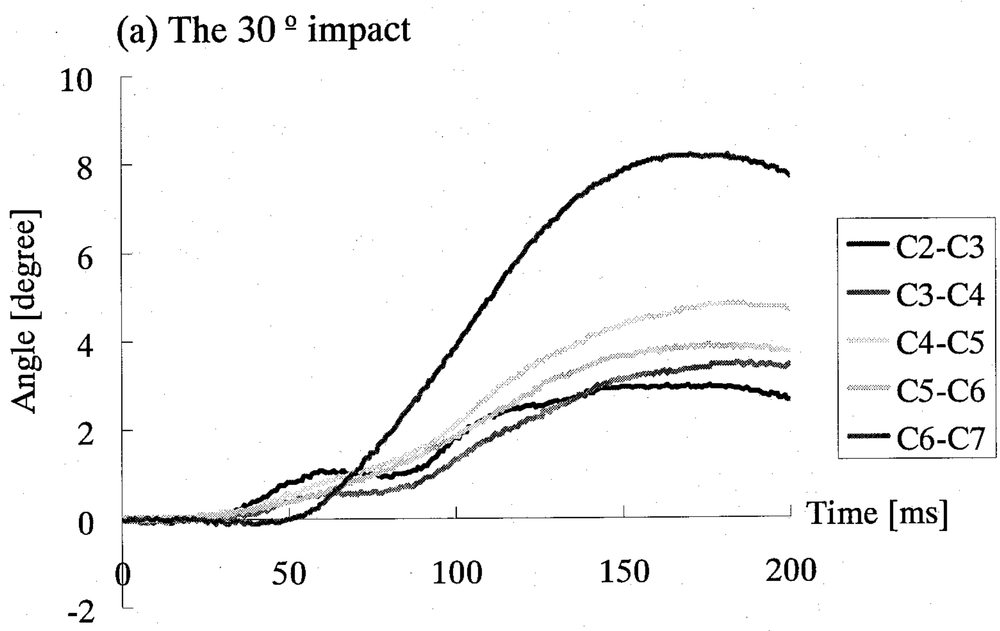


Fig. 2.9(b) Rotation angles of vertebral body around Y-axis in the 30 ° impact

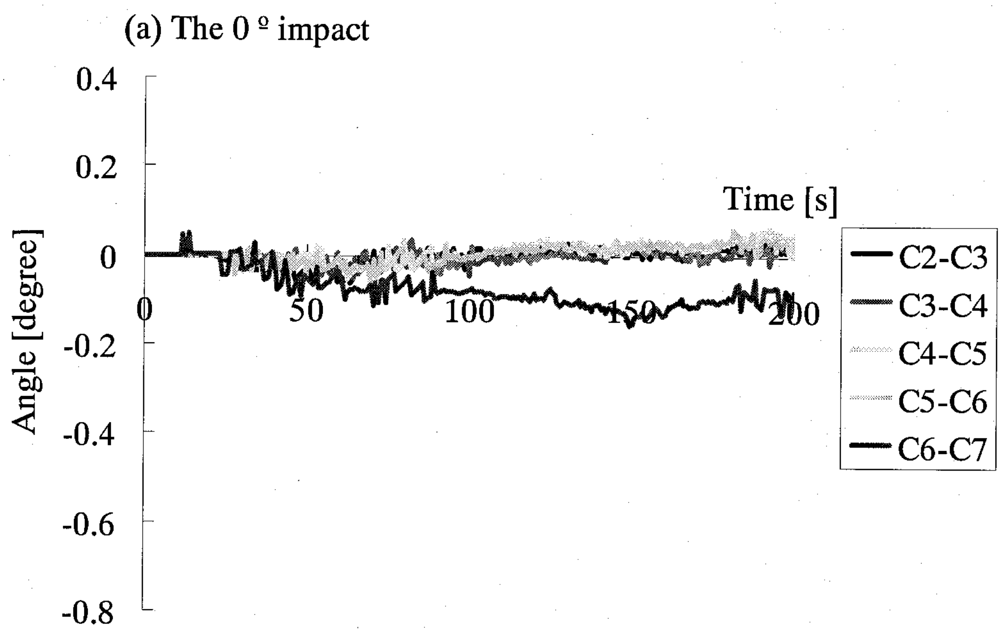


Fig. 2.10(a) Rotation angles of vertebral body around Z-axis in the 0 ° impact

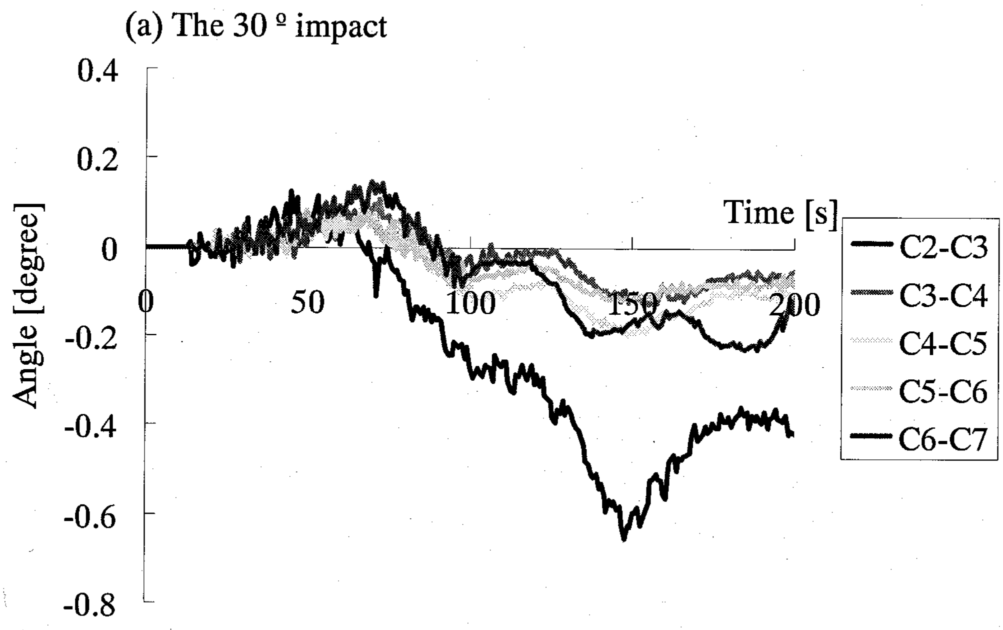


Fig. 2.10(b) Rotation angles of vertebral body around Z-axis in the 30 ° impact

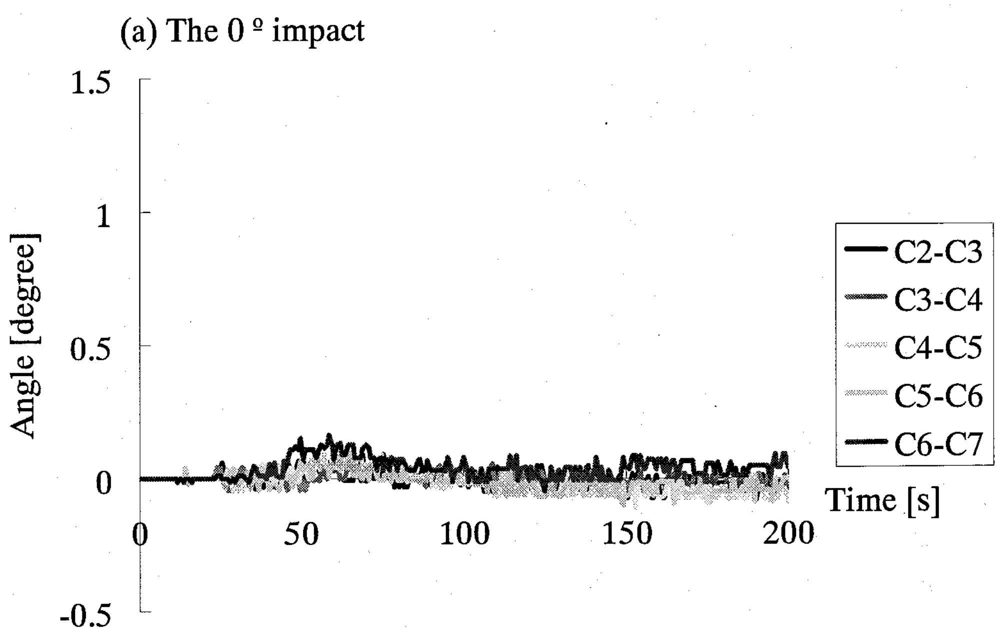


Fig. 2.11(a) Rotation angles of vertebral body around X-axis in the 0 ° impact

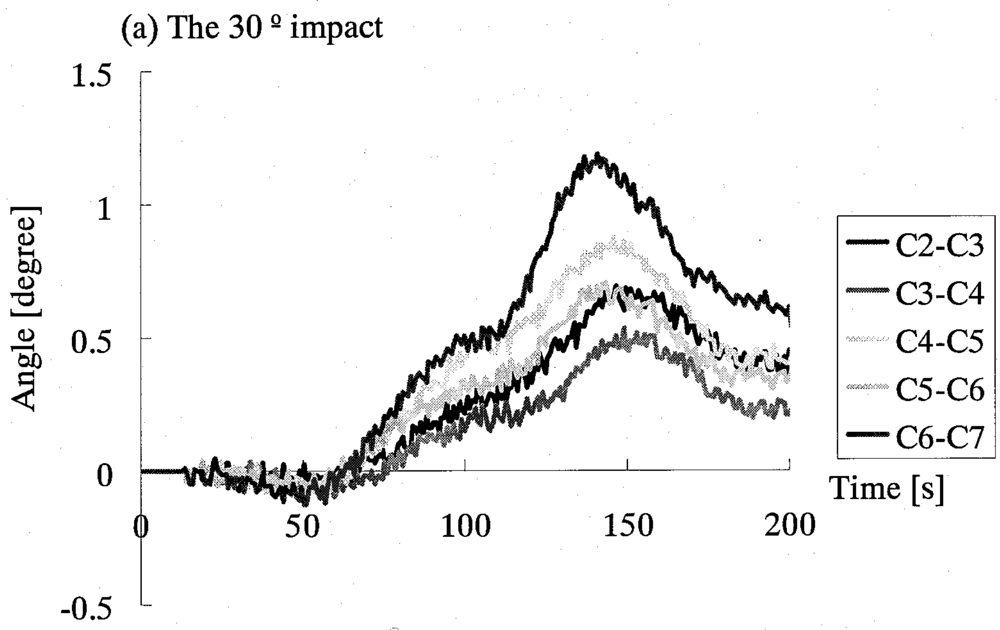


Fig. 2.11(b) Rotation angles of vertebral body around X-axis in the 30 ° impact

2.3.4 Stress analysis on the facet cartilages

The minimum principle (P3) stresses were analyzed on the intervertebral disks and facet cartilages between C5 and C7. In the 0° impact, the stress distribution was symmetric (Fig. 2.12(a)). In the 30° impact, however, the stress distribution was asymmetric (Fig. 2.12(b)), and the P3 stress on the right side facet cartilage was higher than that of the left side in the 30 ° impact (Fig. 2.13). The time history of stress distribution was shown in Fig. 2.12. The P3 stress on the facet cartilage at C6 was higher than that at C5 both in the 0° and 30° impact. In addition, the P3 stresses on right side facet cartilages in the 30° impact were higher than those in the 0° impact (Fig. 2.13). The highest P3 stress was occurred on the right side facet cartilage at C6.

2.4 Discussion

Many studies have been conducted to reveal the mechanism of whiplash injuries. The studies of whiplash injuries have been performed with just two-dimensional analysis. The vertebrae's behavior was observed only in the sagittal section. However, the collision direction is uncertainly on real accidents, and passengers may look aside. Therefore, two-dimensional analysis is not enough to investigate the cervical behavior, as the impact loading is affected various directions to the cervical spine.

In this study, three-dimensional finite element analyses were conducted to investigate the cervical behavior considering the rear-end impact from an oblique direction. As shown in Fig. 2.10, each vertebral body was twisted. It indicates the rotation moment was applied in the cervical spine. The artery passes through the foramen of transverse processes, and there are nerve roots behind the artery in the cervical spine. Since the foramen of transverse's position was misaligned due to the torsional behavior of vertebral bodies, the artery and nerve root might be injured. Especially, torsion angle of C2 on C3 was most largest.

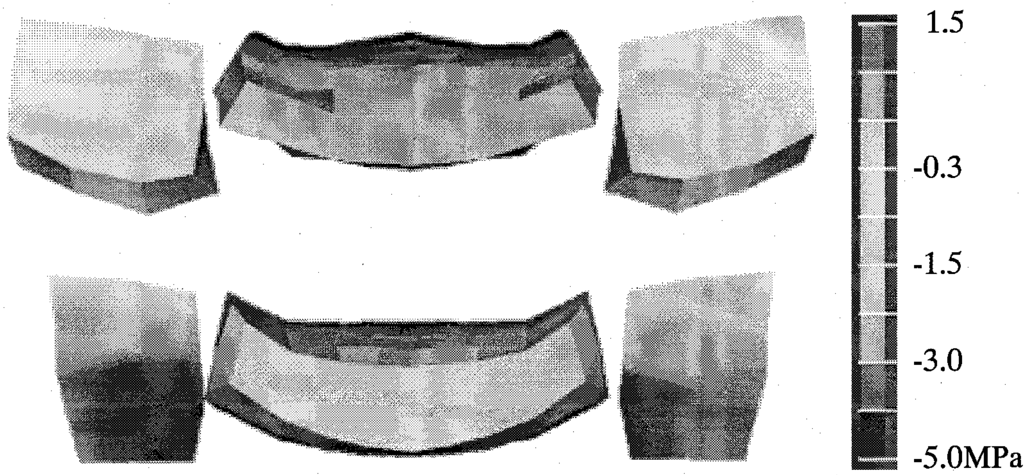


Fig. 2.12(a) P3 stress distribution at 160 ms in the 0 º impact
(Back view of the intervertebral disk and facet cartilages between C5 and C7)

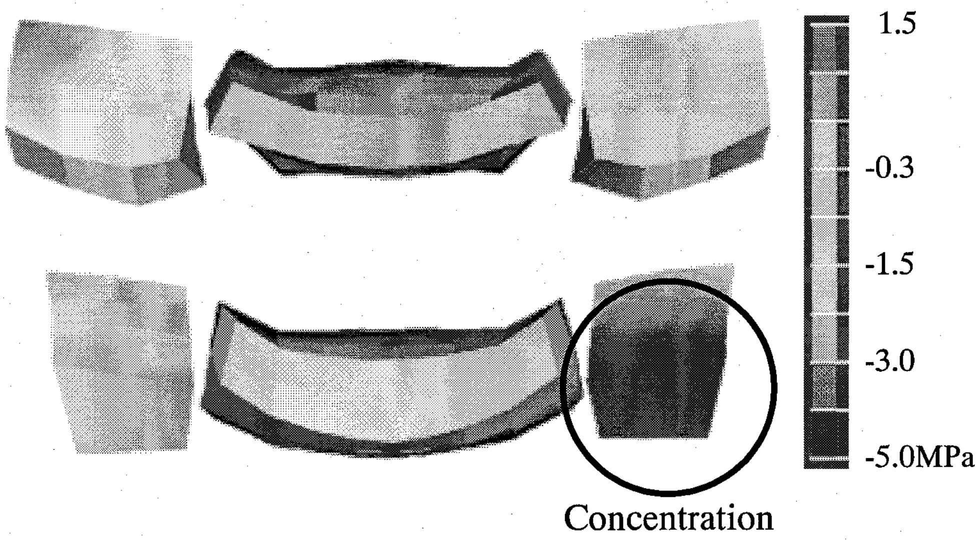


Fig. 2.12(b) P3 stress distribution at 160 ms in the 30 º impact
(Back view of the intervertebral disk and facet cartilages between C5 and C7)

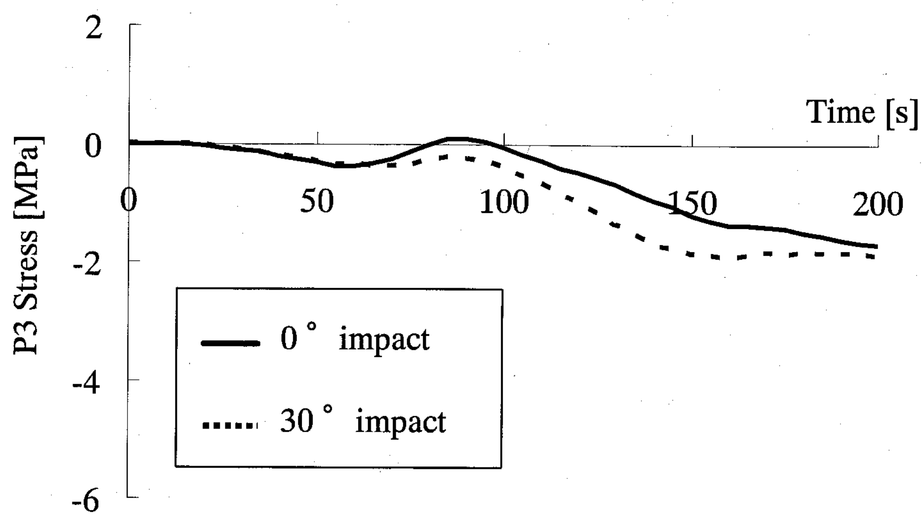


Fig. 2.13(a) The time history of P3 stress on the facet cartilage at C5-C6

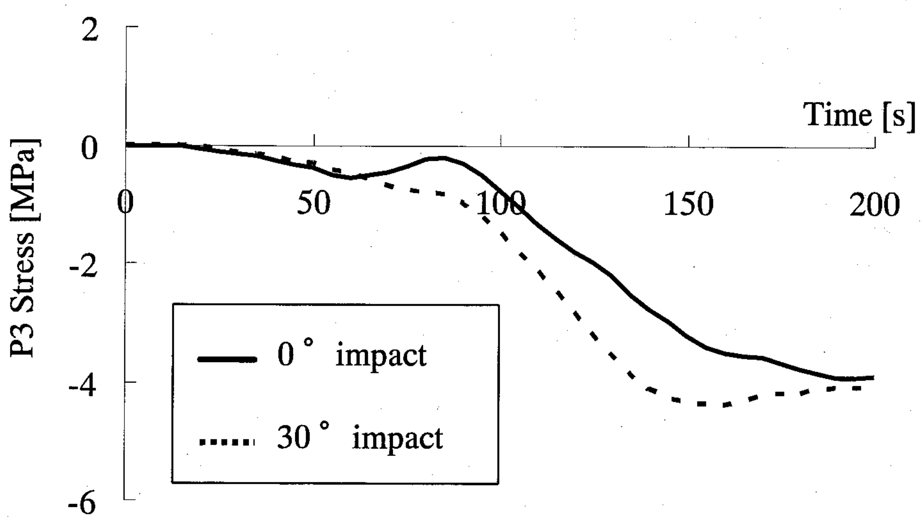


Fig. 2.13(b) The time history of P3 stress on the facet cartilage at C6-C7

It was considered that the soft tissues around the upper cervical spine might be injured easily.

There are many symptoms in whiplash injuries, some symptoms such a Barré-Liéou syndrome are considered due to the blood stream obstruction [23]. Therefore, it was considered that the risk of whiplash injuries increased by the occurrence of torsion in the cervical spine.

In the oblique impact, the cervical spine was bended as shown in Fig. 2.11. It was possible to result in the higher compressive stress on one side (Fig. 2.12, 2.13), and also the higher tension stress could occur on another side compared to the 0 ° impact. Kaneoka et al. reported that the inflammation might occur in the synovial fold by the pinching of the synovial fold between the facet joints because the position of the instantaneous axis of rotation (IAR) shifted [(5)]. Furthermore, Yoganandan et al. also reported that the compression combined with an anteroposterior sliding of the facet joint resulted in the pinching mechanism [(10)]. They suggested that the whiplash injuries were associated with the damage of facet joints. In an oblique impact, it was considered that the IAR shift and facet joint sliding occurred easily because the torsion and bending were included in the cervical behavior. Therefore, it was considered that the risk of whiplash injuries could increase.

Using three-dimensional human whole body finite element model, we could find the complex cervical behavior in rear-end impacts, which could not evaluate with two-dimensional analysis. However, there are some limitations in our study. First, in our finite element model, it didn't consider the effect of muscle activity. In a rear-end impact, it was considered that muscles have no effect up to 150 ms [(24)]. However, some experimental studies suggest that the muscle forces might play an important role for external load [(25), (26)]. Therefore, in the future work, the effect of muscle activity should be included. Next, in the current model, the head restraint was not modeled. Some researches reported the effect of that on the head-neck motion and neck injury in rear-end impact [(16), (27)]. Then, it will make the finite element model included the head restraint in the future study.

2.5 Conclusions

Using three-dimensional human whole body finite element model, the complex cervical behavior such as torsion and bending could be evaluated in the rear-end car collision from an oblique direction. It is useful of evaluating the cervical behavior to investigate the mechanism of whiplash injuries considering the various mechanical parameters.

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CHAPTER 3

THE EFFECT OF IMPACT DIRECTIONS ON THE CERVICAL SPINE IN POSTERIOR-OBLIQUE IMPACTS

3.1 Introduction

In Japan, about 0.6 million people suffer neck injuries in car accidents every year [(1)]. It is well known that neck injuries in rear-end car collisions are the most common injuries resulting from these accidents as "Whiplash Injury". Furthermore, whiplash injuries occurred frequently at speeds lower than 5 km/h [(2)] The injury symptoms include neck pain, headache, dizziness, tinnitus and others [(3)]. It is often difficult to find the pathological lesion by checking used medical imaging devices such as MRI, CT-scan and X-ray, and thus the occurrence mechanism of whiplash injuries is as yet unknown.

Various mechanisms of whiplash injury have been hypothesized based on observed cervical behaviors in rear-end impact tests using dummies [(4) ~ (7)], cadavers [(8) ~ (11)] and volunteers [(12) ~ (14)]. The Hybrid-Ⅲ dummy has conventionally been used to explore cervical behaviors. However, as the neck part of Hybrid-Ⅲ is made of aluminum and hard rubber, it is much stiffer than the necks of living human bodies [(4), (5)]. It was developed that the dummy neck models such as TRID-neck [(6)], BioRID [(7)] improved more flexibility than Hybrid-Ⅲ dummy, while they were not anatomically-correct, and connected with hinge joints. Some researchers have investigated cervical behaviors using whole cadavers [(8), (9)] or isolated cervical spine specimens [(10), (11)]. Cadavers cannot support their posture themselves, meaning that cadavers are too flexible compared to living bodies. There are also studies using volunteers to reproduce occupants neck responses [(12)~(14)], however, volunteers knew the experimental protocol at the beginning of the test and could thus contract their neck muscles in anticipation of the impact, while real occupants are usually unprepared for any impact.

We have developed a new biomechanical neck model called K-D neck to reproduce cervical behaviors of occupants unable to anticipate impending impact [(16)]. This neck model features an anatomical shape with properties simulating the human cervical spine. We have evaluated cervical behaviors with the Hybrid-Ⅲ dummy whose neck was replaced by K-D neck. In addition, numerical analyses were also conducted with pseudo-3D finite element head-neck model which took muscle activities into account [(17)]. In our findings, severe shear movement in an anteroposterior direction occurred between the second cervical vertebra (C2) and third cervical vertebra (C3) [(16), (17)]. As this shear movement was very instantaneously, the soft tissues around vertebral body such as arteries and nerve roots was stretched and sheared at high strain rate, and injured. The injury of arteries and nerve roots correlated with the headache, neck pain, dizziness and numbness in a limb [(19)]. We thought that this was one of the most important causes of whiplash injury reported by much of the literature [(10), (14), (19), (20)], and thus new drivers' seat should be developed to prevent shear movements of vertebral bodies [(18)].

In real accidents, collision may occur from any direction and occupants may be facing directions other than forward when they happen, thus, cervical behaviors can be complex and three-dimensional. Therefore, it is important to evaluate cervical behaviors three-dimensionally. In many previous studies, however, only two-dimensional cervical behaviors have been observed in the sagittal plane. We have investigated three-dimensional cervical behaviors using three-dimensional finite element human whole body model-THUMS [(21)]. In the previous chapter, when the cervical behaviors were evaluated after posterior-oblique impact at the angle of 30°, it was found that severe torsion occurred at C2-C3 and the lateral bending of the cervical spine caused an increase in the compressive stresses on the facet cartilage at C6-C7. In this chapter, we conducted FEM analyses that also considered the posterior-oblique impact at the angle of

15° and 45°, and evaluated how this affected the cervical behaviors.

3.2 Finite element model and methods

3.2.1 Finite element method model

To analyze the cervical behaviors resulting from a rear-end impact, a 3-D finite element model representing a vehicle occupant was constructed with THUMS (Total HUman Model for Safety, automobile occupant model Ver1.52 ß-031117, TOYOTA Central R&D Labs., Inc.), as shown in Fig. 3.1. The THUMS used in this study represents the 50th percentile of American male, with a height of 175 cm, and weighting of 77 kg in a seated posture. The model contains approximately 80,000 elements and includes descriptions of all cortical and cancellous bones, cartilages, ligaments, muscles, tendons, skin, and internal organs. Material properties of the THUMS, such as density, Young's modulus, Poisson ratio, stress-strain curve, stiffness and ultimate stress and strain of bone and soft tissues, were based on previous literature [(22), (23)]. Each body part of the THUMS is selectively validated against published human cadaveric test data, THUMS thus has adequate bio-fidelity to simulate human responses in impacts [(24), (25)]. The cervical parts of THUMS are also constructed to simulate adequately the anatomical shape and properties of the human cervical spine. The modeling details of each component of the THUMS neck segment are shown in Table. 1.

3.2.2 Finite element analysis conditions

FEM analyses were conducted for the first 200 ms after impact and it was assumed that the collision occurred from posterior-oblique directions. Impact angles were set to 0°, 15°, 30° and 45° as shown in Fig. 3.2. The velocity curve obtained from our rear-end sled impact test at 8 km/h [(21)] was applied to the seat part of the FEM model as the loading condition. LS-DYNA Ver.970 (Livermore Software Technology Corp., USA) and Hyper-Works Ver.7.0 (Altair Engineering Inc., USA) were used as the FEM solver and pre-post processor.

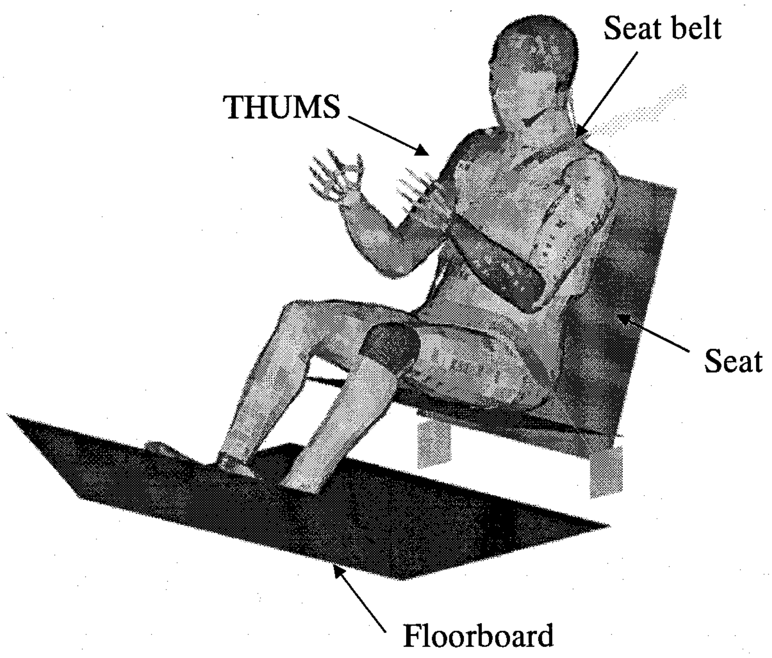


Fig. 3.1 3-D finite element model

Table. 1 Each component of neck segment in the THUMS

Segment	Young's Modulus [MPa]	Poisson ratio	Element type
Cortical bone	5000	0.3	Solid
Cancellous bone	70	0.3	Solid
Nucleus pulposus	0.198	0.499	Solid
Annulus fibrosus	13.3	0.4	Solid
Facet cartilages	12.6	0.4	Solid
LF	15.07	0.22	Shell
LN	30.16	0.22	Shell
ITL	15.08	0.22	Shell
ALL	3.25	0.22	Shell
PLL	3.25	0.22	Shell
Muscles	Hill-type model (passive state)	—	Beam

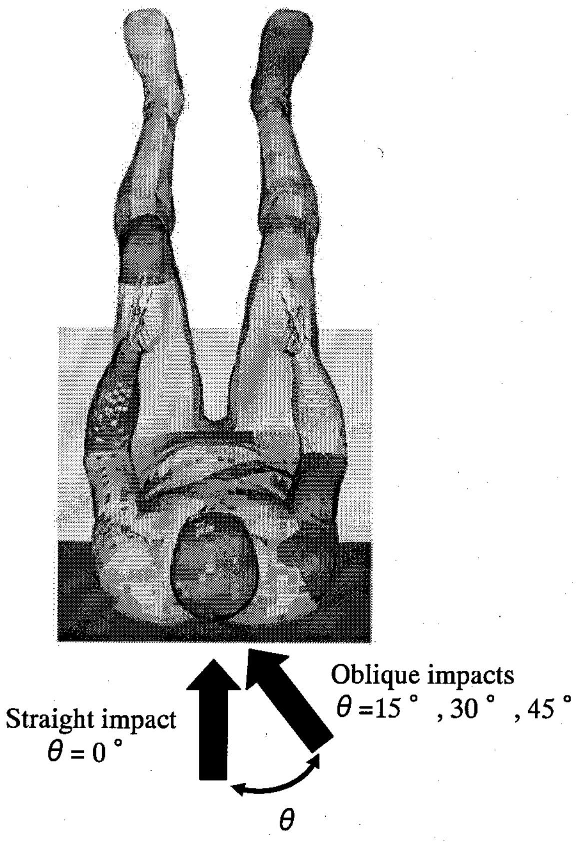


Fig. 3.2 Impact directions

3.2.3 Local coordinate system on the vertebral body

In order to investigate cervical behaviors, the local coordinate system was set on each vertebral body (Fig. 3.3). The local coordinate system was positioned as follows: an origin point was determined on the posterior-center of the upper endplate of the vertebral body: X-axis was directed to the anterior-center of the upper endplate of the vertebral body from the origin point: the Z-axis was directed upward in a direction perpendicular to the X-axis: and lastly, the Y-axis was directed laterally perpendicular to both X and Z-axes. We investigated relative rotational behaviors of adjacent superior vertebra against the local coordinate system. Rotational angles around X-, Y-, and Z-axis of the local coordinate system were measured in the first 200 ms after impact, and sampling frequency was 1kHz. Rotational angles were described as θ_α, θ_β, θ_γ, shown in Fig. 3.4.

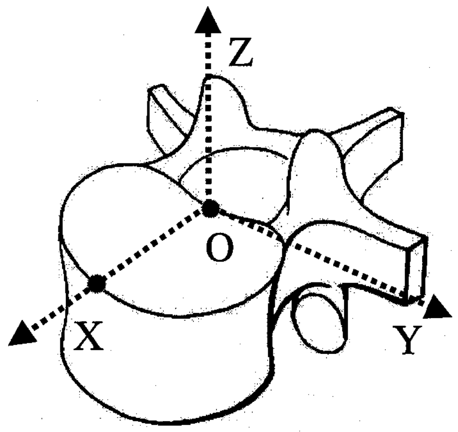
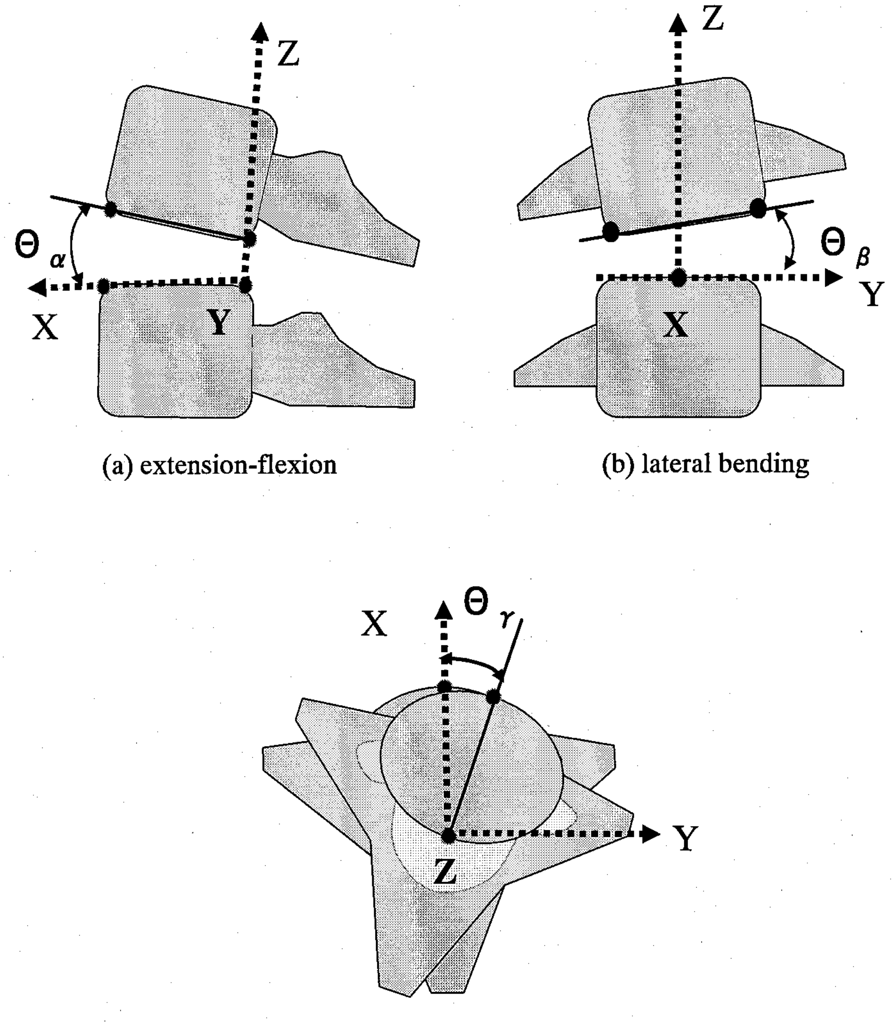


Fig. 3.3 Local coordinate system on the vertebral body



(a) extension-flexion

(b) lateral bending

(c) axial torsion

Fig. 3.4 Rotation angles of vertebral body on the local coordinate system

3.3 RESULTS

3.3.1 Rotational angles of vertebral bodies on the local coordinate system

In the 0° and 15° impacts, The θ_α at C2-C3 decreased from 60 ms to 80 ms (Fig. 3.5(a)), while the θ_α at C6-C7 increased linearity at the same time (Fig. 3.5(b)). This indicates that local flexion occurred in the upper cervical spine (C2-C3) concurrently with local extension in the lower cervical spine (C6-C7). This shows the occurrence of S-curve considered as an important mechanism of whiplash injury. Panjabi et al. reported "S-curve" phenomena in which the cervical spine was formed into an S-shape resulting in local flexion of the upper cervical spine (C1-C3) that cooccurred with the local extension of lower cervical spine (C5-C7) after impact with the rear-end impact test using the cadaveric cervical specimens. It was also reported by some researchers [(26), (27)] that S-curve was caused by the different behavior between the upper and lower cervical spine. Also in our results, as it was shown the different behavior between the upper and lower cervical spine, S-curve may occur in rear-end impacts regardless of the impact angles. The peak of θ_α in the 15° impact was almost the same as that in the 0° impact, while the peak θ_α in the 30° and 45° impacts was smaller than that in the 0° impact. In oblique impacts, as the impact affected also the lateral direction, it was considered that an anteroposterior movement of the head decreased with the increased impact angles.

In posterior-oblique impacts, therefore, the cervical spine was bent in lateral directions (Fig. 3.6). The peak θ_β occurred in the 15° impact and was 0.7° at C2-C3, and 1.2° at C6-C7. The θ_β at C6-C7 was also larger than that between C2 and C3 in the 30° and 45° impacts. This indicates that the lower cervical spine was bent more than the upper cervical spine. In the 45° impact, the θ_β was lower compared with that in the 15° and 30° impacts. With the increased impact angle, as the whole body motion greatened to the lateral direction as shown in Fig. 6, it was considered that the bending moment affected the cervical spine was reduced.

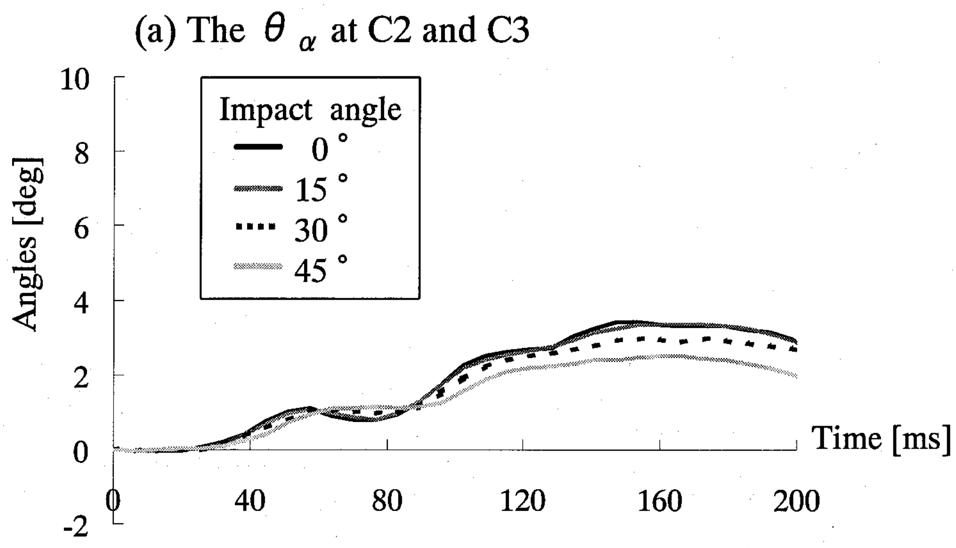


Fig. 3.5(a) Rotation angles of vertebral bodies around Y-axis at C2-C3

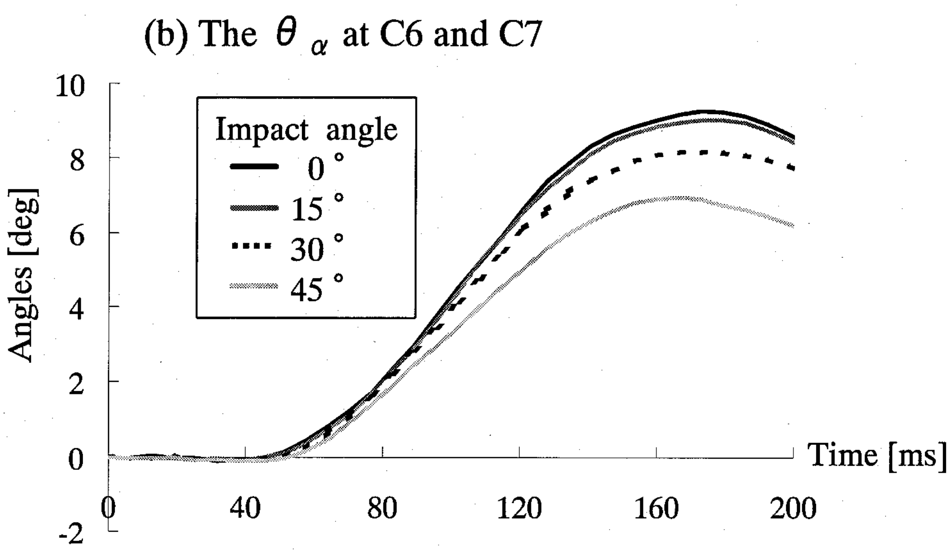


Fig. 3.5(b) Rotation angles of vertebral bodies around Y-axis at C6-C7

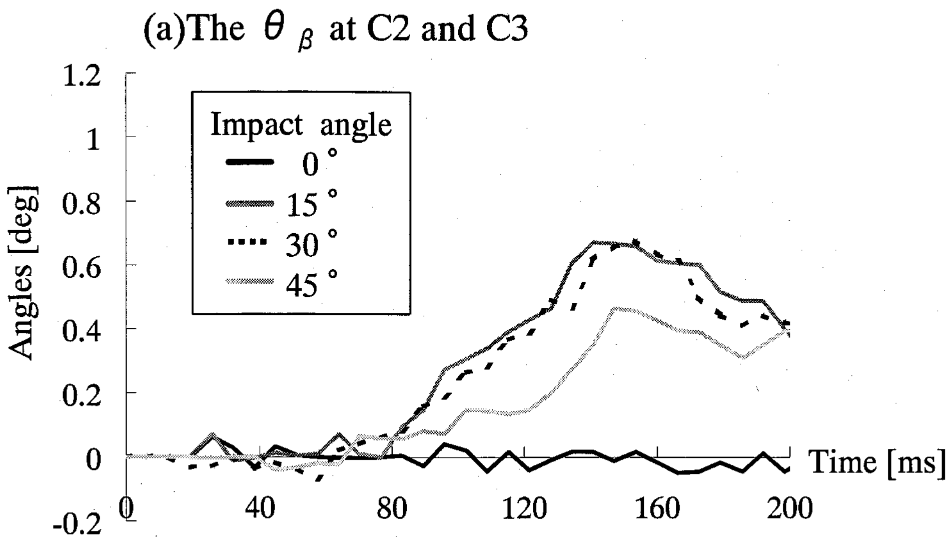


Fig. 3.6(a) Rotation angles of vertebral bodies around X-axis at C2-C3

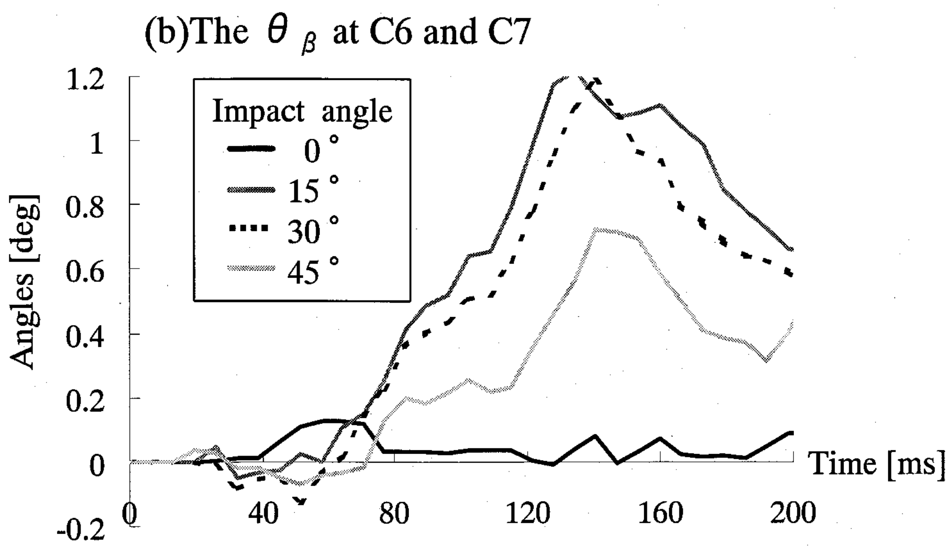


Fig. 3.6(b) Rotation angles of vertebral bodies around X-axis at C6-C7

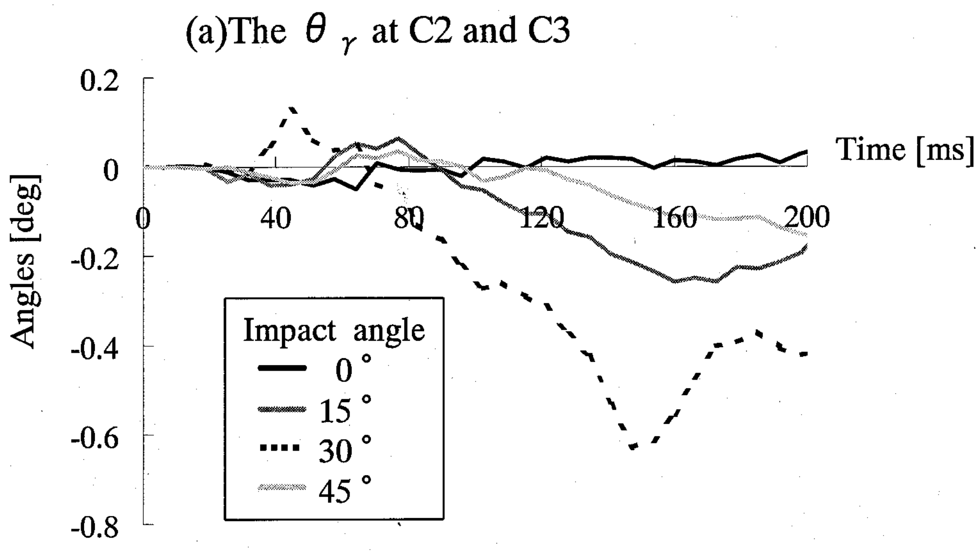


Fig. 3.7(a) Rotation angles of vertebral bodies around Z-axis at C2-C3

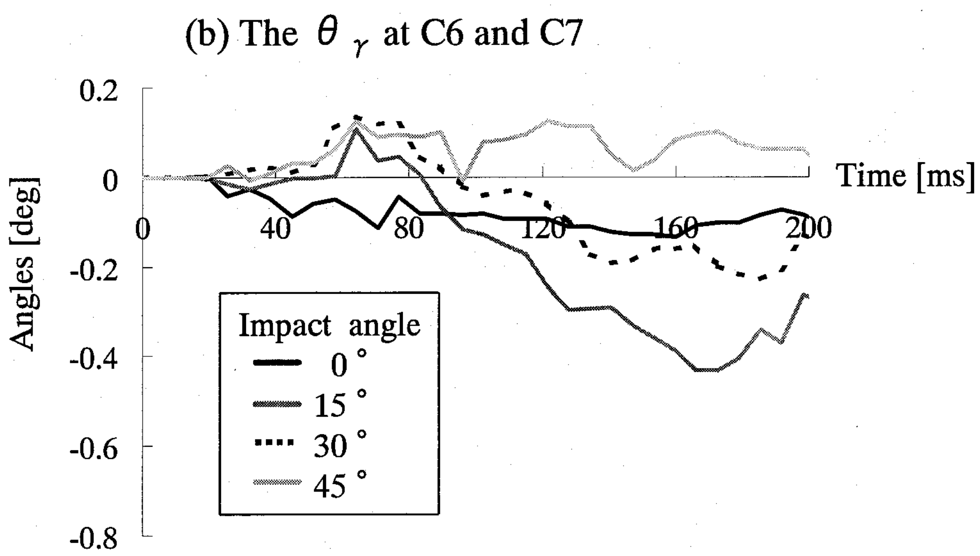


Fig. 3.7(b) Rotation angles of vertebral bodies around Z-axis at C6-C7

Torsional behaviors were also observed in posterior-oblique impacts, as shown in Fig. 3.7. In the 30° impact, the θ_Y at C2-C3 occurred from 30 ms, which was the earliest for any impact. In addition, C2 rotated in left-hand direction until it reached 60 ms, and then changed to right-hand rotation from 60 ms onward. In the 15° and 45° impacts, the θ_Y occurred from 90 ms and 120 ms, respectively. The peak θ_Y at C2-C3 was -0.7° in the 30° impact, which was the largest of any impacts. Interestingly, torsional behavior in the lower cervical spine showed differences compared with the upper cervical spine. The peak θ_Y at C6-C7 occurred in the 15° impact and was -0.45°. The moment affected from the region of C6-C7, and it was larger than at C2-C3 in the 30° impact. However, in the 45° impact, the moment was reduced because the whole body moved laterally, and the θ_Y decreased.

3.3.2 Comparison of the peak stresses in the facet cartilages

In order to evaluate the physical changes of the soft tissues in the cervical spine caused by the rotational motions, we analyzed the compressive, tensile and shear stresses. In particular, it was considered that the whiplash injury involved the injury of facet joints, and some research suggested the injury mechanism on the facet joints. Therefore, we observed the changes of the peak compressive, tensile and shear stresses in the facet cartilages from different impact angles. In the cervical model of the THUMS applied in this study, however, there are some considerable differences compared to an anatomical shape of human. The facet cartilages of the THUMS are formed in the shape close to a rectangular with elastic solid elements, while an anatomical shape is spherical. Especially, the shape of the facet cartilages at C2-C3 focused in this study was spherical. In addition, the joint capsules are not adopted in this model. These are important factors influenced not only stress distributions but also joint kinematics with material properties, and it will need to improve for the more precise stress analyses. Therefore, the stress values were compared relatively.

The peak stresses on the facet cartilages occurred at C6-C7. The compressive stress in the 15° impact increased by 11 % compared to that in the 0° impact (Fig. 3.8(a)). In addition, the shear stresses in the 15° impact increase by 14 % compared to that in the 0° impact (Fig. 3.8(b)). At the same time, in the left-side facet cartilage, the tensile stresses increased in oblique impacts. Most notably, the tensile stress in the 15° impact increased by 34 % as shown in Fig. 3.8(c). In the 45° impact, although the tensile stress increased by 14 %, the compressive and shear stress decreased by 13 % and 17 % compared to that in the 0° impact. Since the rotational angle of the sagittal plane in the 45° impact decreased compared with other impacts, it is thought that the extension of the cervical spine was moderated and thus the compressive stresses were decreased.

As shown in the torsional behavior, the largest torsion angle occurred at C2-C3 in the 30° impact, which also affected the shear stress in the facet cartilage at C2-C3. In particular, the peak shear stress in the 30° impact increased by 27 % as shown in Fig.3.8(d).

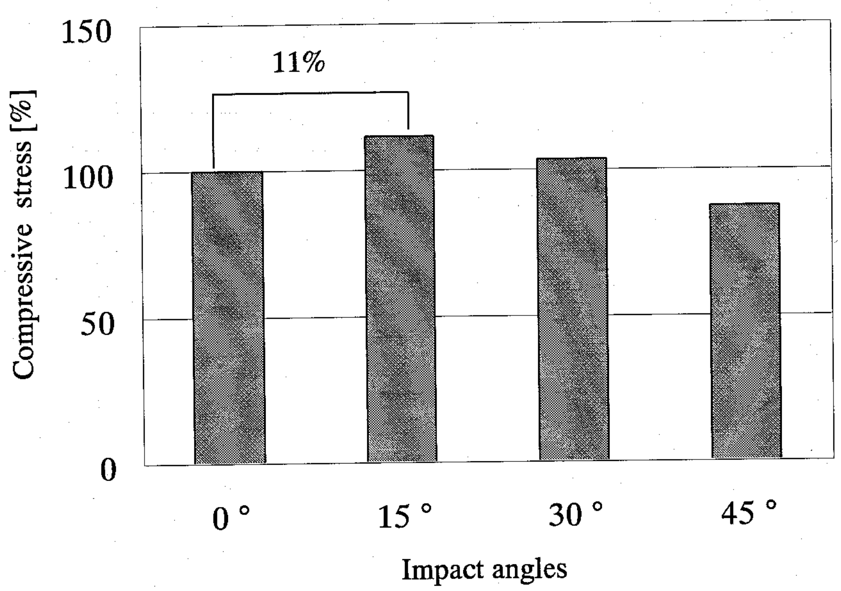


Fig. 3.8(a) Comparison of the peak compressive stresses on the right-side facet cartilages at C6-C7

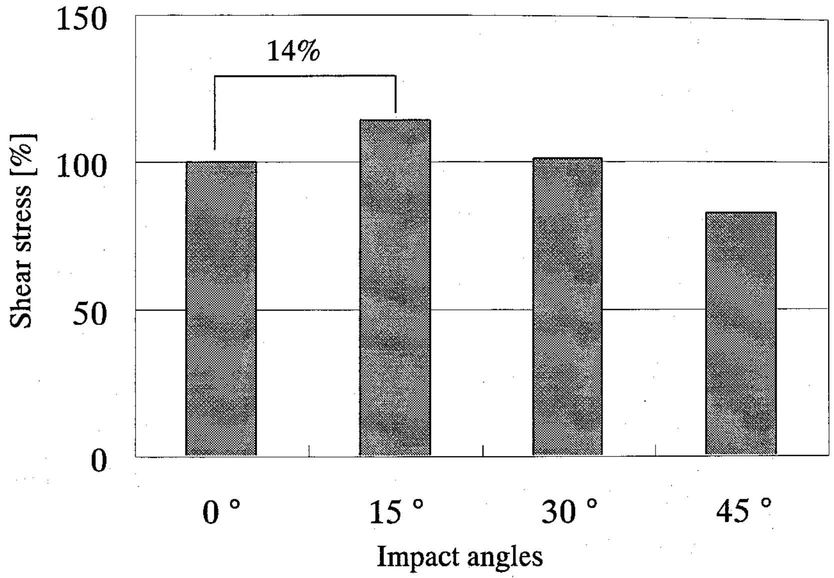


Fig. 3.8(b) Comparison of the peak shear stresses on the right-side facet cartilages at C6-C7

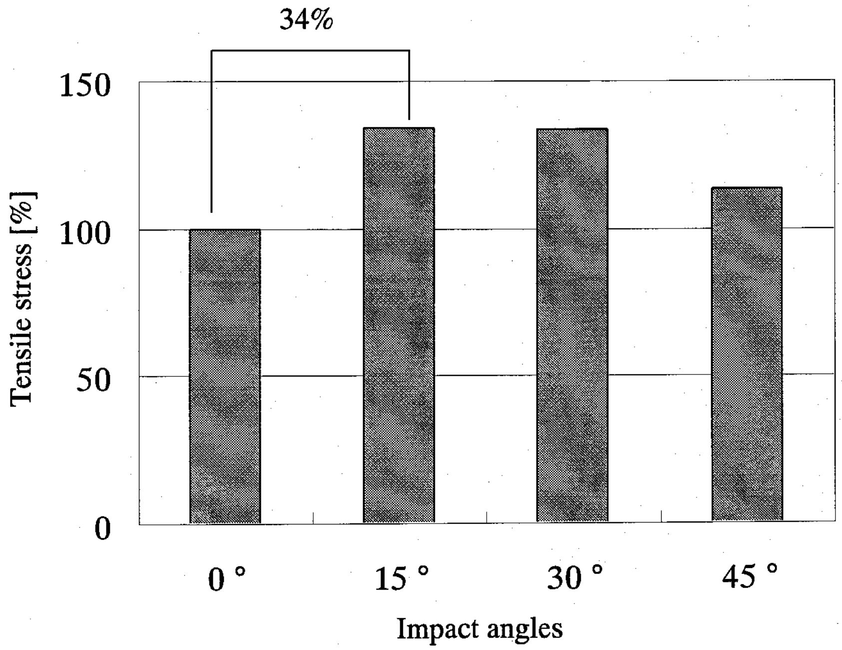


Fig. 3.8(c) Comparison of the peak tensile stresses on the left-side facet cartilages at C6-C7

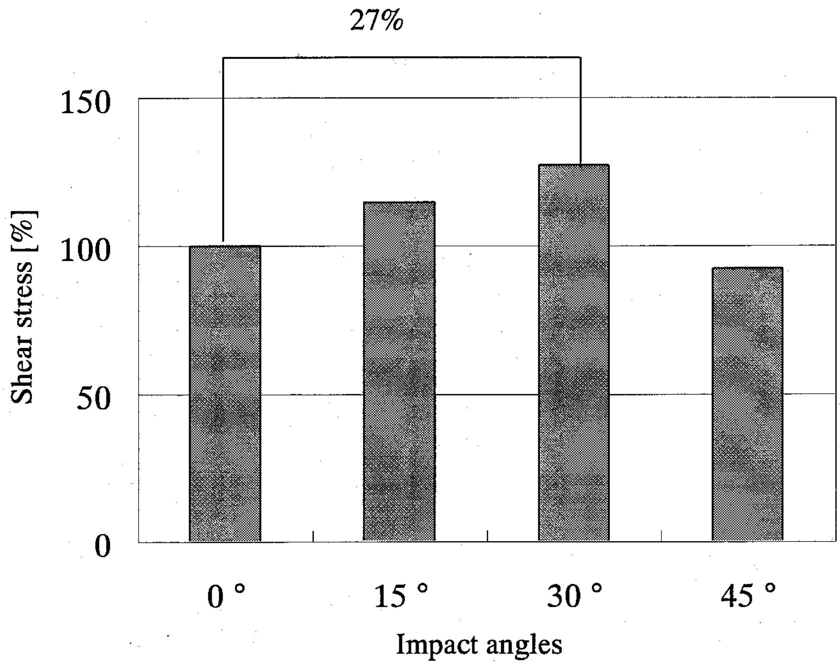


Fig. 3.8(a) Comparison of the peak shear stresses on the right-side facet cartilages at C2-C3

3.4 DISCUSSION

Over the past decades, in many biomechanical studies, cervical behaviors have been evaluated to reveal the mechanism of whiplash injuries using experimental and numerical analyses [(28) ~ (30)] solely on the two-dimensional sagittal plane of the cervical spine. Williams and Belytshko reported the head response using a three-dimensional mathematical model with rigid bodies and beam elements [(28)]. Linder reported on the forces acting on the cervical spine using a newly-developed rigid-body model of a 50th-percentile human [(30)]. However, since these studies were conducted with the cervical model simplified elliptically, particular behaviors of the cervical spine were not evaluated. Furthermore, in real accidents, car collisions may occur from any direction and occupants may be facing several directions other than forward: thus, cervical behaviors during impact can be complex and three-dimensional. We have therefore evaluated the cervical behaviors three-dimensionally as well as the stress distributions, using THUMS, and considering the posterior-oblique impact [21]. In this study, we evaluated the rotational motions at C2-C3 and C6-C7 in posterior-oblique impacts with angles of 15°, 30° and 45° compared with the impact at back. Consequently, the rotation of the vertebral bodies were different depending on impact angles.

We noted in our findings that the cervical spine was bent and twisted in oblique impacts. These caused the head to bend in a lateral direction as well as to extend backward. It was interesting to observe that the bending behavior trended similarly between C2-C3 and C6-C7, while the torsional behaviors were different between C2-C3 and C6-C7. The bending angle at C6-C7 was larger than that at C2-C3 in oblique impacts, and the peak angle at C6-C7 occurred in the 15° impact. In addition, the peak torsion angle at C2-C3 occurred in the 30° impact, while that at C6-C7 occurred in the 15° impact. Since there was greater torque on the head in the 30° impact than that in the 15° impact, it is thought that the larger torque affected the upper cervical spine more severely because it moves more easily than the lower cervical spine.

The bending and torsion caused increase of the peak stress in the facet cartilages during the oblique impacts. The compressive and shear stresses increased in the right-side facet cartilage at C6-C7, and the tensile stresses increased in the left-side facet cartilage. In this analyses, although the peak stresses were not over the injury threshold (25 MPa to the compression and 5 MPa to the shear [31]), the surfaces of the facet cartilage might develop cracks at an extremely low shear stress [31]. As mentioned in the result part, since the shape of facet cartilages in the THUMS applied in this study was different compared to an anatomy of human, it need to evaluate about the accurate damage of the facet joint suffered by the increase of stresses. However, it was sufficiently considered that the facet joints easily suffered an injury caused by the increase of stresses, and resulted in the important cause of whiplash disorders because it was reported that the injury of facet joints echoes the symptoms such as neck pain [32), (33)], and some researchers suggest the injury mechanism of the facet joint: Luan et al. reported that the three-stage motion of the cervical spine causes the injury of facet capsular ligaments [8].

Kaneoka et al. suggested "pinching mechanism" that inflammation might occur in the synovial fold pinched between facet joints [14]. Throughout this study, we assume that the facet joints are subjected to injuries by bending and torsional motion, depending on impact directions.

The torsion and bending behavior occurred very instantaneously. Therefore, the soft tissues around vertebral bodies sheared at high strain rate. It is well known that soft tissues are sensitive to shear forces at high strain rate, and so the risk of injury around C2-C3 in which the torsion angle was large may severely increase. Injury around C2-C3 is related to symptoms such as headache and neck pain, as reported in clinical studies [33), (34]. In addition, there are many symptoms of whiplash related to the injury of arteries, an example being Barré-Liéou syndrome due to blood stream obstruction [34]. However, in our FEM model, since the vertebral arteries and nerve roots were not modeled, we could not evaluate these physical changes In future studies, we should therefore evaluate the injuries caused by the torsional behavior focused the arteries and nerve roots. In addition, there are some limitations in this study. First, it was insufficient to verify the bio-fidelity of the vertebral behaviors of the FEM model. In this paper, as the verification for the bio-fidelity of FEM model, only the flexion-extension responses at C2-C3 and C6-C7 were shown. Therefore, it will need to evaluate the bio-fidelity of all cervical vertebral behaviors three-dimensionally. Next, we did not consider the effect of muscle activity because it seems that muscles have no effect up to 150 ms during rear-end collision, as reported in some papers [35), (36]. However, some experimental studies suggest that muscle might play an important role [37), (38]. We therefore need to consider carefully the effect of muscle activities in real collisions in future studies.

3.5 CONCLUSIONS

It could be found the complex cervical behavior combined the bending and torsion in posterior-oblique impacts with 3-D FEM analyses. Complex behaviors such a bending and torsion occurred by the oblique impacts, and the soft tissues were subjected to injuries by the increase in stress. In real accidents, as impacts may occur from several directions, it is conceivable that the cervical spine will be twisted and bent as shown in this study. We feel this complex behavior is an important cause complicating the mechanism of whiplash injuries. Therefore, in our future study, it will be important to find a method to reduce torsion and bending of the cervical spine.

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CHAPTER 4

A NEW NECK INJURY CRITERION BASED ON THE EFFECT OF ROTATION BETWEEN THE HEAD AND CHEST

4.1 Introduction

A whiplash injury is well known as a common cervical injury suffered by many people who have been in car accidents, and results in expensive insurance payments, e.g., US$ 3.0 billion dollar in Japan, in the year 2000 [1]. Although clinically identifying the injured parts is difficult, clinical symptoms include headache, neck pain, dizziness, tinnitus and others. In order to design a prevention system for whiplash injuries, a criterion for validating the injury severity must be developed. The Neck Injury Criterion (NIC) was developed based on the hypothesis that neural injuries were caused by sudden changes in the pressure gradient in cerebral fluid pressure [2]. The NIC showed the correlation with increased the impact severity, while a definitive correlation with the severity of soft tissue injuries has not been established yet. Alternatively, the Intervertebral Neck Injury Criterion (IV-NIC) was proposed to validate the soft tissues injury in consideration of a hyperextension injury due to intervertebral motion beyond the physiological limit [3]. The IV-NIC showed a good correlation with the injury severity depending on the intervertebral level, and showed possibility as a effective criterion to predict the soft tissues injuries observed clinically [4]. However, both the NIC and IV-NIC were evaluated with two-dimensional components in the anteroposterior, or flexion-extension motion.

In many studies, the cervical behaviors in a rear-end impact have been investigated to reveal the whiplash mechanism [(5)~(15)]. It was assumed that the cause of whiplash injury was soft tissues injury due to the complex cervical behavior during impact, and the injury severity depended on the degree of impact. However, only two-dimensional cervical behaviors were observed in the sagittal section under consideration of the back impact alone. In real accidents, however, collisions may occur from any direction and occupants may be facing directions other than forward when the accidents happen: thus, cervical behaviors can be more complex and three-dimensional. We have investigated three-dimensional cervical behaviors using the finite element human whole body model. In our previous studies, when the impact occurred in oblique directions, severe torsion and lateral bending motion could be found in the cervical behavior, which increased the stresses on the soft tissues [(16), (17)]. This indicates that the injury severity can increase easily due to three-dimensional cervical behaviors. Therefore, the conventional criterion will need to be modified: alternatively, a new injury criterion that considers the three-dimensional behaviors should be developed.

In this study, the Rotational Neck Injury Criterion (RNIC) is proposed based on the hypothesis that the rotation between the head and chest influences an increase in the soft tissue injury severity. In addition, we propose the Total Neck Injury Criterion (TNIC), which is a combination of the NIC and the RNIC. The aim of this study is to evaluate the availability of the TNIC and RNIC and consider the effect of rotation on neck injuries by the numerical simulation with a three-dimensional finite element human model.

4.2 Finite element model and Methods

To evaluate the effect of head-neck motions on a neck injury in posterior-oblique impacts, three-dimensional FEM human body model-THUMS (Total HUman Model for Safety, automobile occupant model Ver1.52 ß-031117, TOYOTA Central R&D Labs., Inc.) was used (Fig. 4.1). The THUMS used in this study represents the 50th percentile of the American male, with a height of 175 cm and weighing 77 kg in a seated posture. The model contains approximately 80,000 elements and includes descriptions of all cortical and cancellous bones, cartilage, ligaments, muscles, tendons, skin, and internal organs. The material properties of the THUMS were based on previous literature [18], [19]. Each body part of the THUMS is selectively validated by bio-fidelity to simulate human responses in impacts by comparison to the published human cadaveric test data [20], [21]. The head-neck segments of the THUMS used in this study are also constructed to simulate the human cervical spine. The modeling details of each component of the THUMS neck segment are shown in Table. 1.

FEM analyses were conducted for the first 120 ms after impact and it was assumed that the collision occurred from posterior-oblique directions. Impact angles were set to 0°, 15°, 30° and 45° as shown in Fig. 4.2. The velocity curve obtained from our rear-end sled impact test at 8 km/h [1] was applied to the seat part of the FEM model as the loading condition. LS-DYNA Ver.970 (Livermore Software Technology Corp., USA) and Hyper-Works Ver.7.0 (Altair Engineering Inc., USA) were used as the FEM solver and pre-post processor.

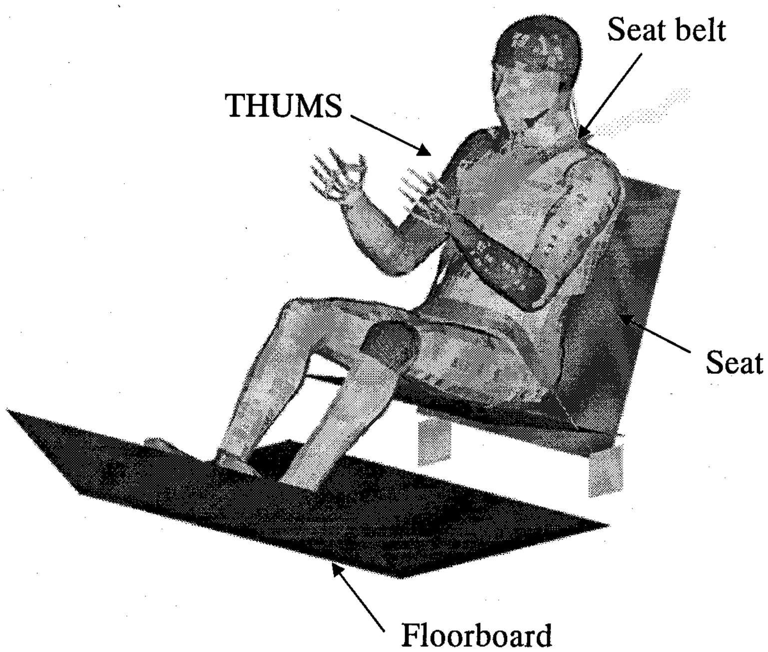


Fig. 4.1 3-D finite element model

Table. 1 Each component of neck segment in the THUMS

Segment	Young's Modulus [MPa]	Poisson ratio	Element type
Cortical bone	5000	0.3	Solid
Cancellous bone	70	0.3	Solid
Nucleus pulposus	0.198	0.499	Solid
Annulus fibrosus	13.3	0.4	Solid
Facet cartilages	12.6	0.4	Solid
LF	15.07	0.22	Shell
LN	30.16	0.22	Shell
ITL	15.08	0.22	Shell
ALL	3.25	0.22	Shell
PLL	3.25	0.22	Shell
Muscles	Hill-type model (passive state)	—	Beam

FEM analyses were conducted for the first 120 ms after impact and it was assumed that the collision occurred from posterior-oblique directions. Impact angles were set to 0, 15, 30 and 45 ° as shown in Fig. 4.2. The velocity curve obtained from our rear-end sled impact test at 8 km/h (1) was applied to the seat part of the FEM model as the loading condition. LS-DYNA Ver.970 (Livermore Software Technology Corp., USA) and Hyper-Works Ver.7.0 (Altair Engineering Inc., USA) were used as the FEM solver and pre-post processor.

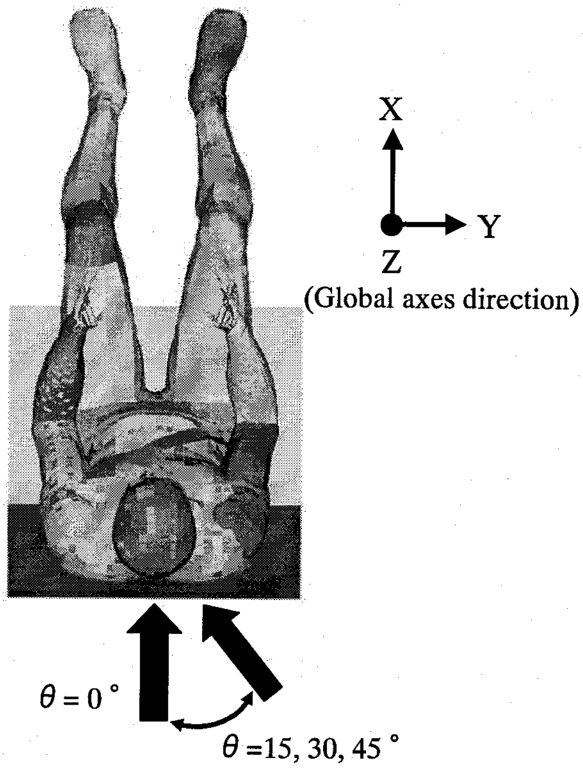


Fig. 4.2 Impact directions

4.3 Injury criterions

4.3.1 Previews by previous criterions (NIC and NIC_m)

The NIC (Neck Injury Criterion) was proposed by Boström et al. to predict neck injuries in low- speed rear-end collisions based on the hypothesis that a neck injury was caused by a change of pressure gradient in the cerebrospinal fluid (CSF) [(2)], and calculated as follows:

NIC = NIC($t = t_{max}$) [m^2/s^2] t_{max}: moment of time at S-shape of the cervical spine

$$NIC(t) = 0.2 \cdot a_{rel}(t) + (v_{rel}(t))^2$$

$a_{rel}(t) = a_x^{T1}(t) - a_x^{Head}(t)$: relative acceleration between T1 and the head

$v_{rel}(t) = \int a_{rel}(t)dt$: relative velocity between T1 and the head

The NIC should not exceed $15 m^2/s^2$ if injuries with long-term consequences are to be avoided [(2)]. The estimation was based on the scaling of the pig anatomy to that of the human and comparing with results from volunteers, which were further supported by Boström et al. and

Eichberger et al..

Next, NIC_m was applied with some modification by Eichberger et al. to the NIC as follows [22].

NIC_m $[m^2/s^2]$: 3 ms Maximum of $NIC_m(t)$ within the first 120 ms

$$NIC_m(t) = 0.2 \cdot a_{res,rel}(t) + (v_{res,rel}(t))^2$$

$a_{res,rel}(t) = a_{T1,res}(t) - a_{Head,res}(t)$: relative resultant acceleration between T1 and the head

$v_{res,rel}(t) = \int a_{res,rel}(t)dt$: relative resultant velocity between T1 and the head

NIC_m was calculated using the resultant acceleration and velocity instead of the longitudinal component Eichberger et al. neglected the lateral (Y-axis) component because the lateral movement of the head was not occurred in their experiment, but we calculated NIC_m with all components including the lateral and vertical directions, in this study.

4.3.2 A new criterion based on the rotation between the head and chest Rotational NIC (RNIC) and Total-NIC (TNIC)

It was assumed that the cause of whiplash injury was not hyperextension of the cervical spine caused by head rotation, but the shear movement between the cervical vertebra caused by the translation of the head relative to the chest. However, the head-neck motions were investigated two-dimensionally only in the sagittal section, and thus the injury severity increased with only the impact severity.

In our previous study, however, it was reported that lateral bending and axial torsion affected the cervical spine, and that caused the stress to increase on the soft tissues, when the collision occurred from a posterior-oblique direction [16], [17]. This indicates that the injury severity was influenced even by the impact direction. From the point of view in our study, the bending and torsion of the cervical spine occurred by the torque caused by the influence of the mass of head and oblique impact. Thus, in order to evaluate a neck injury in a posterior-oblique impact, the three-dimensional movements combining the rotational movements of the cervical spine aside from the translation in anteroposterior direction needed to be considered. It is difficult to take into account all of the cervical behaviors because the cervical behaviors were changed in the vertebral level depending on the impact direction, and thus the relative behaviors between the head and chest were considered an influencing factor on the neck injury. Based on this hypothesis, we propose the Rotational Neck Injury Criterion (RNIC) as a new criterion in consideration of the effect of the rotational movements between the head and chest on a soft tissues injury. The RNIC is calculated by applying the calculation formula of the NIC as follows:

RNIC [$m^2 \cdot rad^2/s^2$] : maximum of RNIC(t) within the first 120 ms after impact

$$RNIC(t) = 0.2^2 \cdot \{\Delta\theta_{rel}(t) \cdot \dot{\omega}_{rel}(t) + (\omega_{rel}(t))^2\}$$

$\dot{\omega}_{rel}(t) = \dot{\omega}_{T1}(t) - \dot{\omega}_{Head}(t)$: resultant angular acceleration of the head relative to T1

$\omega_{rel}(t) = \int \dot{\omega}_{rel}(t)dt$: resultant angular velocity of the head relative to T1

$\Delta\theta_{rel}(t) = \theta_{T1}(t) - \theta_{Head}(t)$: resultant rotation angle of the head relative to T1

The constant 0.2 in the first formula represents the length of the human neck in meters and is used to conform the unit of the RNIC value to the NIC value. Furthermore, in order to comprehensively evaluate the effect of the three-dimensional translation and rotation of the head-neck on the neck injury, we suggest the Total-NIC (TNIC) combined RNIC with NIC as follows:

$$TNIC = NIC_m + RNIC$$

4.4 Results

4.4.1 The accelerations of the head relative to T1

To evaluate the correlation between the head movements and the injury criterions, the translational and rotational accelerations of the head relative to the first thoracic vertebra (T1) on the global axes were investigated. The accelerations were obtained from the center of gravity of the head and T1 in the FEM model.

4.4.1.1 The translational accelerations of the head relative to T1

Figure 4.3 shows the time history of the translational accelerations. In the X- and Z-axis direction, the tendency of the time history curves were almost the same with respect to the impact angles. In the X-axis direction, the negative accelerations occurred, this indicated that the head moved in the posterior direction relative to T1. The peak acceleration decreased with the increased impact angles, and that was -42 m/s^2 at 50 ms in the 0° impact and -31 m/s^2 at 60 ms in the 45° impact. In the Z-axis direction, the acceleration changed from negative to positive between 10 ms and 60 ms, and the negative acceleration again occurred from 60 ms to 100 ms. This demonstrates that the head-neck motion in a rear-end impact is not a simple translation but a complex behavior such as "whiplash". The peak acceleration in the Z-axis decreased with the increased impact angles, the same as that in the X-axis. These results indicate that the impact affecting the head-neck in the anteroposterior and up-down direction was attenuated in the posterior-oblique impacts. When the acceleration occurred in the Y-axis direction in the posterior-oblique impacts, this means that the impact in the lateral direction affected the head-neck, and that might cause a bending in the cervical spine in the lateral direction.

4.4.1.2 The angular accelerations of the head relative to T1

Figure 4.4 shows the results of angular accelerations. Around the X-axis, which was the rotation in the lateral direction, accelerations were not generated in the 0° impact, while the head rotation occurred in the 15°, 30° and 45° impacts. The time history showed a tendency of following the translational acceleration in the Y-axis, which was caused by the impact in the lateral direction. Around the Y-axis, which was the flexion-extension in the sagittal section, the time history curves showed the same tendency in all impacts. Interestingly, the peak accelerations were not proportional to the impact angles differently from the translational accelerations. Around the Z-axis, which was the axial torsion, the time history curves showed a different tendency depending on the time in each impact angle.

It could be found that the acceleration changed from negative to positive in the 30° and 45° impact between 0 and 30 ms, but that was almost not changed until 40 ms in the 15° impact. Interestingly, although the peak acceleration in the 15° impact was higher than that in the 30° impact at 80 ms, the peak acceleration in the 30° impact was higher than that in the 15° impact at 100 ms.

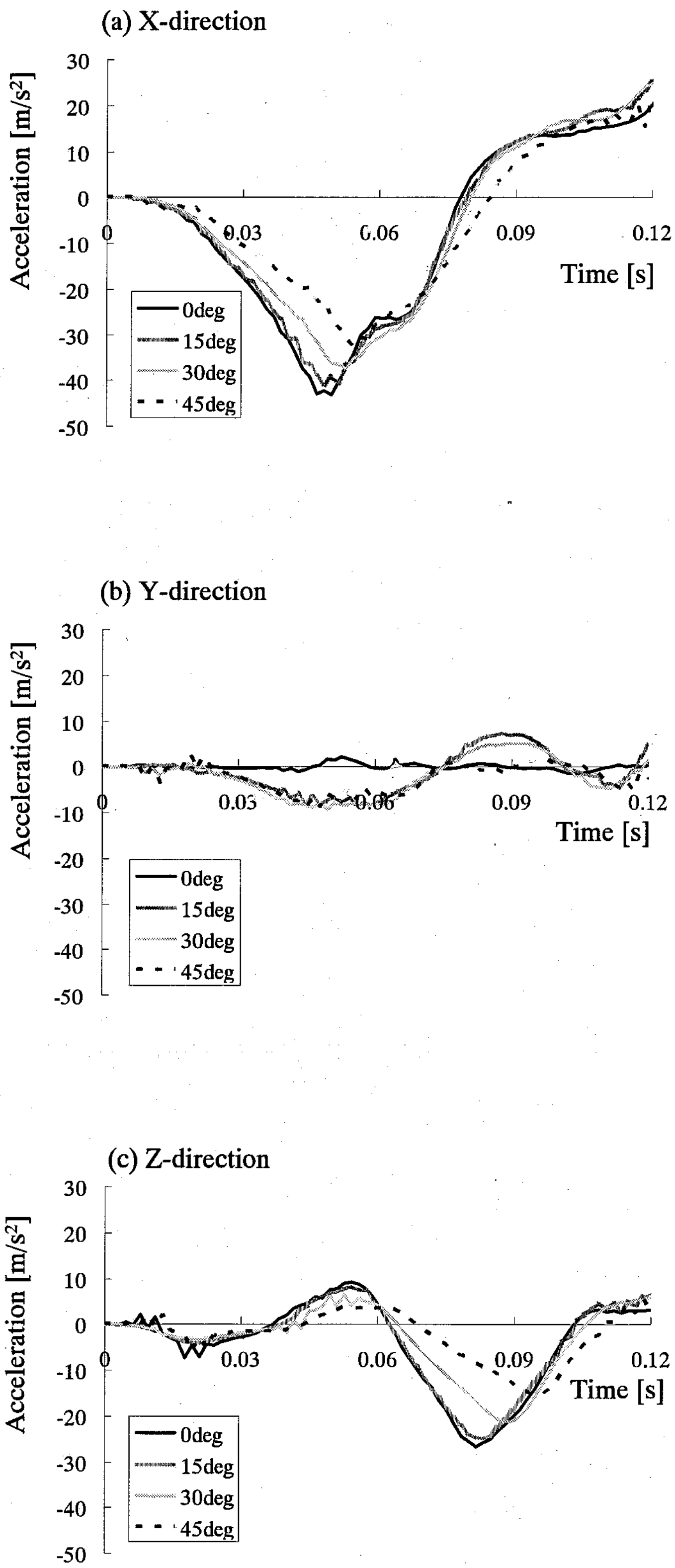


Fig. 4.3 The translational accelerations of the head relative to T1 on the
(a) X-axis (horizontal direction) (b) Y-axis (lateral direction)
(c) Z-axis (vertical direction)

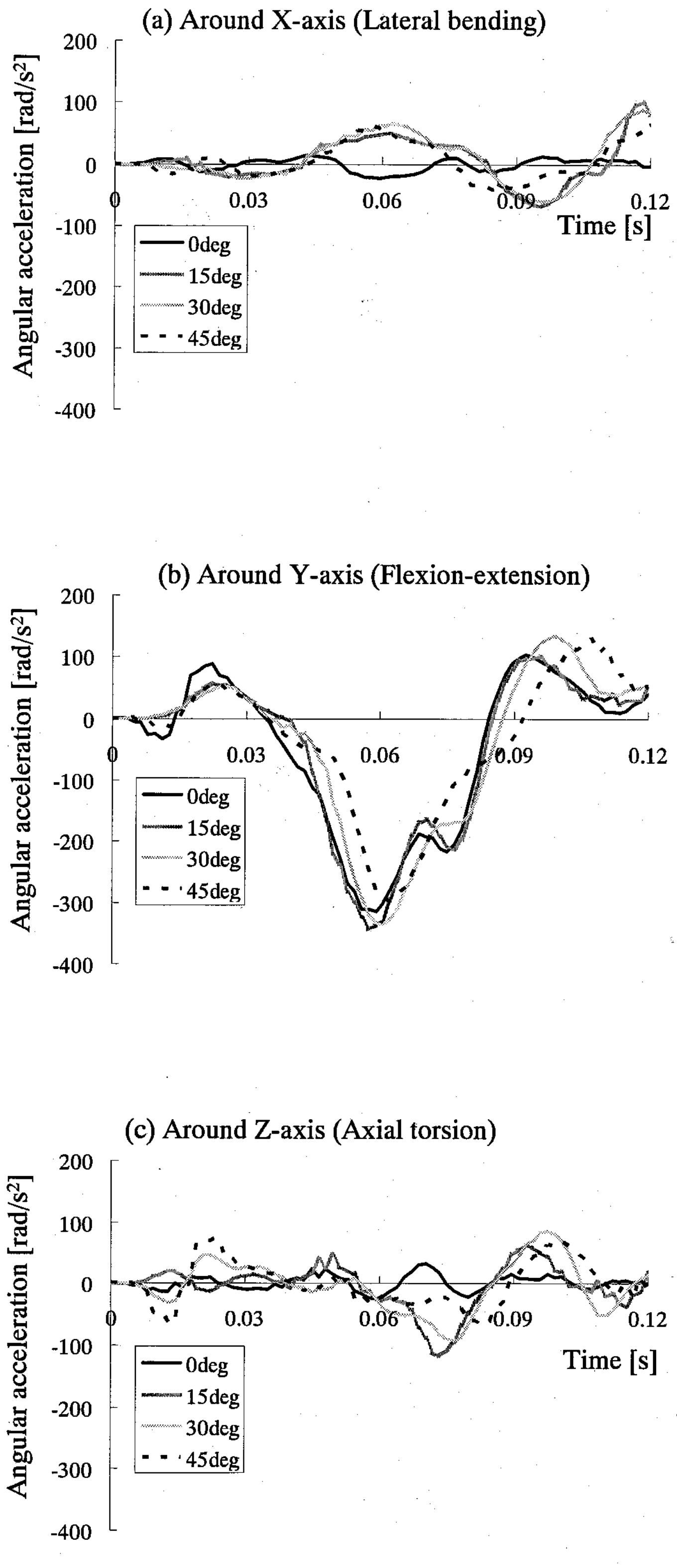


Fig. 4.4 The angular accelerations of the head relative to T1 around the
(a) X-axis (flexion-extension) (b) Y-axis (lateral)
(c) Z-axis (vertical direction)

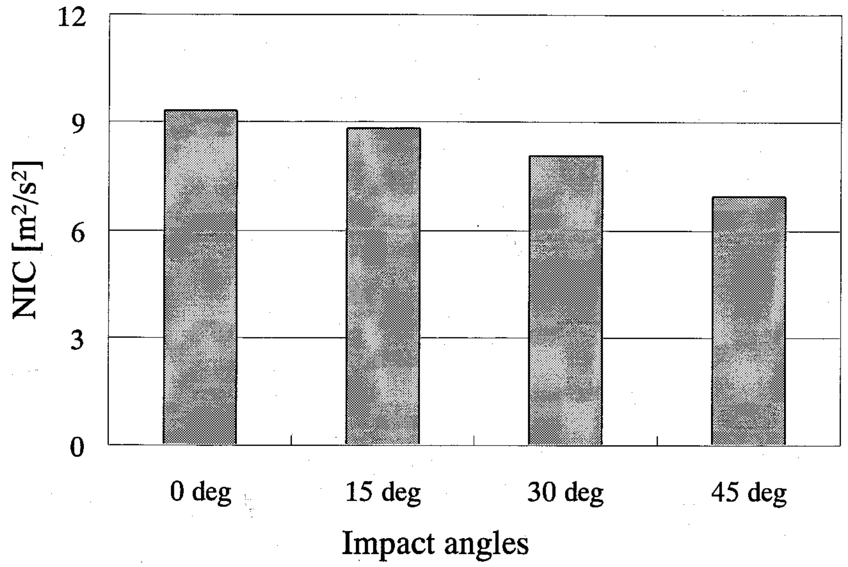
4.4.2 The results of injury criterions

4.4.2.1 NIC and NIC_m

Figure 4.5 shows the results of the NIC. The NIC decreased with the increased impact angles from 9.3 m^2/s^2 in the 0° impact to 6.9 m^2/s^2 in the 45° impact. As shown in Fig. 4.3, the translational acceleration decreased with the increased impact angles, and thus resulted in the decrease of the NIC. In addition, the time when the NIC peaked was almost the same with the translational acceleration in the X-axis. This indicates that the NIC was well related to the acceleration severity. However, in this study, NIC values in any impact angles were not over 15 m^2/s^2 applied as a threshold.

The NIC_m showed a different tendency compared to the NIC, as shown in Fig. 4.6. NIC_m in the 15° impact was 6.5 m^2/s^2, and was highest in all impacts, but that was lower than the NIC. Also in other impacts, NIC_m was lower than the NIC. The times when the NIC_m peaked in all the impacts were after 60 ms, and did not agree with those of any translational acceleration. NIC_m was calculated with the resultant acceleration and velocity of the head relative to T1. Then, NIC_m might be delayed by the influence of the vertical and lateral component of the acceleration and velocity. While the NIC_m in the 15° and 30° impacts was higher than that in the 0° impact, the injury risk increased inversely with the results of the NIC.

(a) Comparison of peak NIC values



(b) Time history curve of NIC

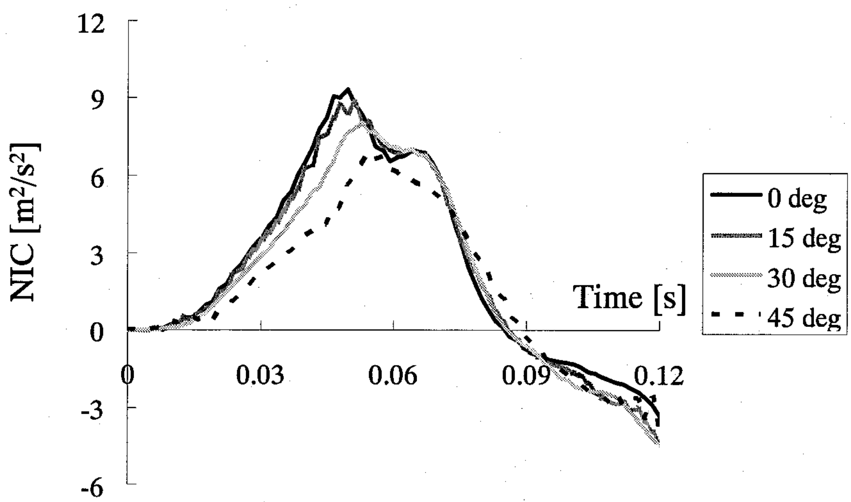
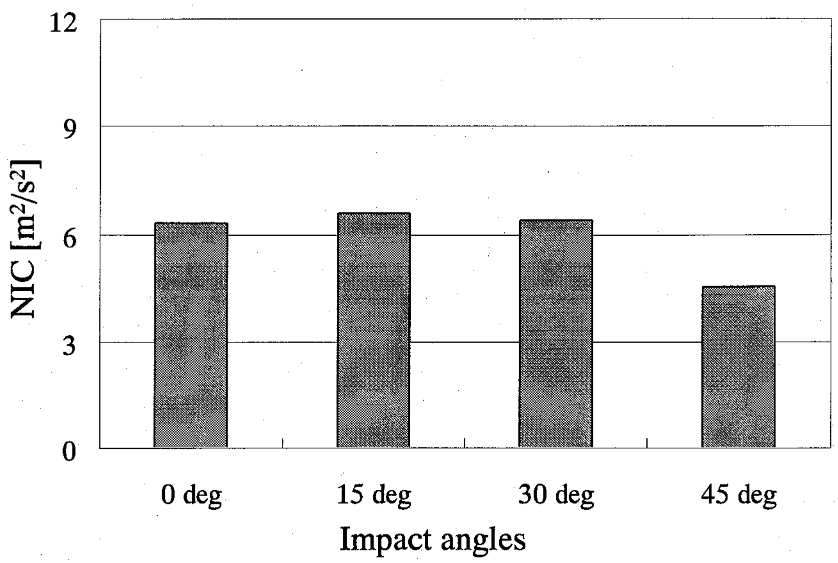


Fig. 4.5 The results of NIC
(a) Comparison of the peak NIC with respect to the impact angles
(b) The time history curves of the NIC

(a) Comparison of peak NIC values



(b) Time history curve of NIC_m

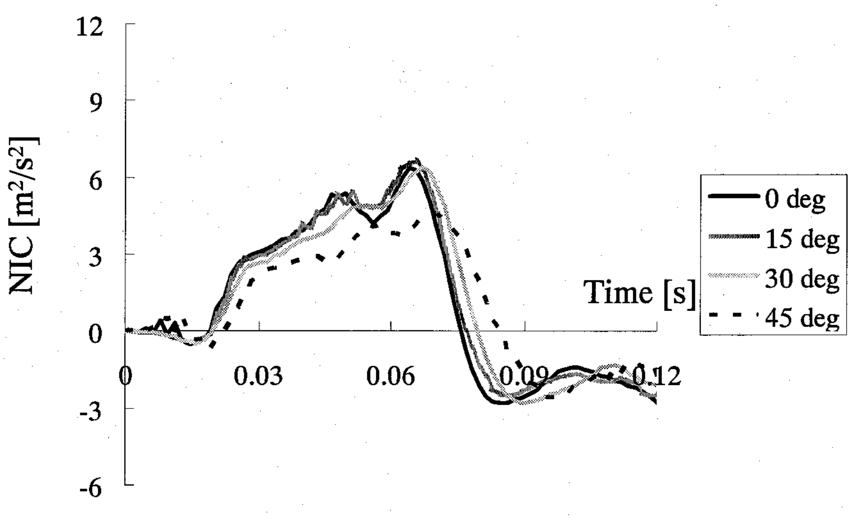


Fig. 4.6 The results of NIC_m
(c) Comparison of the peak NIC_m with respect to the impact angles
(d) The time history curves of the NIC_m

4.4.2.2 RNIC and TNIC

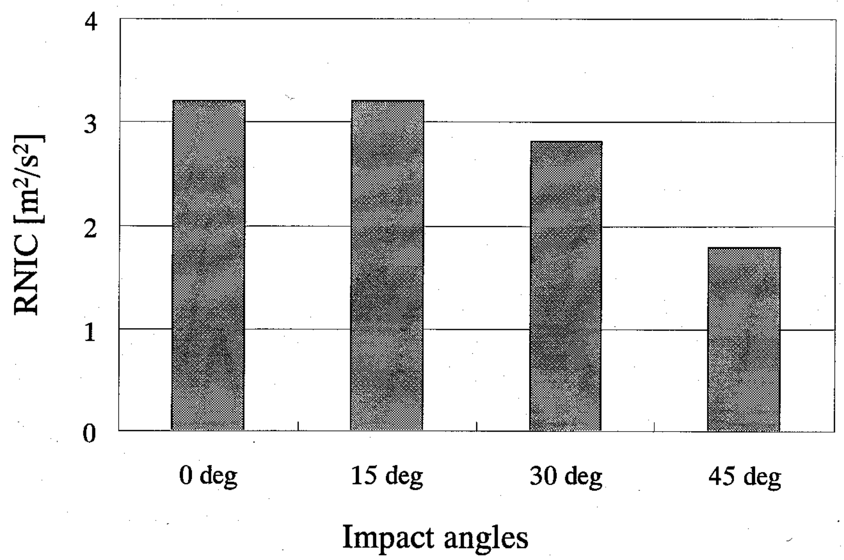
Figure 4.7 shows the results of the RNIC. The RNIC in the 0° and 15° impacts was 3.2 $m^2\ rad^2/s^2$, and indicated the same estimate. In the 30° and 45° impacts, the RNIC decreased to 2.8 and 1.9 $m^2\ rad^2/s^2$, respectively. The time when the RNIC peaked was over 80 ms in all impacts, and later than that of the NIC and NIC_m. It was considered that the RNIC could indicate the injury risk in a different way from the NIC and NIC_m.

The TNIC indicated highest value in the 15° impact, and was 9.8 m^2/s^2 (Fig. 8). The TNIC in the 0° and 30° impact was 9.5 and 9.2 m^2/s^2 respectively, this could indicate the injury risk reported by Panjabi et al.(4), while that in the 45° impact was 6.3 m^2/s^2, lower than the NIC and NIC_m.

4.4.3 Comparison of peak stresses with respect to impact angles

As the results of evaluating soft tissues injury were based on the stress analyses, the peak tensile stresses were over the injury threshold, which was 3 MPa with respect to tension on the intervertebral disk (20), at C5-C6 in the 0°, 15° and 30° impacts (Fig. 4.9). Panjabi et al. reported soft tissues injury due to hyperextension at the C5-C6 level (8), and suggested that the peak NIC as the soft-tissue injury threshold was 8.7 m^2/s^2 (4). Also in our results, the NIC was over 8.7 m^2/s^2 in the 0° and 15° impacts, but not over in the 30° impact. In addition, NIC_m in all the impacts was not over the threshold.

(a) Comparison of peak RNIC values



(b) Time history curve of RNIC

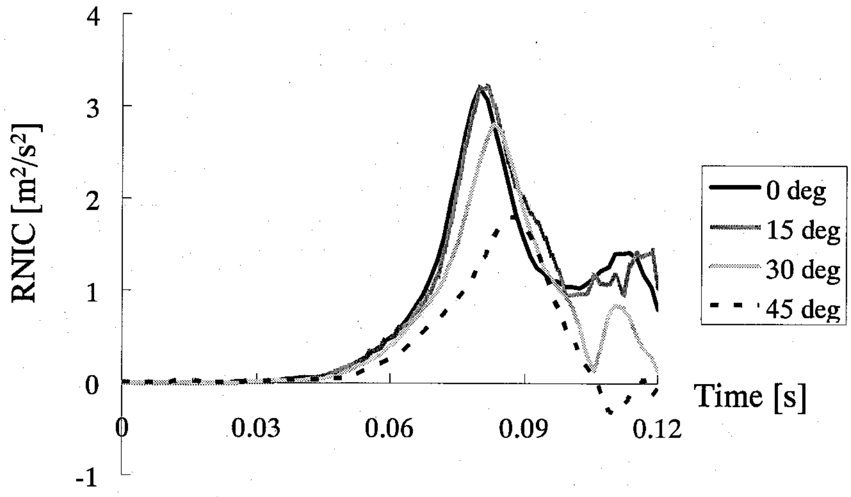


Fig. 4.7 The results of RNIC
(e) Comparison of the peak RNIC with respect to the impact angles
(f) The time history curves of the RNIC

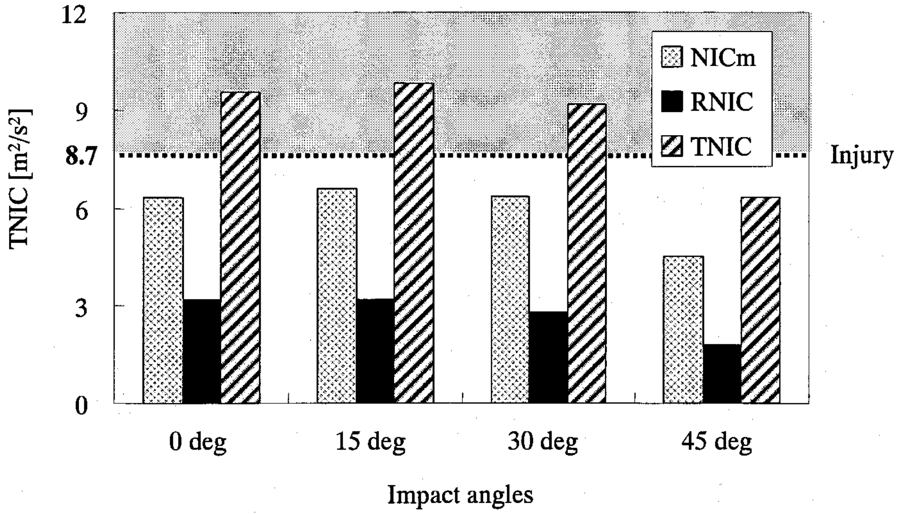


Fig. 4.8 Comparison of the results of NIC_m, RNIC and TNIC

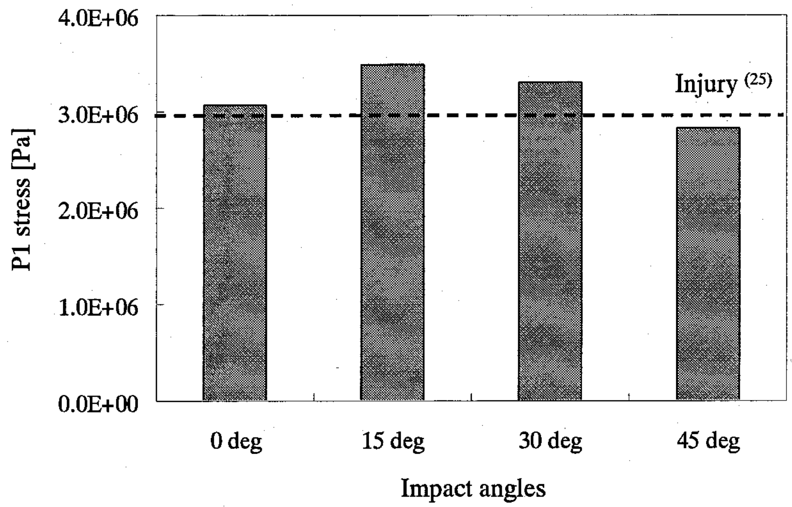


Fig. 4.9 Comparison of the peak tensile stresses on the intervertebral disk at C5-C6

4.5 Discussion

A validated neck injury criterion is an important tool for checking the injury diagnoses and designing a prevention system for the neck injuries in a rear-end impact. The Neck Injury Criterion (NIC) proposed by Boström et al. has been conventionally used to evaluate neck injuries. In the results of applying the NIC in our study, the NIC decreased with the increased impact angles because the translational acceleration in the X-axis decreased, as shown in Fig.

4.3, while lateral acceleration of the head occurred in oblique impacts. The lateral component was unused for calculating the NIC, and thus it was possible to give low evaluations of injury risk in oblique impacts. Alternatively, NIC_m was obtained with the vertical and lateral components aside from the horizontal components. NIC_m in the 15° and 30° impacts indicated higher risk than that in the 0° impact, and those were opposite from the results of the NIC. However, NIC_m were lower than the NIC, NIC_m could give lower evaluations of the injury risk than the NIC.

When the impact occurred in oblique directions, it was difficult to consider the effect of three-dimensional cervical movement on neck injuries because each vertebral behavior changed depending on the impact direction. It was considered that the cervical behaviors were influenced by not only the shear movement but also the rotational behaviors such as lateral bending and axial torsion. Thus, we focused on the influence of rotation between the head and chest. As shown in Fig. 4.4, the angular accelerations affected between the head and chest in oblique impacts, seemed to cause bending and torsion in the cervical spine. In addition, the peak angular accelerations in the oblique impacts were not attenuated, but increased compared to that in the 0° impact differently from the translational acceleration. Therefore, rotation between the head and chest could be assumed as an important factor cause neck injury. The newly proposed criterion RNIC showed the same estimate between the 0° and 15° impacts, and that decreased in the 30° and 45° impacts. In the formula for calculating the RNIC, the relative rotation between the head and chest was an important factor with the angular acceleration. In the 15° impact, although the rotation between the head and chest slightly decreased, the RNIC was same as in the 0° impact due to the increase of the angular acceleration. However, in the 30° and 45° impact, both the rotation and angular acceleration decreased, and thus the RNIC was reduced. In the 30° and 45° impacts, since the whole body significantly moved in the oblique direction more than the case of the 0° and 15° impacts, the impact affecting the head might be attenuated. It was considered that restraint of the body by the seat or seatbelt influenced the severity of the rotation of the head.

In this study, any criterions were not over 15 m^2/s^2 applied as the threshold. Croft et al., however, reported the occurrence of a neck injury in the case when the NIC was below 15 m^2/s^2 in the rear-impact test with volunteers, and this injury was likely due to hyperextension of the neck [(23)]. As the NIC did not consider hyperextension as an injury mechanism, it was not possible to validate in such a case. Panjabi et al. proposed the IV-NIC (Intervertebral Neck Injury Criterion) based on the consideration of injury to the ligament tissues, annulus fibers, and facet joints due to intervertebral motion beyond the physiological limit [(4)]. The IV-NIC demonstrated a good correlation with the injury severity and the NIC, with 95 % confidence limit; the soft-tissue injury threshold was 8.7 m^2/s^2. Also from the results of the stress analyses in this study, the peak tensile stresses over the injury threshold were investigated on the intervertebral disks at C5-C6 in the 0°, 15° and 30° impacts, and the NIC in the 0° and 15 ° impacts were 9.3 and 8.8 m^2/s^2, which were well supported by the results reported by Panjabi et al.. The NIC was 8.0 m^2/s^2 in the 30° impact, which was lower than 8.7 m^2/s^2. In the 30° impact, the injury risk was given a lower evaluation because the NIC was reduced due to the decrease of the translational acceleration in the anteroposterior directions. While the rotational behaviors such as lateral bending and torsion caused increased the stresses on the soft tissues. In the oblique impacts, as the translational acceleration occurred also in the lateral direction, thus NIC_m seemed to be more suitable for the criterion rather than the NIC. However, NIC_m might give a lower evaluation of the injury risk than the NIC due to the effect of the rotation. Therefore, by combining the RNIC with NIC_m, the TNIC showed the value of 9.5, 9.8, and 9.2 m^2/s^2 in the 0°, 15° and 30° impacts, which could indicate the injury risk.

The limitations of the present study must be considered to apply the RNIC and TNIC as a criterion validating whiplash injury in a rear-end impact. Although the RNIC was proposed under the consideration of the head rotation, it has not yet been clear the effect on the soft tissues injury in the cervical spine. Thus, the correlation between the head-neck rotation and the soft tissues injury will need to be evaluated. An additional limitation is that, since our results were based on the numerical simulations using the FEM analyses, it could not be confirmed that the clinical evidences on an actual soft tissues injury affected the impact. Although the FEM model used in this study was validated by the bio-fidelity, some differences compared to human anatomy have remained, the FEM model will need to be improved. While the FEM model is an effective tool to investigate the biomechanical changes in a human body, it is useful to validate the applications of protecting humans from the injuries in car collisions. In a future study, the limitations mentioned above should be improved, and the RNIC will be developed as an effective criterion for whiplash injury in three-dimensional cervical behaviors in a rear-end impact.

4.6 Conclusions

When the impact affected the cervical spine in the oblique directions, conventional injury criterions, the NIC and NIC_m, might give a lower evaluation of injury risk due to the decrease of the translational accelerations, while the soft tissues injury severity might increase, caused by the effect of bending and torsion affecting the cervical spine. The RNIC was based on the head-neck rotation, which could indicate the effect of the rotation on evaluating the injury risk. In addition, the TNIC combining the RNIC with NIC_m was suggested as a new criterion, and that may be used to validate neck injuries in rear-end impacts.

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CHAPTER 5

THE EFFECT OF HEAD ROTATION ON WHIPLASH INJURY IN A REAR-END IMPACT

5.1 Introduction

A whiplash injury is well known as a common cervical injury suffered by many people in car accidents, and results in expensive insurance payments, e.g., US$ 3.0 billion dollar in Japan, in the year 2000 [(1)]. Although clinically identifying the injured parts is difficult, clinical symptoms include headache, neck pain, dizziness, tinnitus and others. In order to design a prevention system for whiplash injuries, a criterion for validating the injury severity must be developed.

In many studies, the cervical behaviors in a rear-end impact have been investigated to reveal the whiplash mechanism [(2) ~ (12)]. It was assumed that the cause of whiplash injury was a soft tissues injury due to the complex cervical behavior during impact, and the injury severity depended on the degree of impact. Only two-dimensional cervical behaviors were observed in the sagittal section under the consideration of the straight impact. In real accidents, however, collision may occur from any directions, and the occupants may be facing directions other than forward when the accidents happen: thus, cervical behaviors can be more complex and three-dimensional. We have investigated three-dimensional cervical behaviors using the finite element human whole body model. When the impact occurred in oblique directions, we found severe torsion and lateral bending motion in the cervical behavior, which increased the stresses on the soft tissues [(13), (14)]. Therefore, in order to prevent a whiplash injury, the three-dimensional cervical motion needs to be considered, and it is important to verify the effectiveness of the preventive system for whiplash injuries. From this view point, a validated neck injury criterion is an important tool for checking a whiplash injury prevention system.

Based on our previous results, the Rotational Neck Injury Criterion (RNIC) was proposed with the rotational components between the head and chest. However, the effect of the head rotation on the neck injury, and the correlation between the RNIC and the injury severity, has been unclear. In this study, we focused on the effect of head rotation on a soft tissues injury and the availability of the RNIC for evaluating whiplash injuries.

5.2 Finite element model and Methods

5.2.1 Finite element head-to cervical model

As shown in Fig. 5.1, the FEM head-to-cervical model was used to evaluate the biomechanical changes during a rear-end impact. The FEM model used in this study was based on the THUMS (Total HUman Model for Safety, automobile occupant model Ver1.52 ß-031117, TOYOTA Central R&D Labs., Inc.), which was a three-dimensional human whole body finite element model developed to evaluate the injuries suffered by a human body during automobile accidents. The material properties in THUMS were based on previous literatures [(15), (16)]. Each body part of the THUMS is selectively validated, and comparing the human responses in impacts were compared with published human cadaveric test data to simulate the bio-fidelity [(17), (18)]. The cervical parts of the THUMS were also constructed to simulate a human anatomical shape and properties of the human cervical spine. The modeling details of each component of the THUMS neck segment are shown in Table 1.

5.2.2 Finite element analysis conditions

The FEM analyses were conducted under the assumption that rotation affected the head during a rear-end impact. First, as the condition of the rear-end impact, the velocity curves (Fig.

5.2) based on our sled test at 5 km/h with Hybrid-Ⅲ dummies were applied to the first thoracic vertebra of the FEM model. Next, as the condition of the head rotation, the angular velocity was applied to the center of gravity of the head. The rotation affected around the X-axis under the two cases (CASE-A and CASE-B): the time when the angular velocity peaks was different between the CASE-A and B. In addition, in each case, three conditions (a, b, c) such that the peak angular velocities were 0.35, 0.7, 1.4 [deg/s] were applied. The angular velocity curves are shown in Fig. 5.3. The rotation around all axes and movement in the Y-axis of the T1 was constrained. The FEM analyses were conducted up to the first 100 ms. LS-DYNA Ver.970 (Livermore Software Technology Corp., USA) and Hyper-works Ver.7.0 (Altair Engineering Inc., USA) were used as the FEM solver and pre-post processor. In order to verify the FEM model, the results of the head displacement relative to the T1 and the head acceleration in the horizontal direction in the simulation were compared with those in previous experiments. The results of the simulation and experiment seemed in good agreement as shown in Fig. 5.4.

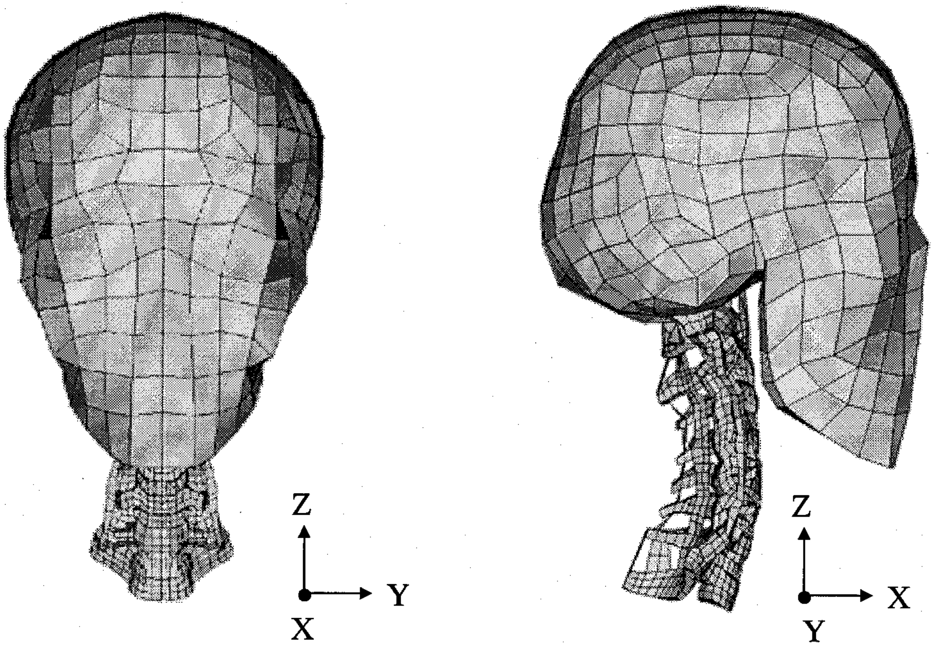


Fig. 5.1 The finite element head-to-cervical model

Table. 1 Each component of neck segment in the THUMS

Segment	Young's Modulus [MPa]	Poisson ratio	Element type
Cortical bone	5000	0.3	Solid
Cancellous bone	70	0.3	Solid
Nucleus pulposus	0.198	0.499	Solid
Annulus fibrosus	13.3	0.4	Solid
Facet cartilages	12.6	0.4	Solid
LF	15.07	0.22	Shell
LN	30.16	0.22	Shell
ITL	15.08	0.22	Shell
ALL	3.25	0.22	Shell
PLL	3.25	0.22	Shell
Muscles	Hill-type model (passive state)	–	Beam

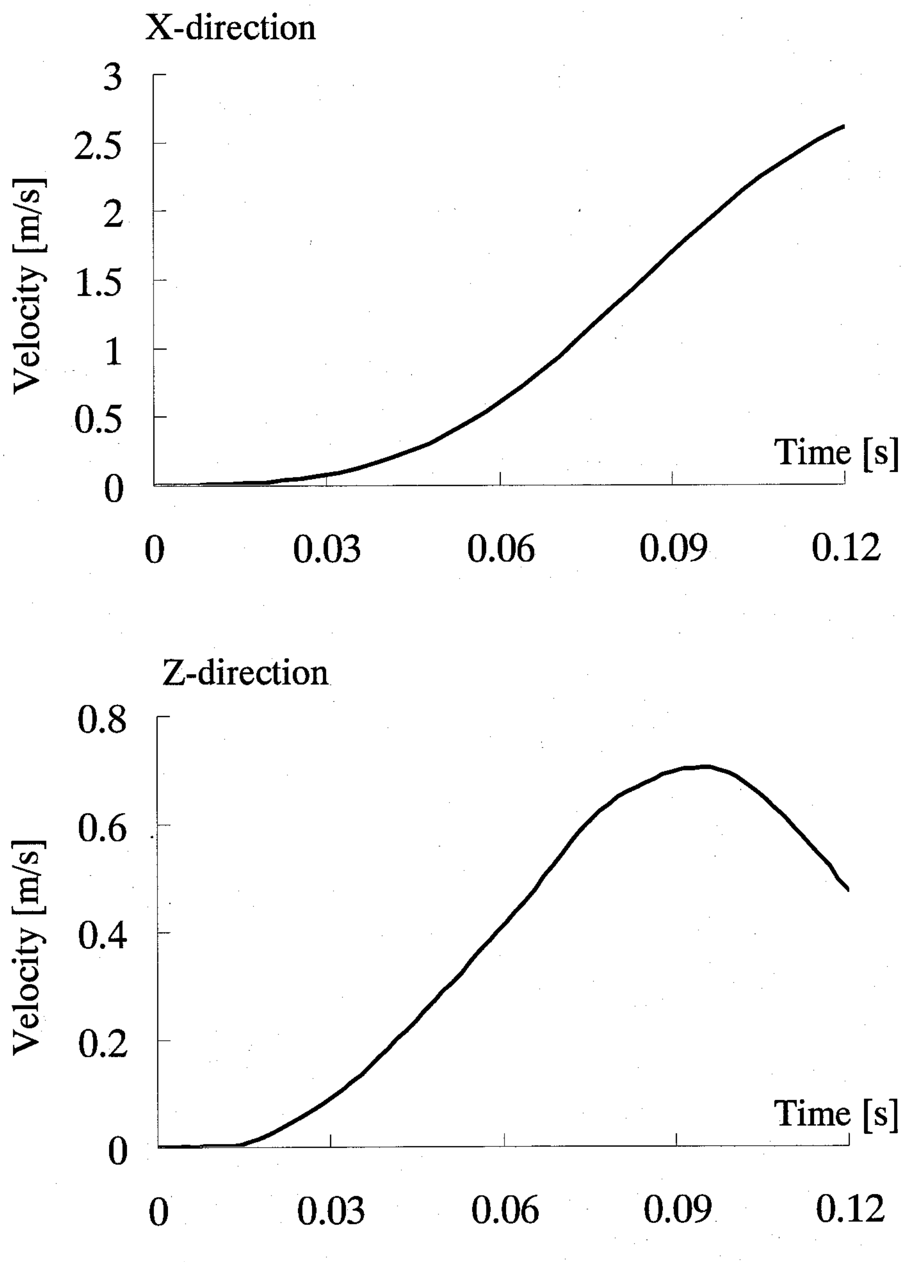


Fig. 5.2 The velocity curve applied to the T1 of the FEM model

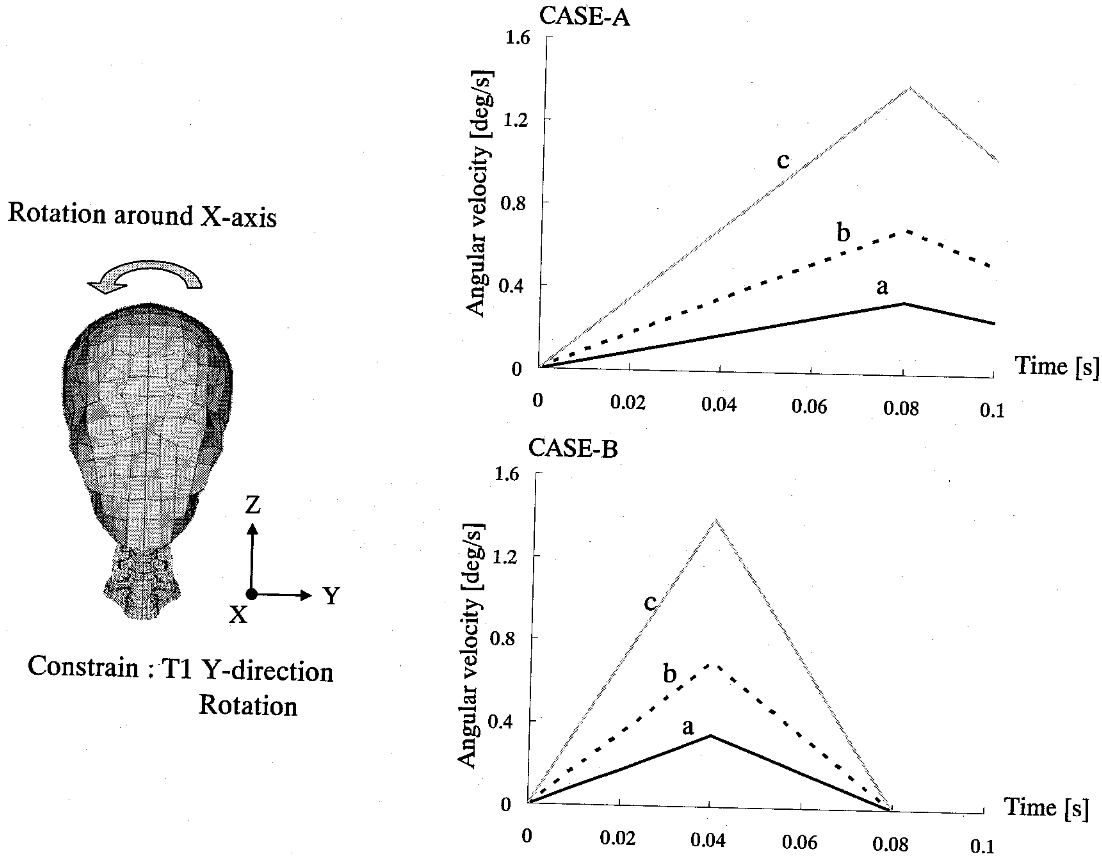


Fig. 5.3 The angular velocity curve applied to the center of mass of the head

5.3 Injury criterions

5.3.1 Neck Injury criterion (NIC)

NIC (Neck Injury Criterion) was proposed by Boström et al. to mathematically model and predict neck injuries in low-speed rear-end collisions (19). The calculation method for NIC can be seen as follows:

NIC = NIC(t = t_{max}) [m^2/s^2] t_{max}: moment of time at the S-shape of the cervical spine

$$NIC(t) = 0.2 \cdot a_{rel}(t) + (v_{rel}(t))^2$$

$a_{rel}(t) = a_x^{T1}(t) - a_x^{Head}(t)$: relative acceleration between T1 and the head

$v_{rel}(t) = \int a_{rel}(t)dt$: relative velocity between T1 and the head

NIC should not exceed $15m^2/s^2$ if injuries with long-term consequences are to be avoided. The estimation was based on the scaling of the pig anatomy to that of the human and comparing with results from volunteers was further supported by Boström et al. and Eichberger et al.

NIC_m was applied some modification by Eichberger et al. to NIC as follows [(20)]. NIC_m was calculated using the resultant acceleration and velocity instead of longitudinal component.

NIC_m $[m^2/s^2]$: 3 ms Maximum of $NIC_m(t)$

$NIC_m(t) = 0.2 \cdot a_{res,rel}(t) + (v_{res,rel}(t))^2$

$a_{res,rel}(t) = a_{T1,res}(t) - a_{Head,res}(t)$: relative resultant acceleration between T1 and the head

$v_{res,rel}(t) = \int a_{res,rel}(t)dt$: relative resultant velocity between T1 and the head

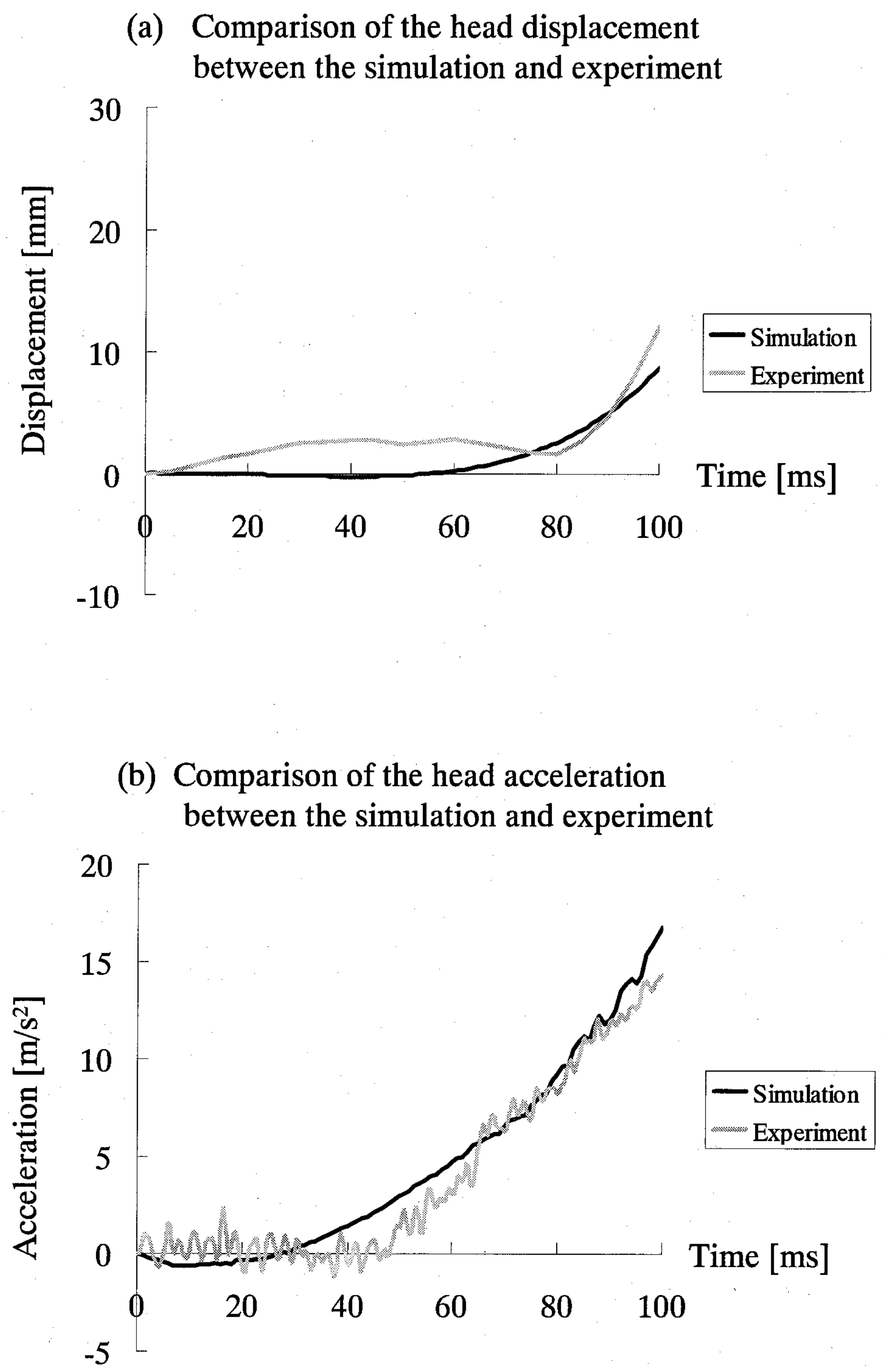


Fig. 5.4 Comparison of the relative head-T1 horizontal displacement and the head acceleration between the simulation and experiment

5.3.2 Rotational Neck Injury Criterion (RNIC)

In our previous study, it was reported that the rotation of the cervical spine such as bending and torsion increased the injury severity when the collision occurred from a posterior-oblique direction. However, it is difficult to take into account the cervical behaviors because the cervical behaviors changed in the vertebral level depending on the impact direction. Therefore, based on the assumption that the rotation between the head and chest influenced cervical behaviors, we proposed the Rotational Neck Injury Criterion (RNIC). The RNIC was calculated as follows:

RNIC [$m^2 \cdot rad^2/s^2$] : maximum of RNIC(t) within the first 120 ms after impact

$$RNIC(t) = 0.2^2 \cdot \{\Delta\theta_{rel}(t) \cdot \dot{\omega}_{rel}(t) + (\omega_{rel}(t))^2\}$$

$\dot{\omega}_{rel}(t) = \dot{\omega}_{T1}(t) - \dot{\omega}_{Head}(t)$: resultant angular acceleration of the head relative to T1

$\omega_{rel}(t) = \int \dot{\omega}_{rel}(t)dt$: resultant angular velocity of the head relative to T1

$\Delta\theta_{rel}(t) = \theta_{T1}(t) - \theta_{Head}(t)$: resultant rotation angle of the head relative to T1

The RNIC was defined as the peak of RNIC(t). The constant 0.2 in the first formula represents the length of the human neck in meters and is used to conform the unit of the RNIC value to the NIC value.

5.4 Results

5.4.1 Injury criterions

Figure 5.5 shows the results of the NIC and NIC_m. The NIC was 7.3 m^2/s^2, and was almost same in all cases compared to the non-rotation case. In this study, the same impact pulse, as shown in Fig. 5.2, was applied to the horizontal direction in all analyses of rear-end impact, and thus the head response in the horizontal direction might be almost the same in all conditions. NIC_m was 6.8 m^2/s^2, and was almost same in all the cases, the same as the NIC. NIC_m was calculated with the resultant acceleration, which did not reflect on the effect of the head rotation.

In addition, there were no cases when the NIC and NIC_m were over 15 m^2/s^2.

Figure 5.6 shows the results of the RNIC. The RNIC increased when the peak angular velocity increased. The RNIC in the non-rotation case was 8.8 $m^2 \cdot rad^2/s^2$, and increased to 9.3 $m^2 \cdot rad^2/s^2$ in CASE-B-c. In this study, since the T1 of the FEM model was constrained, the angular components of the T1 could not be considered for calculating the RNIC. Therefore, the correlation between the RNIC and the injury severity could not be directly evaluated. Thus, the effect of the head rotation on the injury severity of the soft tissues was evaluated with the increase ratio of the RNIC.

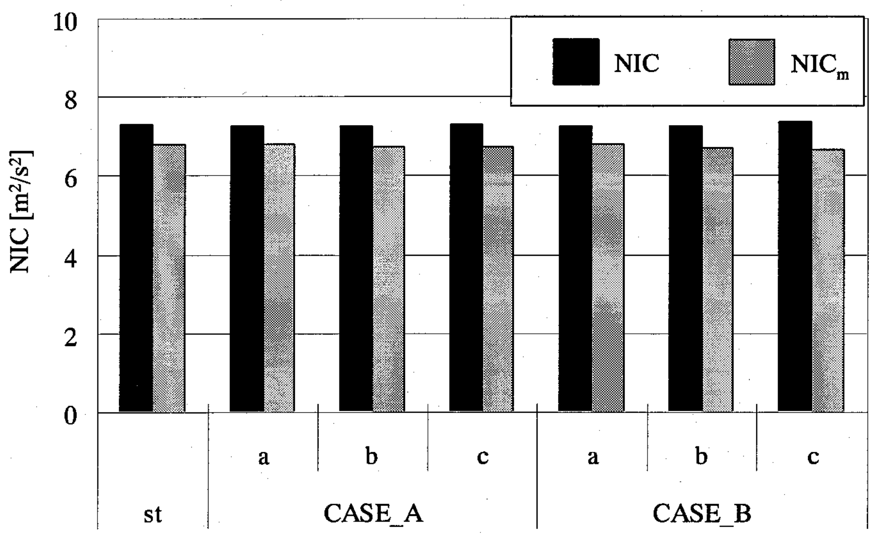


Fig. 5.5 Comparison of the NIC and NIC_m

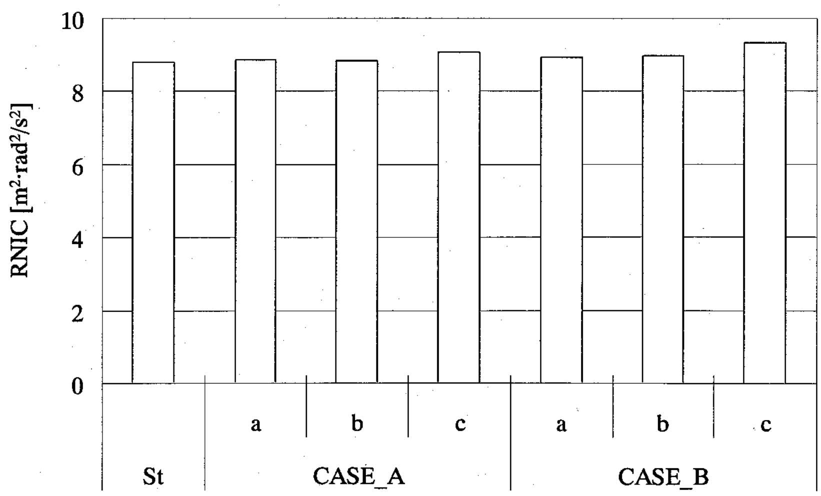


Fig. 5.6 Comparison of the RNIC

5.4.2 Comparison of the peak stresses on the facet cartilages

In our previous studies, we reported that the peak stresses on the facet cartilages increased due to the bending and torsional behaviors affecting the cervical spine. Thus, we focused on the damage to the facet joint, and investigated the stress distributions on the facet joints with increasing angular velocity applied to the head. Additionally, in order to validate the RNIC on the injury severity, we evaluated the correlation between the RNIC and the stress increment.

Figure 5.7 shows a comparison of the peak compressive stresses on the facet cartilages with respect to impact cases. The peak stress at C6-C7 was the highest, but was thought to be due to the constraint applied to the T1. Therefore, the injury severity was evaluated by the increase ratio of peak stress to the non-rotation case. In all the cases, the stress increased as the angular velocity increased, and the increase ratio at C2-C3 and C3-C4 was higher than that at others. Especially, the peak stress at C2-C3 increased by 208% in CASE-B-c, in which the peak angular velocity was 1.4 rad/s, compared to that in the non-rotation case. As shown in Fig. 5.8, the peak shear stresses also increased as the angular velocity increased, and that at C3-C4 increased by 202% in CASE-B-c compared to that in the non-rotation case. Also the peak tensile stress increased most at C2-C3, and that was up 182% compared to that in the non-rotation case. Interestingly, the highest compressive and shear stresses were shown in CASE-B, but the peak tensile stress was shown in CASE-A. In addition, the tensile stress at C3-C4 in CASE-A increased more than that in CASE-B. However, in CASE-B, the tensile stress at C4-C5 increased in the same ratio as for C2-C3, and that at C5-C6 was higher than that in CASE-A. The peak angular velocity was the same between CASE-A and B. In CASE-B, however, the angular velocity peaked at 40 ms, which was earlier than that in CASE-A. This indicated that the angular acceleration in CASE-B was higher than that in CASE-A. In view of this, the compressive and shear stresses on the facet cartilages were proportional to the angular acceleration affecting the head, but the tensile stress was not proportional to that.

Figure 5.10 shows the correlation between the increase ratio of the peak stresses and the

RNIC at the vertebral level. The increase ratio of the compressive stresses was well proportional to the RNIC ratio in all the vertebral levels. Also the increase ratio of the shear stress was well proportional to the RNIC, but that decreased slightly at C4-C5. The increase ratio of the tensile stress was not proportional to the RNIC in the upper cervical levels (C2-C3 and C3-C4), but was well proportional to that in the lower cervical levels (C4-C5 and C5-C6). The rate of stress increase in the upper cervical level was higher than that in the lower level in the same RNIC. Therefore, the RNIC might be used to indicate the injury severity in the upper cervical levels.

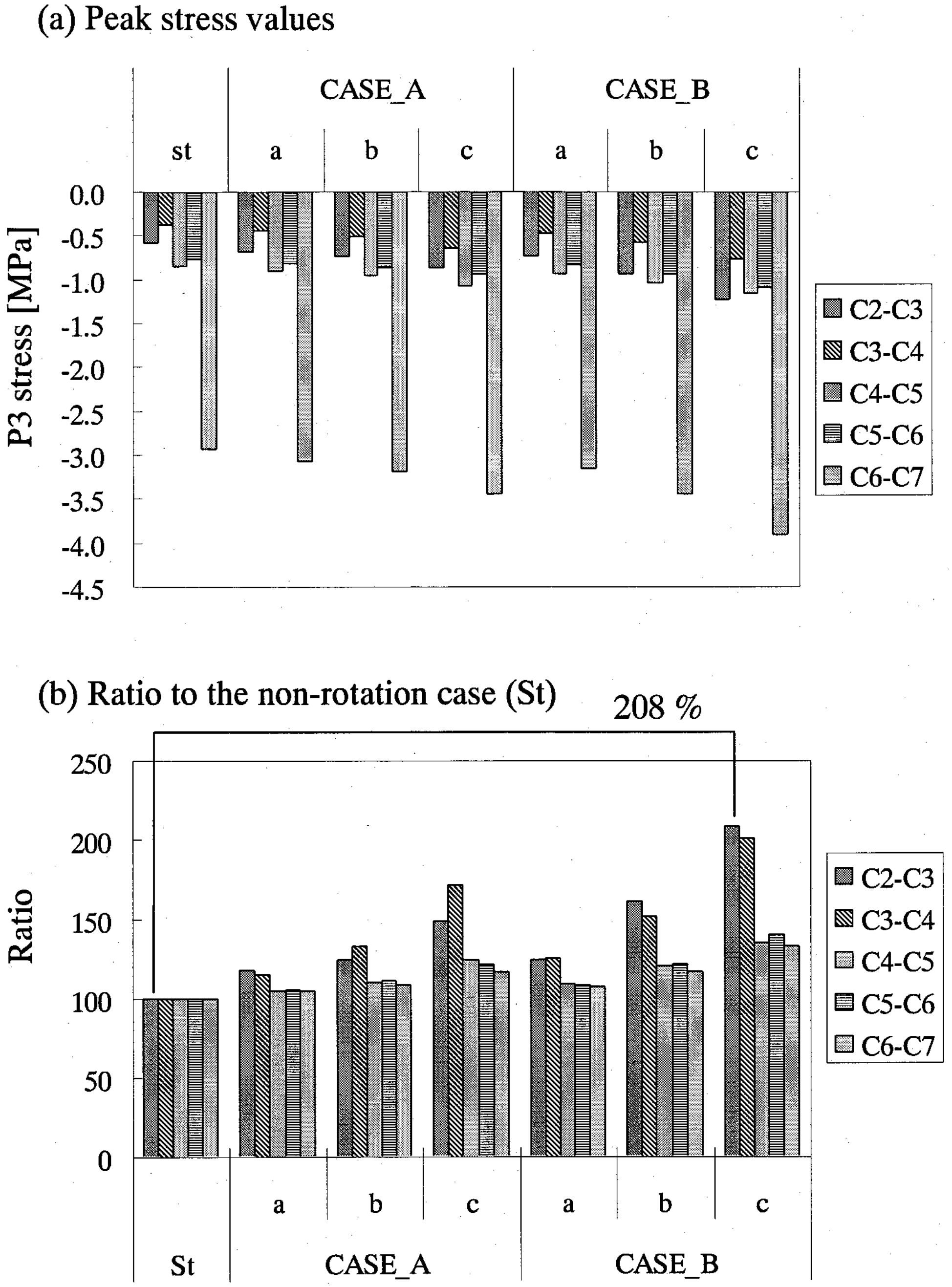


Fig. 5.7 Comparison of the peak P3 stresses on the facet cartilages
(a) Comparison of the peak P3 stresses with respect to the impact conditions
(b) Comparison of the increase ratio to the non-rotation case

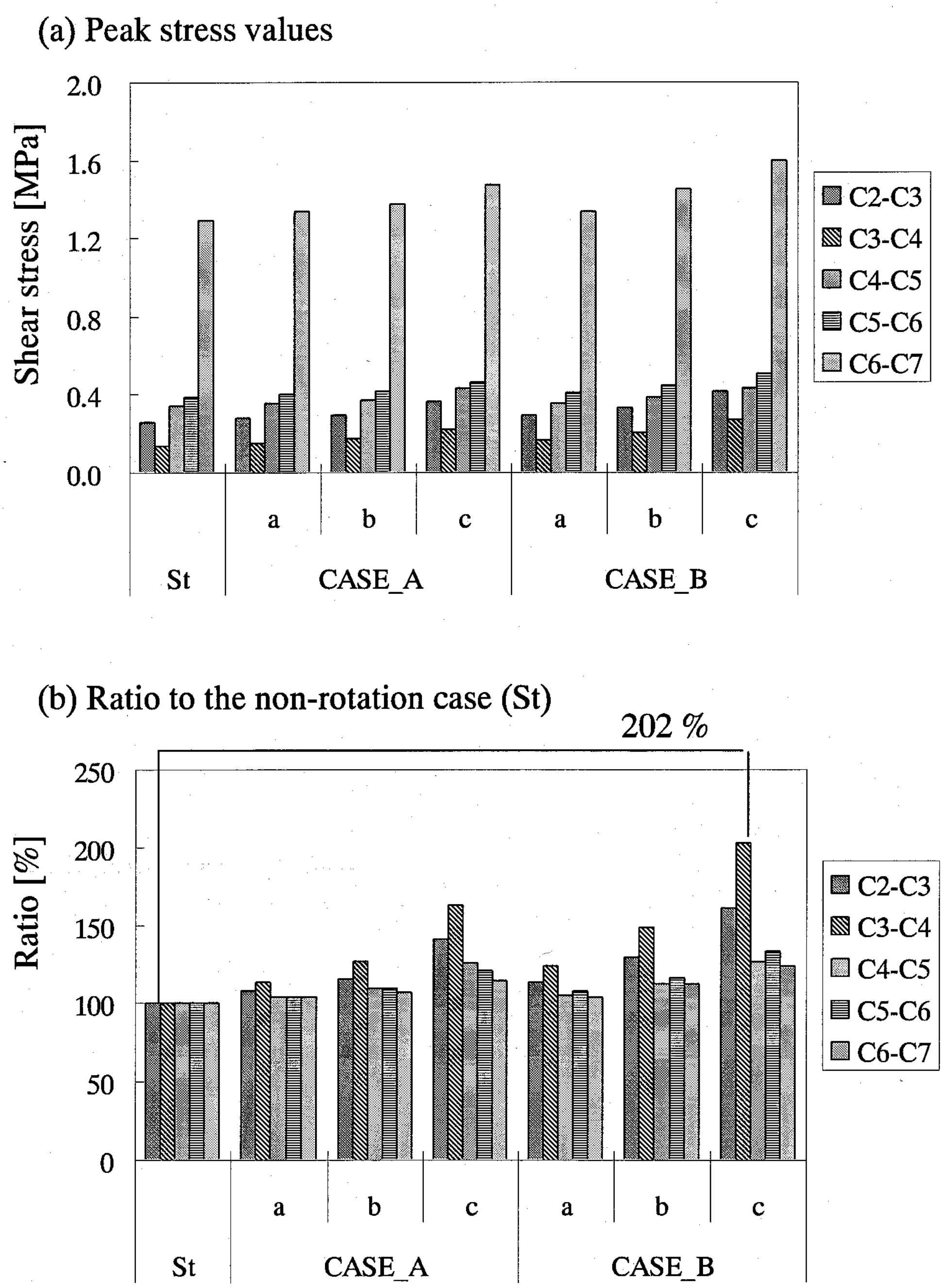
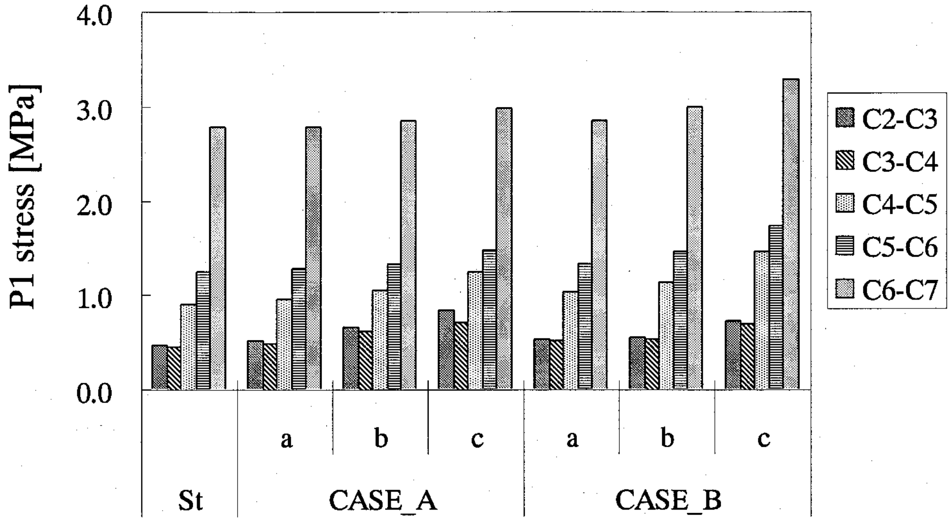


Fig. 5.8 Comparison of the peak shear stresses on the facet cartilages
(a) Comparison of the peak shear stresses with respect to the impact conditions
(b) Comparison of the increase ratio to the non-rotation case

(a) Peak stress values



(b) Ratio to the non-rotation case (St)

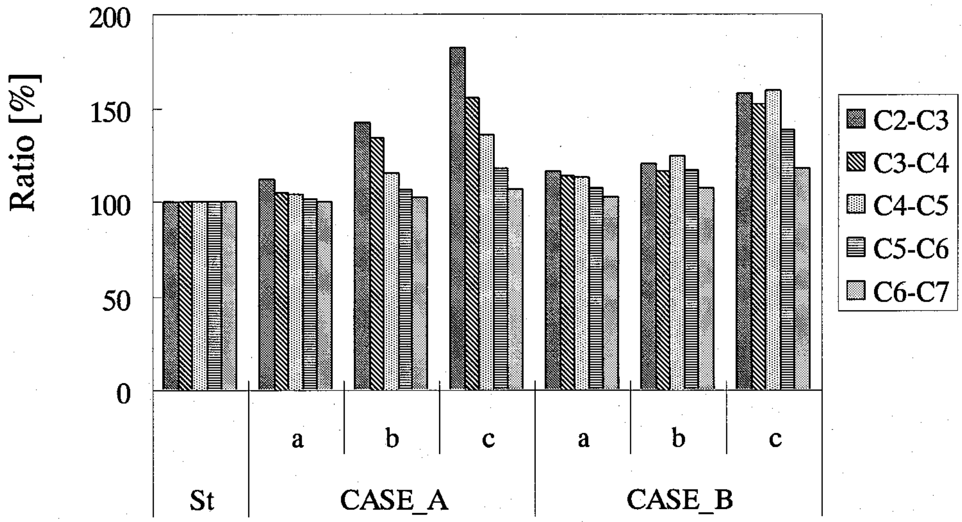


Fig. 5.9 Comparison of the peak P1 stresses on the facet cartilages
(a) Comparison of the peak P1 stresses with respect to the impact conditions
(b) Comparison of the increase ratio to the non-rotation case

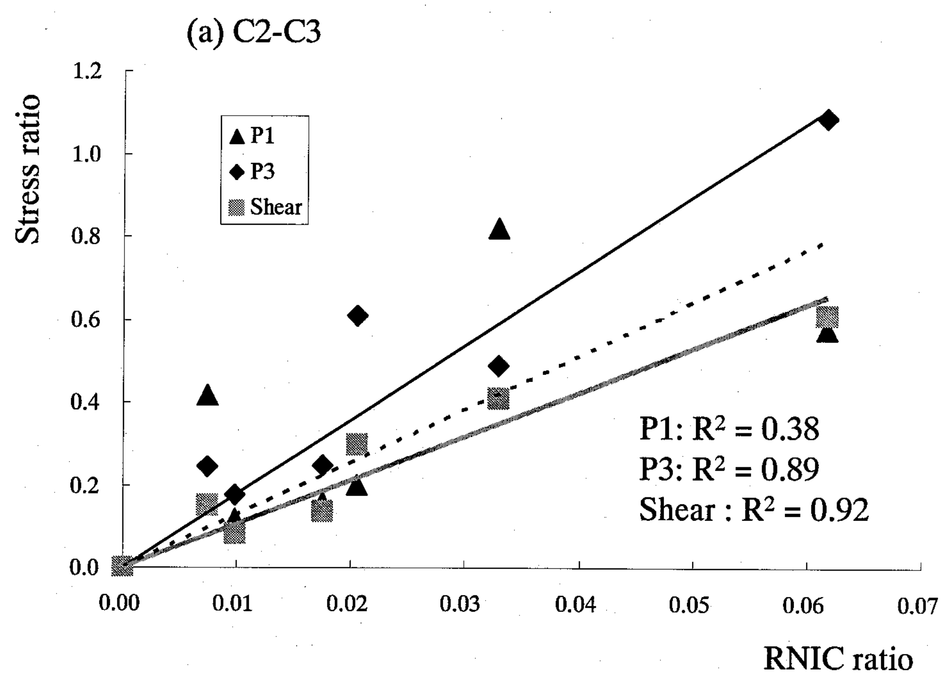


Fig. 5.10(a) The correlation between the increase ratio of the peak stresses and RNIC at C2-C3

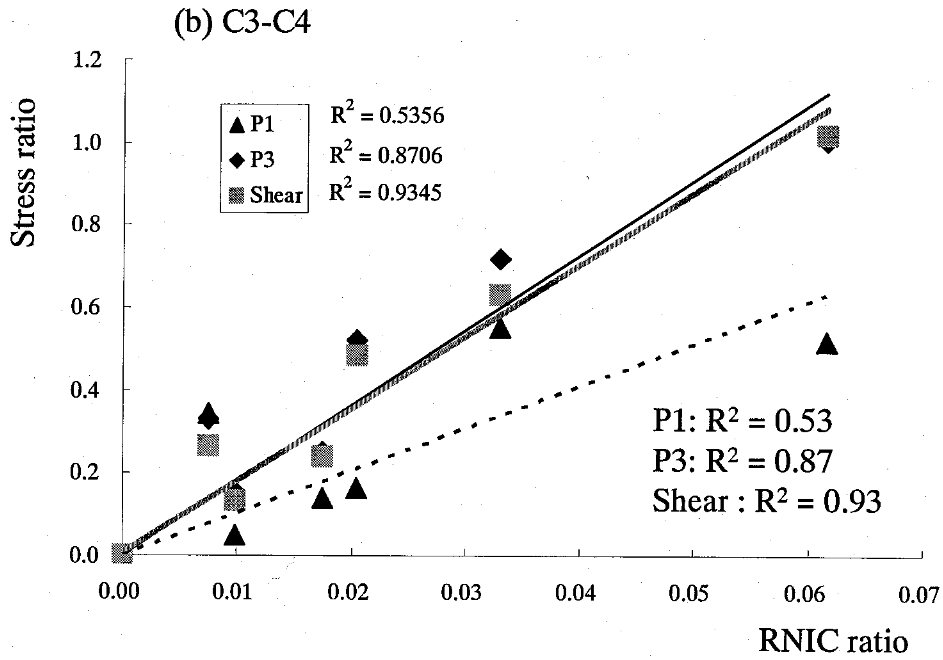


Fig. 5.10(b) The correlation between the increase ratio of the peak stresses and RNIC at C3-C4

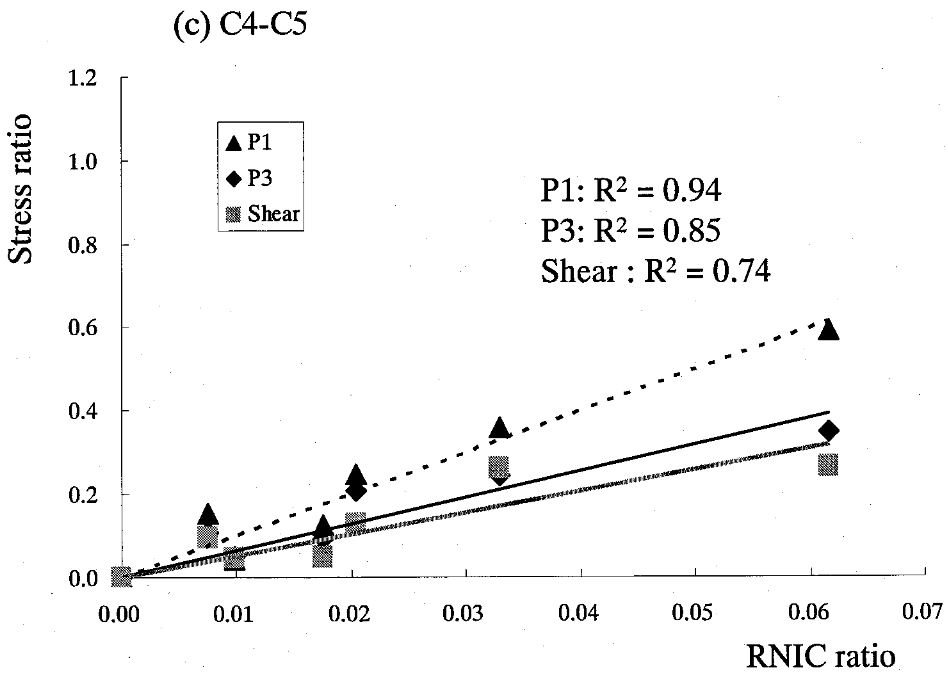


Fig. 5.10(c) The correlation between the increase ratio of the peak stresses and RNIC at C4-C5

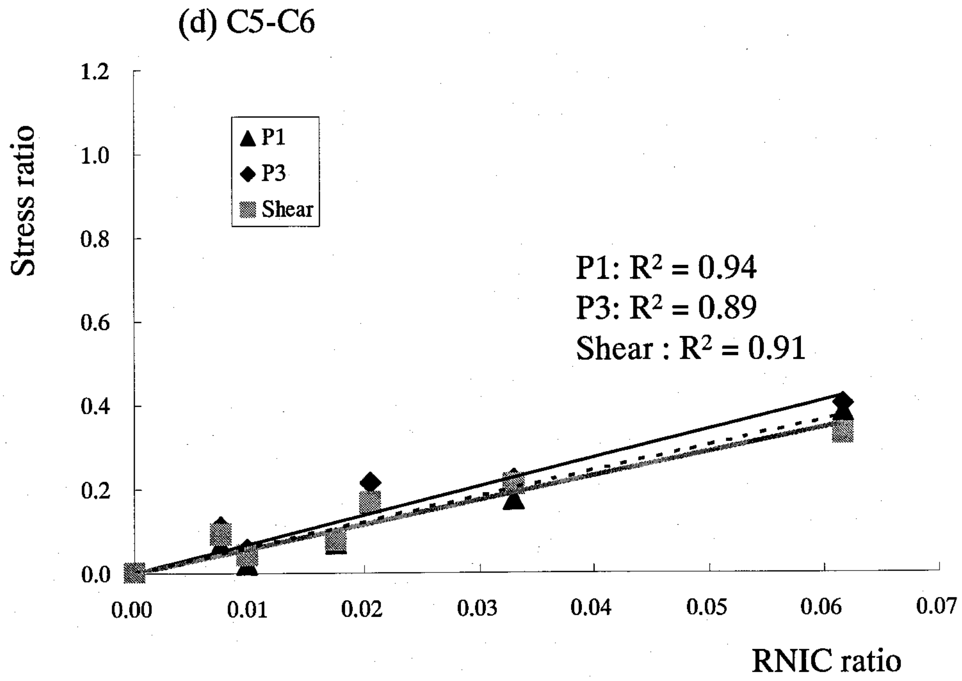


Fig. 5.10(d) The correlation between the increase ratio of the peak stresses and RNIC at C5-C6

5.5 Discussion

In this study, we focused on the effect of head rotation on the injury severity suffered in the facet joints in a rear-end impact. Our results showed that the head rotation could increase the injury severity due to the increasing stresses on the facet joints. In many studies, it was reported that injury of the facet joint was related to the cause and clinical symptoms of whiplash injuries [(6), (8), (21) ~ (25)]. Kaneoka et al. suggested that the neck pain might be caused by inflammation of the synovial fold in the facet joints [(8)]. Yoganandan et al. suggested the pinching mechanism in the facet joints, and that caused headache and neck pain, respectively [(6)]. They reported the compression and sliding as the main causes of injury in the facet joints, and that resulted in the complex cervical behavior such as the S-curve. In our study, it was shown that injury of the facet joints might occur easily due to head rotation. In our previous studies, as the results of the FEM analyses under the assumption that the impact occurred in an oblique direction, it was shown that torsion and bending affected the cervical spine due to head rotation, and easily increased the stresses on the facet joints. In real accidents, collisions may occur from any directions and occupants may be facing directions other than forward when the accidents happen. From this view point, whiplash injury may be caused also by rotation of the head-neck aside from anteroposterior cervical movement such as the S-curve. It was reported that facet joints injury occurred mainly at C5-C6 [(5), (6), (8)], and injury at C2-C3 was indicated as the cause of whiplash disorders [(21), (22)]. Our results showed that the stress increase in the facet joints was influenced by lateral rotation of the head, and the higher risk of injury in the facet joints both of C2-C3 and C5-C6. This is considered to be the cause complicating the mechanism of whiplash injury.

A injury of capsular ligament due to severe elongation around the facet joints was reported as a cause of whiplash injury [(26) ~ (28)]. We showed the injury risk due to the increase of tensile stresses on the facet cartilages. However, the injury severity of the capsular ligaments could not directly evaluated because that was not modeled in the FEM model used in this study.

This was one of the limitations that could be improved in future studies. In addition, there are some soft tissues unmodeled in the FEM model which were involved in important roles in the cervical spine such as the nerve root, artery, and spinal cord. A future study will improve these problems and investigate soft tissue injuries more accurately.

We could also estimate the availability of the RNIC proposed as an injury criterion. The RNIC showed a correlation with the increase of the stresses on the facet joints. Especially, the RNIC was well proportioned to the compressive and shear stresses in the upper cervical level, as shown in Fig. 5.10. Compression and sliding were well supported as important mechanisms at suffered the facet joints in previous literatures. However, the NIC could not relate to our results, because the NIC was based on the translational movements of the head-neck and injury due to the change of the pressure gradient in CSF. Panjabi et al. suggested the IV-NIC (Intervertebral Neck Injury Criterion) for the evaluation of soft tissues injury, which was considered a hyperextension injury due to the intervertebral motion beyond the physiological limit. That showed the good correlation with the NIC and impact severity, and the availability of predicting intervertebral injury. However, since the IV-NIC was obtained from the flexion-extension motion in the sagittal section, the IV-NIC could not evaluate the effect of lateral and axial rotation. On the contrary, our results showed the effect of rotation, and the availability of the RNIC to estimate the injuries due to the rotational behaviors. In this study, a correlation between the relative ratio of the RNIC and the stresses was shown, but the RNIC value could not determined to predict the injury severity. Because rotation of the T1 in the FEM model was constrained, this could not be considered an effect of the T1 behaviors on calculating the RNIC. In a rear-end impact, since the head-neck motions begin after the impact affects to the thorax from the seat back, T1 behaviors are important when considering the effect on whiplash injuries. Some studies have reported the effect of seat back properties on neck injuries [(29)~(31)], thus seat back properties will also need to be considered.

5.6 Conclusions

It was shown that the risk of whiplash injury could increase easily due to the effect of head rotation in the lateral direction, which increased the stresses on the facet joints in a rear-end impact. While the NIC could not indicate the injury risk in this study, the RNIC showed a good correlation with increasing the compressive and shear stresses on the facet joints in the upper cervical level.

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CHAPTER 6

THE EFFECT OF MUCSLE ACTIVITIES ON WHIPLASH INJURY IN A REAR-ENC IMPACT

6.1 Introduction

A whiplash injury is well known as a common cervical injury suffered by many people in car accidents, and results in expensive insurance payments, e.g. US$ 3.0 billion dollar in Japan, in the year 2000 (1). Although identifying the injured parts clinically is difficult, clinical symptoms include headache, neck pain, dizziness, tinnitus and others (2). In order to design a whiplash injury prevention system, a criterion for validating the injury severity must be developed.

Various mechanisms of whiplash injury have been hypothesized based on observed cervical behaviors in rear-end impact tests using dummies (3) ~ (6), cadavers (7) ~ (10) and volunteers (11) ~ (13). The Hybrid-Ⅲ dummy has conventionally been used to explore cervical behaviors. However, as the neck part of Hybrid-Ⅲ is made of aluminum and hard rubber, the dummy's neck is much stiffer than the necks of living human bodies (3), (4). Dummy neck models such as the TRID-neck (5) and BioRID (6) were developed with improved flexibility more than Hybrid-Ⅲ dummy, but they were not anatomically correct, and were connected with hinge joints. Some researchers have investigated cervical behaviors using whole cadavers (7), (8) or isolated cervical spine specimens (9), (10). Cadavers cannot support their posture themselves, meaning that cadavers are too flexible compared to living bodies. There are also studies that use volunteers to reproduce occupants' neck responses (11) ~ (13). However, the volunteers knew the experimental protocol at the beginning of the test and could thus contract their neck muscles in anticipation of the impact, while real occupants are usually unprepared for any impact. It seems that the muscles have no effect up to 150 ms during rear-end impacts, as reported in some papers [(14), (15)], but some experimental studies suggest that muscle might play an important role [(16), (17)].

We have investigated the cervical behaviors in rear-end impacts with the 3-D human whole body FEM model-THUMS, but the muscle activity has never been considered in our previous studies. Yoshida et al. reported that muscle contraction influenced the cervical behaviors with the FEM model in consideration of the muscle activities with the Hill-type muscle element. In this study, we evaluate the effect of the reflexive time of the muscle contraction on cervical behaviors.

6.2 Finite element model and methods

6.2.1 Finite element model

To analyze head-neck behaviors in a rear-end impact, a 3-D FEM model representing a vehicle occupant was constructed with THUMS (Total HUman Model for Safety, automobile occupant model Ver1.52 ß-031117, TOYOTA Central R&D Labs., Inc.) (Fig. 6.1). The THUMS used in this study represents the 50th percentile of the American male, with a height of 175 cm and weighing 77 kg in a seated posture. The material properties of the THUMS were based on previous literatures [(18), (19)]. Each body part of the THUMS is selectively validated against published human cadaveric test data to simulate human responses in impacts [(20), (21)]. The head-neck segments of the THUMS used in this study are also constructed to simulate adequately the anatomical shape and properties of the human cervical spine. The modeling details of each component of the THUMS neck segment are shown in Table 1.

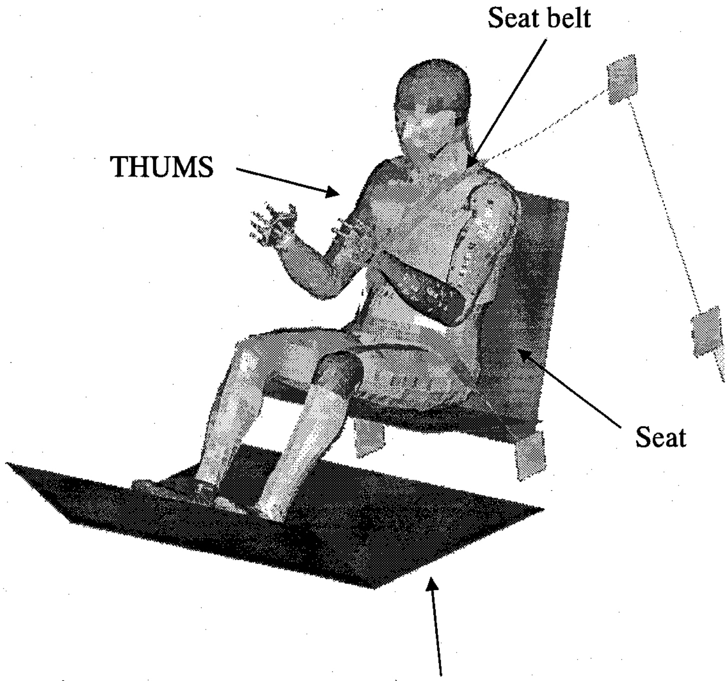

Fig. 6.1 3-D finite element model

Table. 1 Each component of neck segment in the THUMS

Segment	Young's Modulus [MPa]	Poisson ratio	Element type
Cortical bone	5000	0.3	Solid
Cancellous bone	70	0.3	Solid
Nucleus pulposus	0.198	0.499	Solid
Annulus fibrosus	13.3	0.4	Solid
Facet cartilages	12.6	0.4	Solid
LF	15.07	0.22	Shell
LN	30.16	0.22	Shell
ITL	15.08	0.22	Shell
ALL	3.25	0.22	Shell
PLL	3.25	0.22	Shell
Muscles	Hill-type model	—	Beam

6.2.2 Hill-type muscle model

The Hill-type muscle model was adopted as the muscle element in the THUMS [(22), (23)]. The basic model proposed by Hill consists of a contractile element (CE) and a parallel elastic element (PE). The main assumptions of the Hill model are that the contractile element is entirely stress free and freely distensible in the resting state, and is described exactly by Hill's equation. The total force in the muscle F_M is the sum of the force in the CE and the PE because they are parallel:

$$F_M = F^{CE} + F^{PE}$$

The contractile element F^{CE} was defined as follows:

$$F^{CE} = a(t) \cdot F_{\max} \cdot f_{TL}(L) \cdot f_{TV}(V)$$

Here, f_{TL} and f_{TV} are the tension-length and tension-velocity functions for active skeletal muscle, which are specified as the dimensionless length-force curve and the dimensionless velocity-force curve. The dimensionless length L and velocity V are defined as follows:

$$L = \frac{L^M}{L_0}$$

$$V = \frac{V^M}{V_{\max} \cdot S_V(a(t))}$$

Here, L^M, V^M and $a(t)$ are often scaled by the peak isometric force $F_{\max}$, the initial muscle length L_0, and the maximum unloaded CE shortening velocity $V_{\max}$. S_V scales the maximum CE shortening velocity $V_{\max}$ and changes with activation level $a(t)$.

Next, the parallel elastic element F^{PE} was defined as follows:

$$F^{PE} = 0 \quad (L = 1)$$

$$F^{PE} = \frac{F_{max}}{exp(K_{sh}) - 1}\left[exp\left(\frac{K_{sh}}{L_{max}}(L-1)\right) - 1\right] \quad (L > 1)$$

Here, $L_{\max}$ is the relative length at which the force $F_{\max}$ occurs, and K_{sh} is a dimensionless shape parameter controlling the rate of rise of the exponential.

In this study, the peak isometric force $F_{\max}$ was applied to be the stress value of 1 MPa, which corresponds to the upper border of the data reported in the literature in consideration of the cross-sectional area of each muscle [(24), (25)]. The typical normalized tension-length and tension-velocity curves for skeletal muscle were applied as the function curve of f_{TL} and f_{TV} as shown in Fig. 6.2 [(26)]. K_{sh} = 5 and $L_{\max} = 1.05 \cdot L_0$ were implemented for the passive force F^{PE}.

The neck muscles equipped in the THUMS were sternocleidomastoid, scalenus, longus (capitis and colli), rectus capitis, lumped hyoid, multifidus cervicis, splenius (capitis and cervicis), semispinalis (capitis and cervicis), longissimus (capitis and cervicis), levator scapulae, and trapezius (descendes, ascendens, and transversa). The geometry of the muscles was taken from the literature [(27)].

6.2.3 FEM analysis conditions

6.2.3.1 Loading condition

FEM analyses were conducted for the first 200 ms after impact. The velocity curve obtained from our rear-end sled impact test at 8 km/h [(1)] was applied to the seat part of the FEM model. LS-DYNA Ver.970 (Livermore Software Technology Corp., USA) and Hyper-Works Ver.7.0 (Altair Engineering Inc., USA) were used as the FEM solver and pre-post processor.

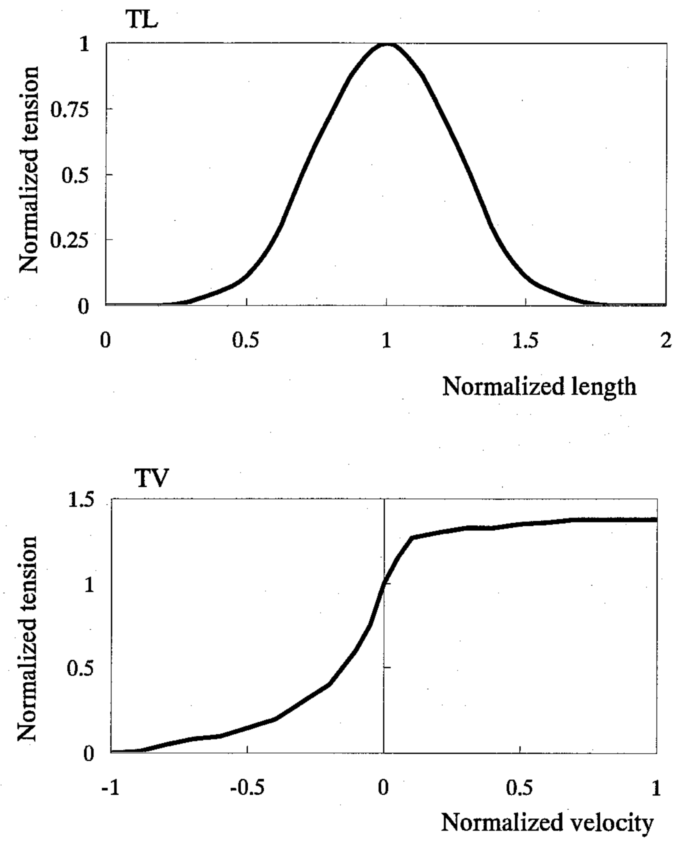

Fig. 6.2 Typical normalized tension-length (TL) and tension-velocity (TV) curves for skeletal muscle

6.2.3.2 Conditions of muscle activity

The muscle activity is defined by the activation level $A = a(t)$, which can depend on the time. In this study, four kinds of activation level were adopted in consideration of the activation level A and the delay time until the muscles were activated as follows:

1) Model A: The activation level A = 0.0 during all of time, in which only the passive force F^{PE} is active. This model was assumed that the occupants were not aware of the impact and the necks were flexible.

2) Model B: The activation level A = 1.0 during all of time, in which both of the contractile force F^{CE} and passive force F^{PE} are active. Thais model was assumed that the occupants were aware of the impact and the neck became rigid.

3) Model C: The activation level was applied as the function of the time. The time history

curve of the activation was shown in Fig. 6.3. An initial delay time until 54 ms was applied as the reflexive contraction of the neck muscles in response to the impact [28]. This model assumed that the occupants were aware of the impact almost at the same time as the collision.

4) Model D: The activation level increased from 54 ms before of the impact. This model assumed that the occupants were aware of the impact just before the collision.

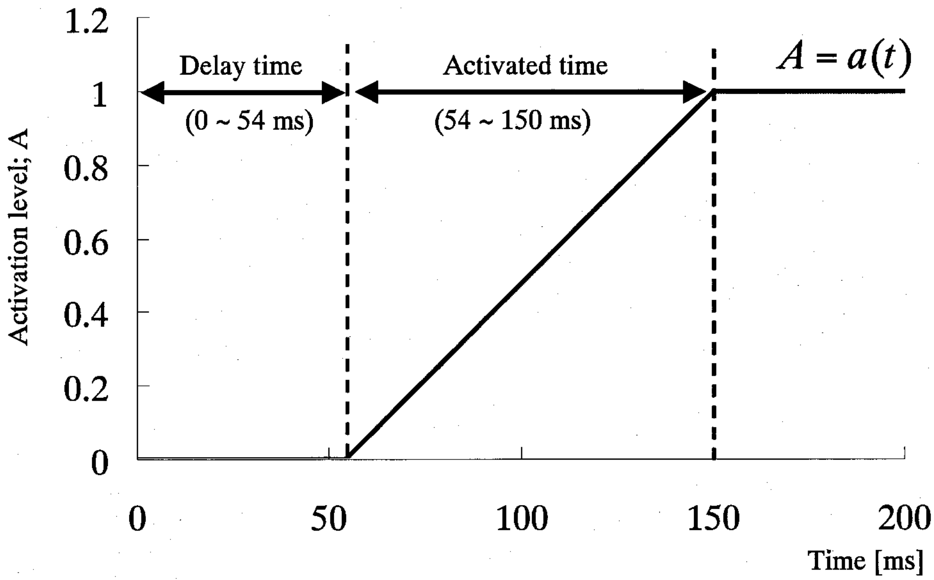

Fig. 6.3 The time history curve for the muscle activation

6.3 Results

As shown in Figs. 6.4 and 6.5, in Model B, C and D, in which the muscles were activated, the head displacements relative to T1 in the horizontal and vertical directions were reduced compared to the displacement in Model A. The peak horizontal displacements in Model B, C and D were decreased by almost 50% compared to that in Model A, as shown in Fig. 6.7. In addition, the head rotation angles were reduced in Model B, C and D (Fig. 6.6). This indicated that the head position was kept and the head extension to the back was restrained due to the muscle contraction. In particular, in Model D,in which the muscles were activated before the impact, the peak displacements were the most decreased of all the models, which means that head-neck extension can be restrained by tightening them before the collision. Therefore, when the muscles are tightened, it seems that neck injury due to hyperextension could be prevented.

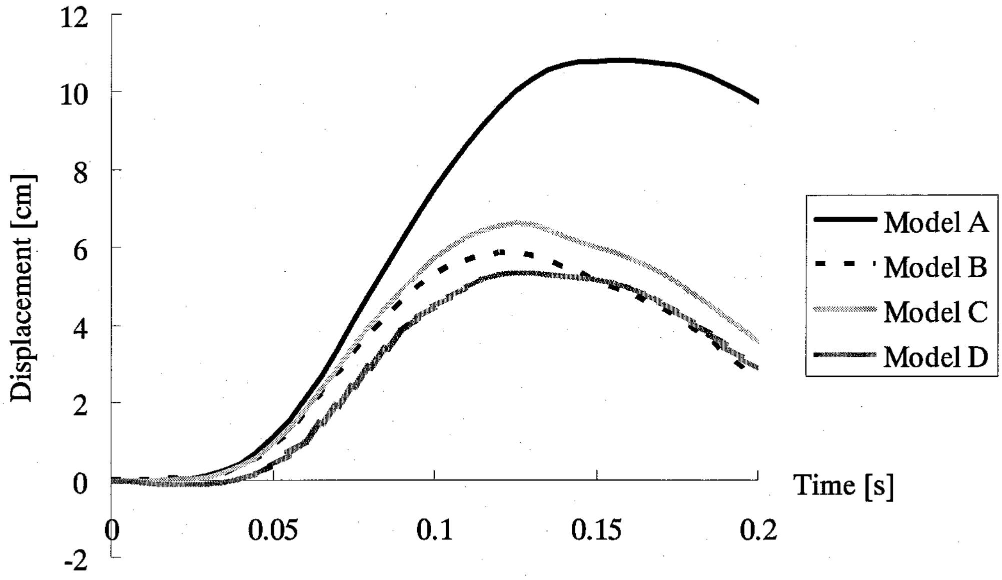

Fig. 6.4 The head horizontal displacement relative to T1

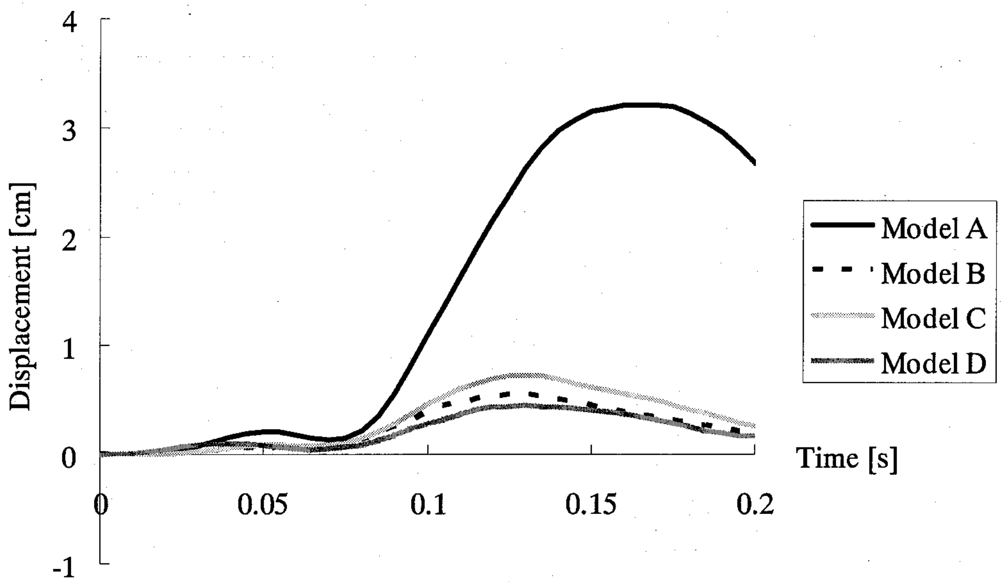

Fig. 6.5 The head vertical displacement relative to T1

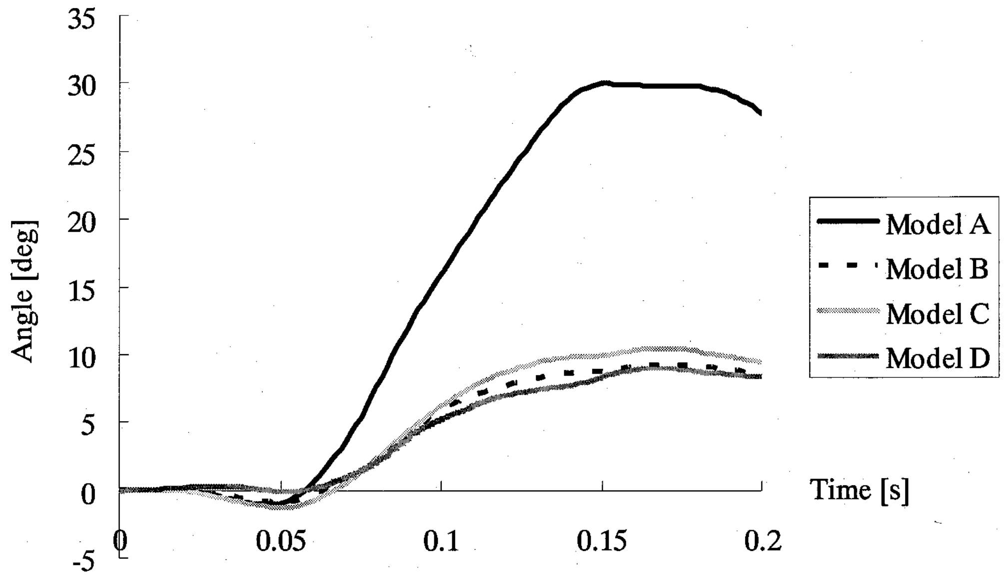

Fig. 6.6 The head rotation angles relative to T1

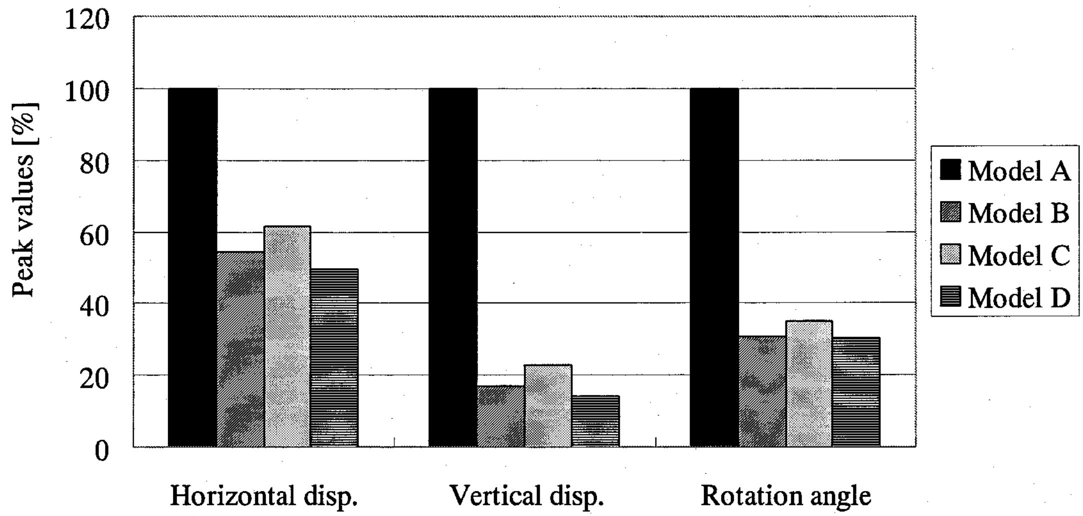

Fig. 6.7 Comparison of the relative peak values of the horizontal and vertical displacement, rotation angles

Figure 6.8 shows the rotation angles between the adjacent vertebral bodies. In Model A, from 50 ms to 90 ms, the vertebral behaviors showed the differences between the upper cervical (C2-C3) and lower cervical (C6-C7) levels, when the local flexion occurred at C2-C3 concurrently with local extension at C6-C7. This shows the occurrence of the S-curve, which is considered an important mechanism of whiplash injury. However, in Models B, C and D, the local flexion at C2-C3 was not shown, and thus it seemed that the S-curve did not occur.

(a) Model A

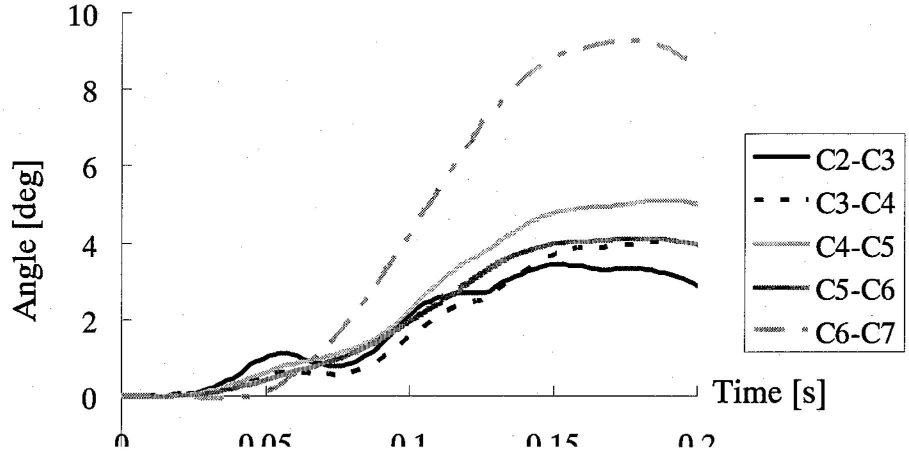

Fig. 6.8(a) The rotation angles of the adjacent vertebral bodies in the Model A

(b) Model B

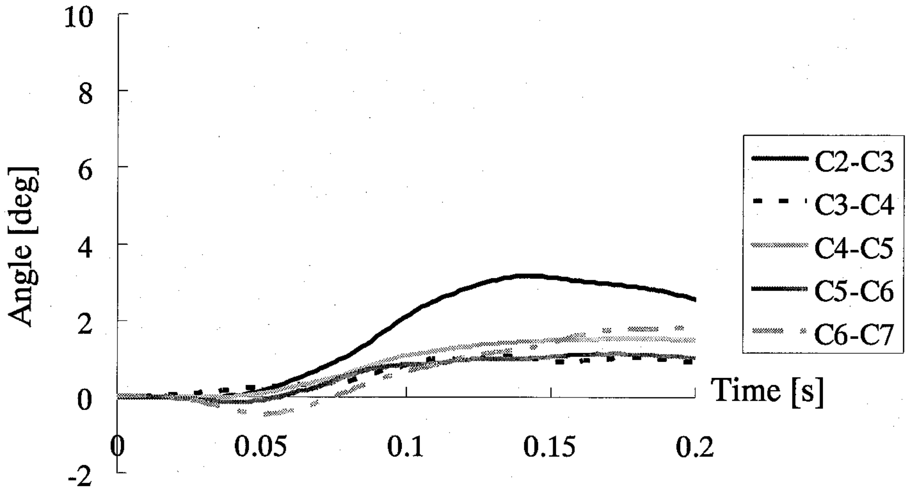

Fig. 6.8(b) The rotation angles of the adjacent vertebral bodies in the Model B

Interestingly, the peak rotation angles at C2-C3 in Models B, C and D decreased by about 10 % compared to that in Model A, while at the other vertebral bodies, the peak rotation angles decreased by about 70% (Fig. 6.9). This means that global head-neck motions were restrained by muscles contraction, but local movements in the cervical vertebra could not be restrained.

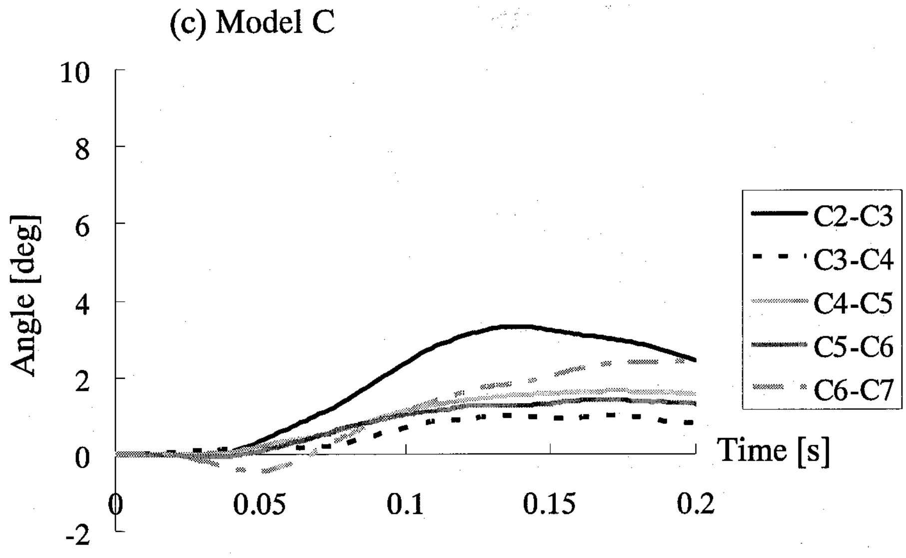

Fig. 6.8(c) The rotation angles of the adjacent vertebral bodies in the Model C

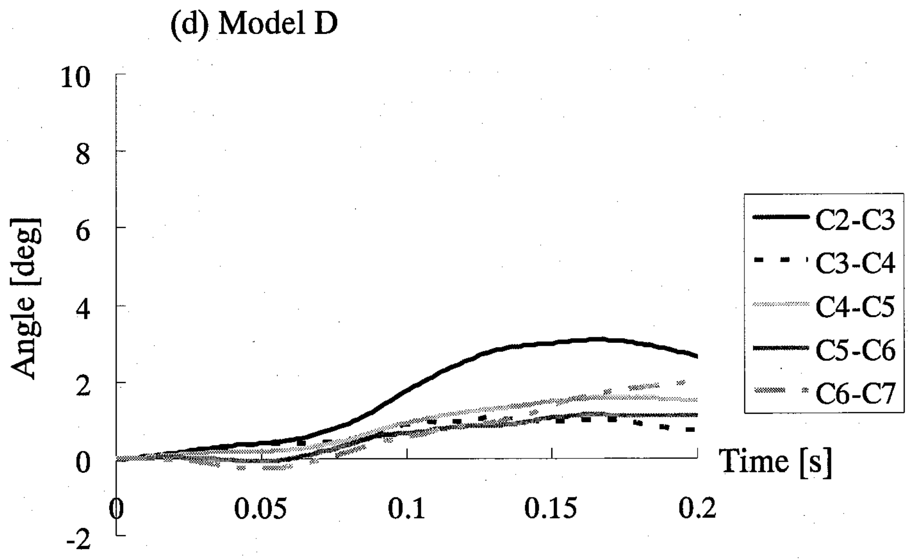

Fig. 6.8(d) The rotation angles of the adjacent vertebral bodies in the Model D

Figure 6.10 shows a comparison of the peak von Mises stresses in the intervertebral disks at C2-C3. Although the peak stresses at C6-C7 in Models B, C, and D drastically decreased compared with that in Model A, the peak stress at C2-C3 scarcely decreased. In Model C, in which the muscles began to contract just after the impact, the peak stress was reduced by 68% at C6-C7, but the peak stress at C2-C3 was reduced only by 7%. Even in Model D, in which the muscle contraction began before the impact, the peak stress at C2-C3 was reduced by 21%. This indicates that the injury could occur around C2-C3 even in the case when the occupants were aware before the collision occurred.

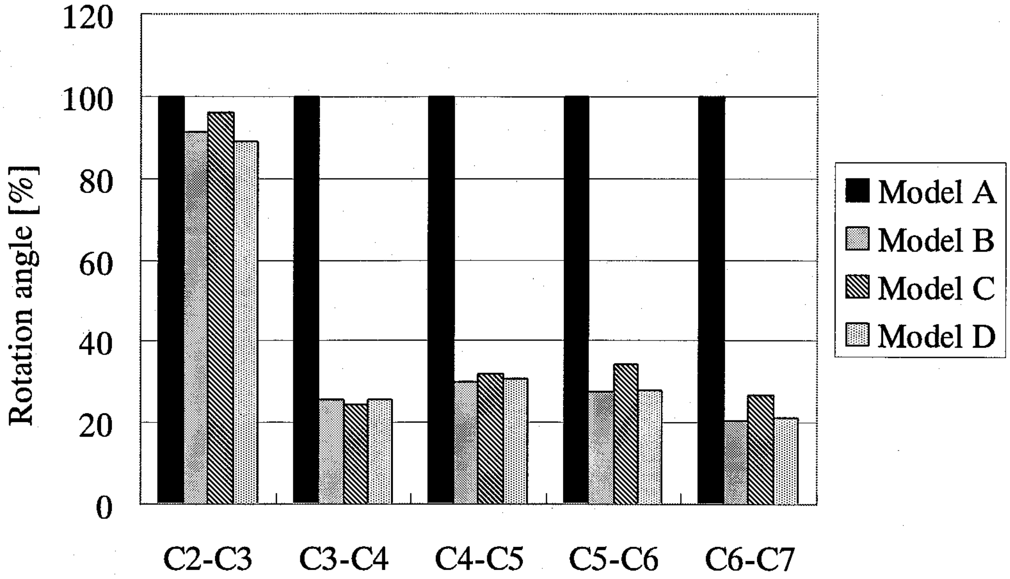

Fig. 6.9 Comparison of the peak rotation angles between the vertebral bodies

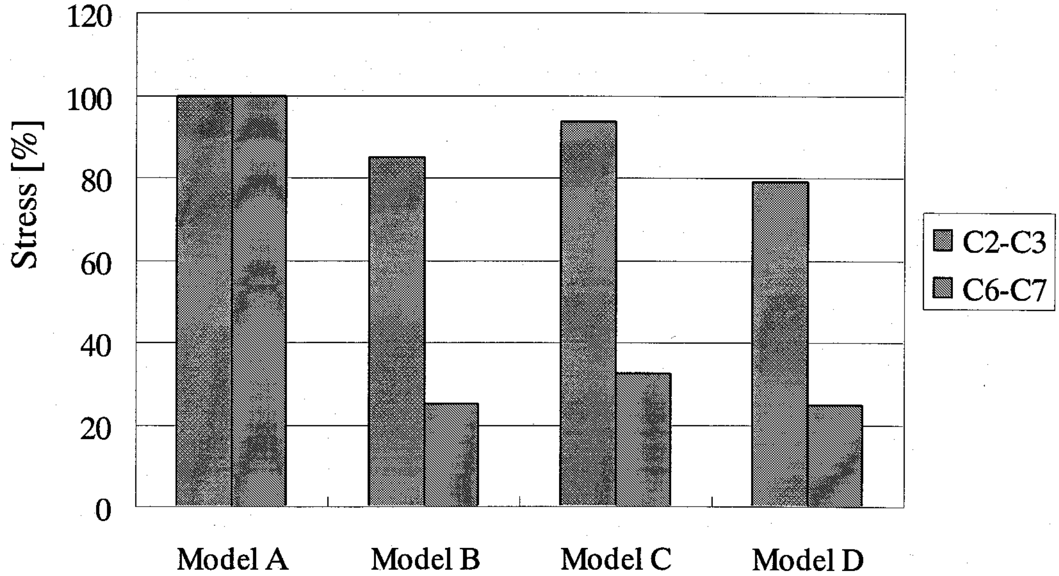

Fig. 6.10 Comparison of the peak von mises stresses on the intervertebral disks

6.4 Discussion

In this study, the effect of muscle contraction focused on the head-cervical behaviors during a rear-end impact using FEM analyses. In many studies with cadavers and volunteers, the role of the muscles were indicated as an important factor influencing cervical motions, and also as a considerable limitation in experiments for reproducing occupants' neck motion in real accidents. The results of our FEM analyses showed that when muscle contraction was occurred, the head-neck extension to the back was restrained during the impact. In the case when muscle contraction did not occur, such as in Model A, this means that the occupants were not aware of the collision, the cervical behaviors showed the S-curve phenomena (9). The S-curve phenomenon has been supported in many studies, and has been assumed to be an important cause of whiplash injury.

When muscle contraction occurred, the vertebral behaviors did not show a different tendency between the upper and lower levels, and thus the S-curve seemed not to have occurred. While there was a slight decrease of the peak rotation angles at C2-C3 under the activated muscles, the peak stresses at C2-C3 also decreased by 21% at the maximum. The injury around the region at C2-C3 agreed with the results of previous studies that used experimental analysis in the sled test (29) ~ (31).

It was reported that a 54 ms reflex delay before beginning the muscle contraction was implemented as the result of EMG under the volunteers subjected to whiplash loading (28). We conducted FEM analyses in consideration of the delay time for muscle activation. In the results, the head-neck behaviors showed differences with respect to the reflex delay time of beginning the muscle activation. The rotation angle between the vertebral bodies and the peak stress on the intervertebral disk were more reduced in the case of beginning the muscle contraction before the impact compared with the case of that at the moment of the impact. In spite of affecting muscle contraction, the risk of injury was indicated at C2-C3.

There are some limitations in this study about the consideration of muscle activation for evaluating the cervical behaviors in rear-end impacts. First, in this analysis, the contraction in all of the muscles began at the same time, but it was reported that the beginning of muscle contraction was different depending on the kind of the muscles involved [32]. In addition, although 1 MPa was applied as the peak contraction force in all of the muscles, the peak contraction force under the whiplash loading was different depending on the kind of muscle [32]. In a future study, the FEM model should be improved regarding the limitations mentioned above for evaluating the cervical behaviors under a state that simulates occupants in real accidents.

6.5 Conclusions

The FEM analyses were conducted to investigate the effect of muscle activity considered the reflexive time before beginning muscle contraction on head-cervical behaviors in a rear-end impact. The head and cervical movements could be restrained by muscle contraction. In addition, head-cervical behaviors were influenced also by the delay time of the beginning of the muscle contraction. However, at the C2-C3 level, there was a possibility of an injury occurring even in the case when the muscle contraction was affected.

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CHAPTER 7

SUMMARY

CHAPTER 1

In Japan, there are approximately 360,000 people suffered the neck injuries in rear-end car collisions in a year. A whiplash injury is well known as a common neck injury occurred in rear-end car collisions, but the occurrence mechanism is yet unknown because the sufficient resolution has not been found to identify injuries even with presently available imaging methods such as MRI, X-ray and CT-scan. There are many studies to reveal a cause of whiplash injury with the experimental and numerical analysis and some hypotheses were suggested from their results. As the experimental methods, the rear-end impact tests were conducted with anthropometric dummies, cadavers and volunteers. However, since neck properties of them were different from that of occupants who were unaware of the impact, and thus it was difficult to accurately reproduce the cervical behaviors during the impact. In the numerical analysis, a head-neck model was represented using rigid bodies or finite element method. The rigid body model was very simplified such as elliptical shape, it was difficult to evaluate the biomechanical changes of the cervical spine that had the complex structure. While the finite element method was useful to estimate the cervical behaviors and the soft tissues injury, it was only investigated two-dimensional cervical behaviors in the sagittal section. From this view point, three-dimensional FEM analysis can be effective tool for evaluating the biomechanical changes of the cervical spine in rear-end impacts.

CHAPTER 2

Many studies have been conducted to reveal a cause of whiplash injury with the experimental and numerical analysis, and the cervical behaviors were investigated during a rear-end impact. Studies of whiplash injuries have been performed with two-dimensional analysis in the sagittal section. However, collisions were happening in various directions on real accidents like frontal, side, oblique and offset-collision. Thus, we investigated the cervical behaviors with three-dimensional finite element human whole body model-THUMS. The FEM impact analysis was conducted on the assumption that a collision occurred from a posterior-oblique direction with 30 degrees and the rotation angles of the vertebral bodies were investigated three-dimensionally in the sagittal, lateral and axial planes. In the results, torsion and lateral bending affected the cervical spine in an oblique impact and the stress increasing was also occurred in the facet cartilages. It was considered to be one of the important factors that the risk of whiplash injuries could increase.

CHAPTER 3

In real accidents, collisions may occur from any directions and occupants may be facing directions other than forward when they happen. Thus, the cervical behaviors can be complex and three-dimensional. We investigated the cervical behaviors in rear-end impacts with three-dimensional FEM analyses. In this chapter, we focused the effect of impact directions on the cervical behaviors and analyzed the stresses occurred in the facet joints regarding to whiplash disorders. The cervical behaviors showed complex motion combining axial torsion and lateral bending, and the vertebral behaviors were changed depending on impact angles. In addition, bending and torsion increased the peak stress in the facet cartilages in oblique impacts. It was sufficiently considered that the facet joints easily injured due to the increase of stresses, and resulted in the important cause of whiplash disorders

CHAPTER 4

In order to design a whiplash injury prevention system, a criterion for estimating the injury severity must be developed. Although the Neck Injury Criterion (NIC) was used conventionally, the NIC was calculated with only translational components in the horizontal direction. From previous results in our studies, we thought that rotational movements such as torsion and bending were also the important factor of whiplash injury in rear-end impacts. Thus, the Rotational NIC (RNIC) was proposed as a new injury criterion and the Total NIC combining the RNIC and NIC_m. In posterior-oblique impacts, the NIC and NIC_m might give the lower evaluation of injury risk because those did not consider the effect of rotational components on neck injuries. The RNIC could reflect the effect of rotation for evaluating the injury, the TNIC combining the RNIC with NIC_m could indicate the injury severity. The RNIC and TNIC may be used to validate neck injuries in rear-end impacts.

CHAPTER 5

The RNIC based on rotational motions were proposed as a new criterion, but the correlation with the injury severity have never been cleared. In this chapter, we evaluated the effect of head rotation on the injury severity suffered the cervical spine in rear-end impacts and investigated the correlation between the RNIC and the soft injury severity. The FEM analyses were conducted the assumption that head rotation was affected during a rear-end impact. In the results, the peak stresses on the facet joints increased with the increased angular velocity applied the head. Especially, in the upper cervical level (C2-C3), the peak compressive and shear stresses severely increased. In addition, the RNIC was well proportional to the stress increasing, and it may be used as a criterion to predict the soft tissues injury.

CHAPTER 6

In experimental studies with cadavers and volunteers, the role of muscles was considered as an important factor influencing the cervical behaviors. Cadavers cannot support their posture themselves, meaning that cadavers are too flexible compared with a living body. Volunteers knew the experimental protocol at the beginning of the test and could tight their neck muscles before the impact. However, in real accidents, occupants may be usually not aware to any impacts. In this study, the FEM analyses were conducted in consideration of the muscle contraction using the THUMS model. Hill-type muscle element was applied in the neck part of the THUMS, the effect of muscle contraction was investigated on head-cervical behavior during impacts. In the results, the head motions were restrained by muscle contraction and the vertebral dislocation was also reduced. Furthermore, the reflexive delay time before beginning the muscle contraction was influenced the head-cervical behaviors. When the muscle contraction began before the impact, the head and vertebral motions were almost restrained. However, the stress level at C2-C3 was scarcely reduced in the activation state of muscles. This indicates that the region around C2-C3 was very vulnerable and could involve the cause of whiplash injury.

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