Development of a Mechanical Impedance Model-Based Computer Simulator for Evaluation of an Active Headrest Mechanism in Rear-End Impact*

Yoshiyuki TANAKA**, Shunsuke FUKUSHIMA***, Masaya YAMASHITA†,
Yoshinobu OOTANI† and Toshio TSUJI***

** Graduate school of engineering, Nagasaki University

1-14 Bunkyo-machi, Nagasaki, Nagasaki, Japan

E-mail: ytnk@nagasaki-u.ac.jp

*** Graduate school of engineering, Hiroshima University

1-4-1 Kagamiyama, Higashi-hiroshima, Hiroshima, Japan

† DELTA KOGYO CO.,LTD.

1-14 Futyu-cho, Aki-gun, Hiroshima, Japan

Abstract

Recent years have seen increasing demand for the development of a car seat with an active headrest (AHR) that is expected as one of strong mechanisms to reduce neck injuries caused by rear-end collisions. This paper develops a mechanical impedance model-based computer simulator that can be used to analyze and design an effective AHR seat by means of the 7 criteria in line with the specifications of Euro-NCAP such as *NIC* and *Nkm*. The developed simulator can reproduce the dynamic behavior of a crash-test dummy (BioRID II) as measured in an actual rear-end collision test. The effectiveness of AHR originally developed to reduce neck injuries was quantitatively evaluated through a set of computer simulations. Furthermore, in order to improve the performance of the AHR seat, effects of the seat parameters such as the seat joint stiffness, the headrest position, and the viscoelastic parameters of the seat surface were investigated.

Key words : Neck Injury, Active Headrest, Mechanical Impedance, Rear-End Collision, Euro NCAP

1. Introduction

Rear-end collisions accounted for about 30 % of all vehicle accidents (about 800,000) in Japan in 2007⁽¹⁾; about 45 % of individuals hit in such accidents suffered whiplash and spinal cord injuries (referred to in general as neck injuries)⁽²⁾. Accordingly, there is an increasing need for a car seat that is effective in reducing neck injuries to produce safer automobiles.

Many of recent studies have tried to elucidate the neck injury mechanism using computer simulation techniques on the basis of several experimental evidences^{(3)–(6)}, and they can be roughly classified into the studies using multi-body dynamics^{(7)–(11)} and the studies using the finite element method (FEM)^{(12)–(19)}. Garcia et al.⁽⁷⁾, for example, developed a four-segment dynamic model of the cervical spine and head in the sagittal plane, and analyzed the form of the cervical spine according to impact speeds. Himmetoglu et al.⁽¹¹⁾ created a multi-body human model, and demonstrated that the human model can enough reproduce the rear-impact response of a 50th-percentile male. However, these previous studies based on multi-body dynamics did

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not consider the interactive effects between human driver and car seat in rear-end impacts although such interactions would be important in designing seat parameters. Meanwhile, Furusu⁽¹²⁾ created a model of the human cervical vertebrae using the FEM and examined the correlation between cervical vertebrae injury and shear force along their sagittal axis. Kirkpatrick et al.⁽¹³⁾ constructed a computer model of a human driver using LS-DYNA (a multipurpose analysis tool based on FEM) and ran various situations of vehicle collisions. Stahlschmidt et al.^{(14)–(16)} also presented a new validation test setup using the BioRID II dummy modeled by FEM, and demonstrated that the BioRID II model can create the behavior of the actual dummy in the impact test. Huang et al.^{(18),(19)} then created a FEM model of a human and a seat, and investigated human dynamic behavior in relation to the seatback joint angle as well as the influence on the cervical part according to the headrest position. However, the FEM based method takes a long time in simulation, and requires extensive parameter inputs although it will be extremely difficult to determine them accurately.

On the other hand, in the very recent year, an active headrest (AHR) has been proposed as an effective safety device for reducing neck injuries, in which the headrest will move to the driver's head before the neck is bending in rear-end collisions. Some practical mechanisms have been developed by trial and error adjustments with collision tests, and the effectiveness of AHR seats was mainly discussed by experimental tests^{(20),(21)} although to repeat experiments requires a lot of time and cost. For such the problem, a few computational researches using a FEM model have been tried to analyze the effectiveness of AHR seats^{(22),(23)}, but under the limited conditions in a low speed rear impact. There have still been the strong need of a computer-aided-design tool for developing more effective AHR mechanisms with significant reliability, and thus, this research paper aims to develop a computer simulator using a human-seat model based on mechanical impedance to evaluate an original AHR mechanism.

This paper is organized as follows: Section 2 outlines the developed simulator consisting of a human-seat model created in the framework of multi-body dynamics. Section 3 describes measurement experiments to estimate viscoelastic parameters of seat-surface, and explains the regulation step so that the developed simulator can reproduce dynamic behaviors of human and seat observed in an actual rear-end collision test with the BioRID II dummy. Section 4 verifies the effectiveness of the developed simulator as well as that of the proposed original AHR mechanism, and declares the most appropriate headrest position to reduce neck injuries by the AHR.

2. Impedance Model-Based Simulator for neck injuries

2.1. Outline of the simulator using a human-seat model

Figure 1(a) outlines the flow in evaluating a car seat against neck injuries by the developed simulator. The simulator can provide dynamic behaviors of dummy and seat in rear-end impact test on the sagittal plane, and can evaluate the risk of neck injuries by using the 7 criteria in line with the specifications of Euro-NCAP: 1) $NIC^{(4)}$, 2) $Nkm^{(5)}$, 3) the acceleration of the first thoracic vertebra (T1) in the x direction Acc_{T1} , 4) the velocity of the head relative to the sled in the x direction $V_{Rebound}$, 5) the shear force of the upper neck in the x direction F_x , 6) the tension force of the upper neck in the z direction F_z , and 7) the contact time of head to headrest T_{HRC} . Note that the simulator has a database of the physical parameters of a human-seat model, such as the stature and the weight of a human driver, the size and the weight of seat parts, and the impedance properties of driver's skin and seat surface, which had been measured or estimated in advance.

Figure 1(b) shows a schematic representation of the human-seat model in which the interaction between human and seat is expressed by setting virtual connections between the backs of head and torso and the surfaces of headrest and seatback. The

Fig. 1 The developed simulation system based on mechanical impedance

human model is expressed by 14 rigid links with 14 DOFs; 2 prismatic joints and 6 rotational joints (h_1 : the hip joint, h_2 : the 5th lumbar vertebrae joint, h_3 : the 1st lumbar vertebrae joint, h_4 : the 7th cervical vertebrae joint, h_5 : the 1st cervical vertebrae joint, h_6 : the shoulder joint). Note that the neck is expressed by 6 vertebral disks with a beam model based on the Timoshenko beam theory⁽²⁵⁾. The model of an AHR seat is expressed by 4 rigid links with 4 DOFs, which has one prismatic joint and 3 rotational joints (s_1 : the joint between bearing surface and seatback, s_2 : the joint between seatback and headrest, s_3 : the joint in the AHR mechanism). The AHR mechanism utilized in this study holds up the driver's head and reduces neck injuries by moving the headrest forward right after the rear-end collision as shown in Fig.1(c).

The contour of the human-seat model is created based on a picture of the dummy and seat in a rear-end impact test as shown in Fig.2(a), and the whole model is constructed on a computer simulator by ADAMS (General-purpose mechanical analysis software, MSC Software) as presented in Fig.2(b).

2.2. Formulation of human-seat dynamics

Dynamic behavior of the human-seat model in the two-dimensional space is computed with the following dynamic equation as:

$$M_{q_i} \ddot{q}_i + h(q_i, \dot{q}_i) + g(q_i) + B_{q_i} \dot{q}_i + K_{q_i} (q_i - q_{v_i}) = \tau_{ext_i}, \quad (1)$$

Fig. 2 Contours of the human-seat model

where $q_i \in \mathbb{R}^n$ ($i \in h$: human, s : seat) represents the vector of joint-angle and/or translational-position, n is the degree of freedom in the human/seat model, and q_{vi} represents the virtual equilibrium angle/position for the stiffness. $M_{q_i}, B_{q_i}, K_{q_i} \in \mathbb{R}^{n \times n}$ represent the inertia, viscosity, and stiffness matrices, respectively; $h(q_i, \dot{q}_i) \in \mathbb{R}^n$ refers to the centrifugal and coriolis forces; $g(q_i) \in \mathbb{R}^n$ relates to the gravity force; and $\tau_{ext_i} \in \mathbb{R}^n$ is the external force that is caused by vertical interactive forces $F_{ints}^j \in \mathbb{R}^2$ ($j = 1, \dots, m$) at m contact points between the human and the seat.

The interactive force F_{ints}^j applied from the seat to the human at a contact point is given as follows:

$$F_{ints}^j = f_{ints}^j \frac{x_{ch}^j - x_{cs}^j}{x_c^j}, \quad (2)$$

$$f_{ints}^j = \begin{cases} -B_{cs}^j \dot{x}_c^j - K_{cs}^j (x_c^j - x_{cv}^j) & (x_c^j - x_{cv}^j < -d_{skin}^j) \\ -(\frac{B_{cs}^j B_{ch}^j}{B_{cs}^j + B_{ch}^j}) \dot{x}_c^j - (\frac{K_{cs}^j K_{ch}^j}{K_{cs}^j + K_{ch}^j}) (x_c^j - x_{cv}^j) & (-d_{skin}^j \leq x_c^j - x_{cv}^j \leq 0) \\ 0 & (x_c^j - x_{cv}^j > 0), \end{cases} \quad (3)$$

where $B_{ci}^j, K_{ci}^j \in \mathbb{R}^1$ indicate the surface viscosity and stiffness at the j -th contact point, respectively; x_c^j is the distance of the virtual contact points between the point on human rigid link $x_{ch}^j \in \mathbb{R}^2$ and the point on seat's rigid link $x_{cs}^j \in \mathbb{R}^2$; x_{cv}^j is the equilibrium position for K_{ci}^j ; and d_{skin}^j is the thickness of the skin.

The headrest of the original AHR seat does not activate to reduce neck injury until the interactive force F_{act} between the lower torso and the reaction plate in the seatback is under the specified threshold $F_{threshold}^{(26)}$. The headrest moves forward to the head at high speed just after the force exceeds the threshold. Thus the resistant torque at the AHR-joint τ_{act} can be calculated as

$$\tau_{act} = \begin{cases} -l \times F_{act} & (F_{act} > F_{threshold}) \\ 0 & (F_{act} < F_{threshold}), \end{cases} \quad (4)$$

where l is the distance between the reaction plate and the AHR-joint. The maximum angular displacement of the AHR joint is set at 30 [deg] ($0 < s_3 < 30$) based on experimental results with some trial car seats, and the value of F_{act} was obtained by the sensor-function in the simulator by ADAMS.

3. Measurement experiments for setting the model parameters

3.1. Loading tests for viscoelastic estimation of a car seat

Dynamics of the seat surface to a normal force $dF_{cs}(t)$ can be expressed by

$$B_{cs}(dX)d\dot{X}(t) + K_{cs}(dX)dX(t) = -dF_{cs}(t), \quad (5)$$

Fig. 3 Schematic illustration of the loading test for the seatback surface

where $dX(t)$ denotes the displacement of the seat surface from onset time; $B_{cs}(dX)$ and $K_{cs}(dX)$ represent the seat viscosity and stiffness parameters. The viscoelasticity was modeled by the following nonlinear polynomial expressions with respect to the displacement dX as

$$K_{cs}(dX) = \sum_{p=0}^5 a_p dX^p, \quad B_{cs}(dX) = \sum_{q=0}^5 b_q dX^q, \quad (6)$$

where the parameters a_p and b_q were determined by a non-linear fitting method with the measured data via two experimental tests; a quasi-static test and a dynamic loading test.

Figure 3(a) shows a schematic illustration of the experimental set-up for the quasi-static loading test. The table on which the seat is fixed moves upwards toward the hemispheric rigid load (mass: 6.8 [kg], diameter: $\phi = 165$ [mm]) attached to the tip of the spring (rate of spring k_{spring} : 3.16 [kgf/mm]) at a constant speed of 4.5 [mm/s] to deform the seat surface by the rigid load. The displacement of the table and the spring (dX_{table} and dX_{spring}) is measured by a strain gauge displacement sensor (sampling frequency: 1 [kHz]) to calculate dX and dF_{cs} by

$$\begin{cases} dX(t) = dX_{table}(t) - dX_{spring}(t) \\ dF_{cs}(t) = k_{spring} dX_{spring}(t). \end{cases} \quad (7)$$

Since a rigid load deliberately presses the seat surface under $d\dot{X} \approx 0$ in the quasi-static test, the viscous term in Eq.(5) can be omitted as

$$K_{cs}(dX) dX(t) = -dF_{cs}(t). \quad (8)$$

Fig. 4 Examples of force-deformation curves measured by the loading tests

Fig. 5 Viscoelastic properties of the seat

Accordingly, the stiffness $K_{cs}(dX)$ can be estimated by fitting the displacement dX and the external force dF_{cs} into Eq.(8).

While Fig. 3(b) shows a schematic illustration of the experimental set-up for the dynamic loading test. The hemispheric rigid load is attached to the tip of the movable frame of the testing machine, and falls freely to the seat fixed to the table from 300 [mm] above. A marker on the rigid load is observed using a high-speed camera (sampling rate: 1 [kHz]) to measure the displacement of the seat surface dX during deformations. In the dynamic loading test, the rigid load with a sufficient large mass M_0 falls freely and deforms the seat surface. The dynamic equation of the rigid load is then expressed as

$$F_{cs}(t) = M_0 \ddot{X}(t) + M_0 g, \quad (9)$$

where $\ddot{X}(t)$ represents the acceleration of the rigid load after contact with the seat surface. That is, the force applied to the seat by the rigid load F_{cs} can be calculated by Eq.(9). The viscosity $B_{cs}(dX)$ can be estimated by fitting the displacement dX and the calculated force dF_{cs} into Eq.(5) with the stiffness $K_{cs}(dX)$ that have already be estimated in the quasi-static loading test.

In this paper, such viscoelastic parameters of seat surface for computer simulation were estimated at 4 point as illustrated in Fig.3(c); the upper, middle, and lower parts on the halfway line of the seatback and the central part of a headrest. Therefore, the number of contact points was set at $m = 4$ in the constructed human-seat model.

Fig. 6 Time profiles of the sled acceleration for Euro-NCAP criteria

Figure 4 shows the force-displacement curves measured at the seatback middle in the quasi-static test for estimating the seat stiffness and the dynamic loading test for estimating the seat viscosity. It suggests that differences between the curves depend on the effect of seat viscosity. The seat stiffness and viscosity estimated at the specified four points are presented in Fig. 5. It can be confirmed that viscoelastic characteristics of seat surface change non-linearly according to the displacement at all the measured points. These non-linear dynamic properties of an actual seat, estimated by the two-step procedure mentioned above, are equipped into the human-seat model for simulation analysis.

3.2. Fine adjustment of unknown physical parameters with rear-end collision tests

Figure 6(a) shows the experimental apparatus of a rear-end impact test, which can reproduce dynamic behavior of the seat and the dummy in an actual rear-end accident. A seat with the AHR mechanism and a dummy (BioRID-II, Denton ATD) are set on the sled of a crash simulation system with 1 DOF (Instron Structural Testing Systems). Markers are attached to the joint positions of the seat and the dummy, and the posture of each is measured using Image Express Motion (Sensors Applications). The forces acting on the dummy are measured using built-in force sensors. The impact test following Euro-NCAP protocol adopts the different three levels of sled-acceleration input severity (low, medium and high) as show in Fig.6(b). Acceleration toward the x -axis was induced to the sled, in which the solid line represents an example of acceleration profile measured. Initial postures of the seat and the dummy were set based on Euro-NCAP protocol.

The simulation test was carried out with inducing the measured acceleration of sled as an input for the human-seat model (See Fig. 6(b)), where the simulation duration was 0.18 [s] with sampling time 0.0001 [s]. The physical and dynamic parameters of the human model were set based on the specification of BioRID-II, and the stiffness and viscosity of human skin were set as $K_{ch} = 366$ [N/m] and $B_{ch} = 1.23$ [Ns/m], respectively, while its thickness was set as $d_{skin} = 0.005$ [m] reported in the literature⁽²⁷⁾. The viscoelasticity of the seat surface was set at the estimated value outlined in the previous section.

Fig. 7 Comparison of experimental and simulated dynamic behavior

The 35 unknown parameters in the human-seat model were determined by the two-step optimization procedure with the measured data and the Euro-NCAP criteria such as NIC , Nkm , and so on. The criterion NIC can be calculated as

$$\begin{cases} NIC(t) = 0.2a_{rel}(t) + (v_{rel}(t))^2 \\ a_{rel}(t) = a_{T_1}(t) - a_{head}(t) \\ v_{rel}(t) = v_{T_1}(t) - v_{head}(t), \end{cases} \quad (10)$$

where $a_{T_1}(t)$ and $a_{head}(t)$ represent the acceleration of the first thoracic vertebra and the head in the x direction, respectively; $v_{T_1}(t)$ and $v_{head}(t)$ are the velocities of the first thoracic vertebra and the head in the x direction, respectively; and $a_{rel}(t)$ and $v_{rel}(t)$ are the relative acceleration and velocity between the first thoracic vertebra and the head. The criterion Nkm can be calculated as

$$\begin{cases} Nkm(t) = \frac{|F_x|}{F_{int}} + \frac{|M_y|}{M_{int}} \\ F_{int} = 845 \text{ [N]} \\ M_{int} = \begin{cases} 88.1 \text{ [Nm]} & (M_y > 0) \\ 47.5 \text{ [Nm]} & (M_y < 0), \end{cases} \end{cases} \quad (11)$$

where F_x represents the shear force of the upper neck in the x direction, M_y represents the sagittal bending moment of the upper neck, and F_{int} and M_{int} represent the intercept values of the shear force and bending moment, respectively.

Fig. 8 Comparison of experimental and simulated results

The first adjustment of parameters was done using the Generalized Reduced Gradient (GRG) Method⁽²⁸⁾ under the following optimization function as

$$J_1 = \sum_{i=1}^{10} u_i(t) \{Val_{mes}^i(t) - Val_{sim}^i(t)\}^2,$$

$$u_i(t) = \begin{cases} 0 & (t < time_{low}^i) \\ c_i & (time_{low}^i \leq t < time_{high}^i) \\ 0 & (t \geq time_{high}^i), \end{cases} \quad (12)$$

where $Val_{mes}^i(t)$ and $Val_{sim}^i(t)$ are measurement data and simulation data in the rear-end impact test such as postures, accelerations, forces, and neck injury criteria of the dummy, respectively; c_i is a weighting factor determined for the i -th data, $time_{low}^i$ and $time_{high}^i$ are lower limit and higher limit of the weighting term, respectively. As minimizing the function J_1 , it is expected that the simulator reproduces the dynamic behavior of the dummy and seat observed in the rear-impact test. Note that initial values were set by manual to make the behavior of the human model similar to that of the dummy in appearance.

The fine adjustment was then conducted using Genetic Algorithm (Global Opti-

Fig. 9 The Euro-NCAP calculation rule for neck injury criteria

mization Toolbox, Mathworks Inc.) with the fitness function given by

$$J_2 = \sum_{i=1}^{11} c_i \left(\frac{Val_{mes}^i(t) - Val_{sim}^i(t)}{\max\{Val_{mes}^i(t)\} - \min\{Val_{mes}^i(t)\}} \right)^2, \quad (13)$$

$$c_i = \begin{cases} 1 & (1 \leq i \leq 5 : \text{for angular data}) \\ 5 & (6 \leq i \leq 11 : \text{for index data}). \end{cases}$$

The GA parameters set in this study were as follows; Population number: 30, Bit length: 8, Crossover rate: 90 %, Mutation rate: 1%, Repetition time: 300. The search range for each unknown parameter of the human-seat model was set based on the value obtained via the rough adjustment by GRG.

Figure 7 shows the experimental and simulated dynamic behavior at 0.00, 0.06, 0.12 and 0.18 [s] from the top down. The simulated postures of human and seat well agree with the captured ones by the high-speed camera at each time. The seatback turns backward as the time goes on, while the head throws back and contacts to the headrest. Fig. 8(a) shows time histories of headrest angle, seatback angle, head angle, neck angle and torso angle. It should be noted that each angle represents that from the x -axis in the world coordinate system. Solid lines represent simulated results, and broken lines represent experimental results; it can be seen that the two almost coincide. Fig. 8(b) shows time histories of the neck injury criteria that were used in the parameter adjustments. The solid lines represent the simulation results, and the broken lines represent the experimental results; it can be seen that the two almost coincide except for the upper neck tension force. The headrest contact time is 0.064 [s] in the experiment and 0.0679 [s] in the simulation. These results demonstrate that the proposed model is capable of reproducing the measured movement of the dummy and seat in rear-end impact testing.

4. Computer simulations for the AHR validation

4.1. Effectiveness of the AHR mechanism

The effectiveness of the AHR mechanism was verified using seven criteria in line with the specifications of Euro-NCAP; NIC , Nkm , Acc_{T1} , $V_{Rebound}$, Fx , Fz , and the time of first contact between rear of the head and the headrest T_{HRC} . The maximum values of these criteria are then converted from 0-point to 0.5-point along the specified rule as shown in Fig. 9, where higher point indicates better ability for reducing neck injuries. Simulation analyses were performed for a seat with and without the AHR mechanism according to the three controlled acceleration inputs, i.e., Low severity, Medium severity, and High severity shown in Fig. 6(b). The duration of the simulations was 0.22 [s], and the sampling time for dynamic calculation was 0.0001 [s].

Figure 10 shows the simulation results with and without the AHR. The figure (a) presents the time histories of headrest angle, seatback angle, head angle, neck angle and torso angle. The angular difference related to the AHR mechanism is clearly found in the time profiles of headrest angle after 0.04 [s] around. The figure (b) demonstrates time histories of the interaction force between body and seatback and the resistant torque that was utilized to actuate the AHR mechanism instantaneously.

(a) Joint angles

(b) Interaction force and torque

(c) Neck injury criteria

Fig. 10 Time histories of neck injury criteria with/without the AHR (Medium severity)

Fig. 11 Scores for neck injury criteria with/without the AHR

The resistant torque was suddenly reduced at 0.04 [s] around because the magnitude of the interaction force was over the threshold, and the AHR-mechanism was activated so that the headrest moved forward and contacted with the back of the head to reduce neck injuries. The figure(c) shows time histories of the neck injury criteria for the medium severity, where the top/bottom of gray areas represent the higher/lower performance limit for each evaluation data and the peaks of waves are marked with a black triangle. Note that these time-series data are displayed until the end of headrest contact, which is defined as the time at which the head first loses contact with the headrest⁽²⁴⁾.

Fig. 11 shows the neck injury criteria scores for the seat with and without the AHR according to the acceleration inputs. The scores of the AHR seat are higher (better) or equal than those of the non-AHR seat except for the scores of *NIC* under conditions of high severity, *Nkm* under conditions of medium severity, and upper neck shear force *Fx*. In particular, there is a marked difference in scores of *NIC*, *T_{HRC}*, and *V_{Rebound}* between the seats with and without the AHR. It can be confirmed that the average scores for the AHR seat are higher than that for the non-AHR one. However, the average score for the seat with the AHR under conditions of high severity is lower than that for the same seat under conditions of low severity and medium severity. The performance of the AHR seat may improve if seat parameters could be regulated properly.

4.2. Influence of seat parameters into neck injury

Some experimental studies using non-AHR seats reports indicate that the joint stiffness between the bearing surface and the seatback and the headrest position affects the risk of neck injury in some degree⁽⁶⁾. On the basis of these empirical evidences, the performance of the AHR seat was analyzed according to the seat joint stiffness, the headrest position, and the viscoelastic properties of seat surfaces.

Figure 11 also shows the simulation results for neck injury criteria scores according to the joint stiffness and the severity of the acceleration input, in which the standard value of joint stiffness is 142.6 [Nm/deg]. The average scores show the performance

Fig. 12 Scores for neck injury criteria according to the headrest position (180 [Nm/deg])

becomes best at 180 [Nm/deg]; this is because the force is easily transmitted to the reaction plate of the AHR mechanism. However, the average scores for high severity are still much lower than those for the others.

In order to further improve the performance of the AHR seat, an effect of the headrest position was investigated with fixing the seat joint stiffness at 180 [Nm/deg]. The headrest was moved against the head position as shown in Fig.12(a). The origin of the headrest position (the white circle) was defined as vertical axis was the distance from the top of the head in a vertical line to the top-front point on the headrest and horizontal axis was the distance from the top of the occipital on a level line to a point on the front of the headrest on a level line. Fig. 12(b) shows the data maps of average scores of the neck injury criteria depending on the headrest position. The quadrangular marker indicates the basic position of the headrest, and the double circle in the figures indicates the best position (x -axis = 40 [mm] and z -axis = 0 [mm]) obtained the highest average score at medium severity. White circles indicate the average scores of the seven criteria obtained at the simulation, and the data between the white circles are calculated by cubic spline interpolation based on Delaunay triangulation. The scores are improved when the vertical position of the headrest from the head is heightened and the horizontal distance from the head to the headrest is decreased. Fig. 12(c) shows the score bars under the conditions of the headrest positions at the

Fig. 13 Scores for neck injury criteria according to the headrest position (180 [Nm/deg])

basic position and the double circle position. It can be seen that the average scores are improved by regulating the headrest position for all conditions, especially for the high severity.

Finally, viscoelastic properties of the seat surface presented in Fig. 5 were optimized for the improvement of seat performance by GRG, in which the magnitudes of stiffness functions and viscosity functions were regulated to maximize the average score for neck injury criteria along the Euro-NCAP calculation rule. Note that other seat parameters were same with the data obtained through the fine adjustment i.e.; seat joint stiffness was at 142.6 [Nm/deg] and the headrest position at the basic position. Fig. 13(a) shows the optimized amplitudes of the viscoelastic function of seat surface according to the seat part, while the figure (b) shows the scores of the neck injury criteria before and after optimization according to the sled acceleration. It should be noticed that the scores for the high severity condition is much improved by the optimization of seat surface visco-elasticity. The results reveal that the material of seat cushions should be carefully determined in the development of an effective seat against neck injuries. In this manner, the proposed simulator can assist to improve and design an effective AHR seat to reduce neck injuries.

5. Conclusion

The simulator using human-seat model had been developed based on mechanical impedance for analyzing the risk of neck injuries. The model with the AHR mechanism

enabled the recreation of dummy and seat movement in rear-end impact testing to some extent, and allowed the risk of neck injuries to be evaluated using seven criteria types. The simulation experiments performed in the study indicated that the AHR mechanism is effective in reducing such injuries. Future research will be coordinated further to assess the behavior between the human-seat model and the dummy in rear-end impact testing. The relationship between human body size and the risk of neck injury will also be analyzed, and a new mechanism will be designed with the aim of reducing neck injuries.

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