Finite Element Analysis for Investigating the Effects of Muscle Activation on Head-neck Injury Risks of Drivers Rear-ended by a Car after an Autonomous Emergency Braking

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ABSTRACT: Avoidance of frontal collisions by autonomous emergency braking (AEB) in sudden traffic jams has the potential to cause rear-end collisions by the following cars. In this study, finite element analyses using a human body model were performed to investigate how the muscle activations of drivers rear-ended by the following car could affect head-neck injury risks. Three muscle conditions of sleeping, relaxed, and braced drivers were assumed using a developed muscle controller. The simulation results suggest that vehicle systems that let drivers brace themselves at the onset of the AEB might be effective in reducing head-neck injury risks in this type of collisions.

KEY WORDS: Safety, rear end collision, anthropomorphic dummy, muscle activation, head-neck injury risk [C1]

1. Introduction

In recent years, commercial vehicles worldwide are being equipped with advanced vehicle safety technologies such as autonomous emergency braking (AEB) and lane keeping assist (LKA). This kind of advanced safety technology is effective in reducing the number of vehicular accidents as well as the number of fatalities. For instance, the AEB would be effective in avoiding frontal collision of the AEB-equipped car in sudden traffic jams. However, when the AEB-equipped car is rear-ended by the following car, there is a possibility that the AEB-equipped car’s driver could sustain head-neck injuries, because the driver is impacted from behind since his upper body is already leaning forward due to the deceleration of the AEB.

Eis et al. (2005) found that 59% of the occupants had no injury and 39% had AIS (Abbreviated Injury Scale) 1 injury during single rear impact crashes, in a study involving 1,272 struck vehicle occupants using data obtained from the German In-Depth Accident Study (GIDAS) (2). They also found that most of the AIS 1 injured occupants with soft tissue neck injury sustained the injuries from crashes with struck vehicle of delta-V less than or equal to 15km/h. Hell et al. (1999) reported on a comparison of the injury patterns with the injury costs in car to car accidents of 28 cars with 41 belted occupants, head injuries predominated at 31.7%, followed by thorax injuries at 17.1% and abdominal injuries at 14.6% for AIS2+ injuries. They also showed that the highest costs involve the head at 9,000 Euro followed by the pelvis at 6,000 Euro and abdomen at 3,000 Euro in rear-end collisions. These studies suggest that AIS1 injury with mostly soft tissue neck injury is predominant in struck vehicle occupants involved in rear-end collisions, while head injury is predominant in both frequency and costs among AIS2+ injury, which could occur with higher impact velocities of more than 15 km/h.

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2. Method

2.1. Human body FE model for the prediction of occupant kinematics and injury outcomes

2.1.1. Human body FE model

In our previous studies, we developed a human body FE model called THUMS (Total HUman Model for Safety) Version 5, which includes active muscles of the entire body except the face (3). Figure 1 shows an outline of THUMS Version 5 AM50 (American Male 50 percentile) model. The body size of THUMS is close to that of an American mid-size adult male (height of 175
from the test data predicted by THUMS were almost the same as those obtained from Ono et al. (1997) \(^{(7)}\). Comparisons of the head rotational angle time history curves for rear impact between THUMS and volunteer test data of a rear impact at 8 km/h, obtained from White et al. (2009) \(^{(6)}\), and volunteer test data on a rear impact at the same velocity, obtained from Ono et al. (1997) \(^{(7)}\). Comparisons of the head rotational angle time history curves for rear impact between THUMS and individual muscles of the entire body \(^{(4)}\).

As for the rear-end impacts, THUMS was validated against cadaver test data of a rear impact at 8 km/h, obtained from White et al. (2009) \(^{(6)}\), and volunteer test data on a rear impact at the same velocity, obtained from Ono et al. (1997) \(^{(7)}\). Comparisons of the head rotational angle time history curves for rear impact between THUMS and the test data indicated that the peak head rotational angles predicted by THUMS were almost as those obtained from the test data \(^{(8)}\).

In our previous studies, we developed a muscle controller to simulate the muscle activation patterns in this study. In addition, the THUMS includes 262 muscle bar elements modeled by Hill type muscle contractile material properties, and can reproduce the driver’s muscle activation conditions by inputting the activation time history curves to individual muscles of the entire body \(^{(4)}\).

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In force control, the controller calculates the control signal \( v_h \) based on the errors \( e_h(N) \), which is the difference between the current reaction force and the target reaction force at the steering or brake pedal using the following equation:

\[
r_h(t) = K_{p,h} \cdot e_h(t) + K_{i,h} \int_0^t e_h(\tau) d\tau + K_{d,h} \frac{d e_h(t)}{dt}
\]  

(3)

where the suffix \( h \) (=1, ..., 4) represents each reaction force exerted by the hand or foot on the right or left side and \( K_{p,n} \) (N\(^{-1}\)), \( K_{i,n} \) (N\(^{-1}\) s\(^{-1}\)), and \( K_{d,n} \) (s/N) are the PID gains for the force control. In this study, the target reaction forces were set to 300 N for each hand and 400 N for each foot, based on the forces measured from volunteer test data on braced occupants\(^{(1)}\). Using \( v_h \) obtained from equation (3), the muscle activation level \( a_i \) of each muscle for the force control is determined by the following equation:

\[
a_i(t) = \sum_h P_{ih} \cdot v_h(t)
\]  

(4)

where \( P_{ih} \) are the percentage contributions of each muscle to the pushing force, which were set according to the information about the role of each muscle, as described in anatomical texts\(^{(10)}\) and the measured EMG (Electromyography) data from volunteer tests in a braced position, which was conducted in our previous study\(^{(12)}\). For example, the percentage contributions of the right Biceps Femoris (Long Head) and right Gluteus Maximus for the pushing force on the right brake pedal are 0.3 and 0.1, respectively.

The muscle activation levels for the relaxed drivers are predicted using equation (2), while those for the braced drivers are predicted using the following equation:

\[
a_{total}(t) = a_{p}(t) + a_{i}(t)
\]  

(5).

2.2. Simulation setup

The purpose of this study is to investigate how the muscle activations of drivers could affect head-neck injury risks of the drivers rear-ended by the following car after an AEB. Therefore, we simulated a crash event in which the drivers sustained rear-ended impacts after deceleration change resulting from an AEB, using a human body FE model (THUMS Version 5 AM50) and a vehicle sled FE model. Figure 3 shows the simulation setup for a rear-end collision after an AEB. THUMS was seated with a 3-point seatbelt on the automotive seat of the vehicle sled FE model of a Ford Taurus vehicle body, which was originally developed by the National Crash Analysis Center of the George Washington University under contract with the Federal Highway Administration and the NHTSA and was modified by JSOL Corporation. As shown in Figure 3, acceleration time history curves were applied to the vehicle sled model in the following sequential order: a gravity of 1.0G for making a sitting posture, a deceleration of 0.8G for simulating an AEB, and rear-end impact decelerations corresponding to 16 km/h or 24 km/h. The deceleration curve of 0.8G for simulating an AEB was obtained from a volunteer test for reproducing an AEB that was conducted by Ejima et al. (2008)\(^{(13)}\). The deceleration curves of rear-end impacts corresponding to 16 km/h or 24 km/h were obtained from two series of cadaver tests for reproducing low-speed rear-ended impacts that were conducted by White et al. (2009)\(^{(10)}\) and Kang et al. (2012)\(^{(14)}\), respectively. We performed the simulations without muscle activation to simulate sleeping drivers, with posture control to simulate relaxed drivers, and with total control of posture and force to simulate braced drivers.

3. Results

The following sub-sections describe the simulation results regarding the effects of muscle activations and impact velocities on driver’s kinematics, brain injury outcomes, and neck injury outcomes.

3.1. Effect of driver conditions on driver’s kinematics

Figures 4 and 5 show comparisons of the predicted driver’s kinematics among three muscle activation conditions simulating sleeping, relaxed, and braced drivers in rear-end collisions at 16 km/h and 24 km/h after an AEB, respectively. The driver postures after the AEB were different for the three driver conditions. In the case of 16 km/h, the relaxed driver leaned forward while the braced driver held back his upper body. At 250 ms after the collision, the head of the relaxed driver had a larger backward rotation, while the head of the braced driver was almost maintained at the original posture. Until 25 ms after collision, the kinematics of the sleeping driver were similar to those of the relaxed driver. However, at 250 ms after collision, the head of the sleeping driver moved downward with a backward rotation of the torso. In the case of 24 km/h, the relaxed driver leaned forward similar to the sleeping driver, while the braced driver held back his upper body in comparison with posture of the relaxed driver, at 25 ms after collision. In addition, the head positions in the case of 24 km/h were supposed to move backward in comparison with those in the case of 16 km/h, 180 ms after collision. Figure 6 shows the comparisons of translational accelerations of the driver’s head among three driver conditions. The highest peak translational accelerations were found in the braced driver at 16 km/h and in the sleeping driver at 24 km/h. Figure 7 shows the comparisons of rotational acceleration of the driver’s head among three driver conditions. The highest peak rotational accelerations were found in the braced driver at 16 km/h and in the sleeping driver at 24 km/h. At 24 km/h, the peak translational and rotational accelerations of the relaxed driver were higher than those of the braced driver.
Fig. 4 Predicted driver’s kinematics in the case of rear-end collision at 16 km/h following AEB.

Fig. 5 Predicted driver’s kinematics in the case of rear-end collision at 24 km/h following AEB.

3.2. Effects of driver conditions on brain injury outcomes

There are several brain injury predictors that use a head/brain FE model. Among the predictors, we selected the maximum 1st principal strain (hereinafter, referred to as MPS), the maximum shear strain, and CSDM (Cumulative Strain Damage Measure) 10% and CSDM 15% to predict the brain injury outcomes. This is because these predictors had high correlation coefficients with RIC (Rotational Injury Criterion), which was proposed as one of the MTBI (mild traumatic brain injury) criteria in our previous study, and showed high correlation with head rotational acceleration data, for both MTBI and non-MTBI, obtained from American football players (15). The CSDM is defined as the percentage volume of the brain that exceeds a specified MPS threshold. The CSDM where the MPS threshold is set to 10% is expressed as “CSDM 10%” in this study. Figure 8 shows the comparisons of the maximum 1st principal strain and maximum shear strain among the three driver conditions for 16 km/h and 24 km/h. The braced drivers tend to have less strain in the brain than the relaxed drivers. At 24 km/h, the maximum 1st principal strains were 0.2 and 0.17, while the maximum shear strains were 0.24 and 0.18 for the relaxed and braced drivers, respectively. In addition, the maximum strains of the sleeping driver were less than those of the other drivers at 16 km/h; however, they were much higher than the other drivers at 24 km/h. Figure 9 shows the comparisons of the CSDM 10% and CSDM 15% among the three driver conditions for two impact velocities. At 24 km/h, the values of CSDM 10% for the relaxed and braced drivers were 7.8 and 1.8, respectively, while the values of CSDM 15% for the relaxed and braced drivers were 0.6 and 0.0, respectively. In addition, the values of CSDM 10% and CSDM 15% for the sleeping driver were 35.0 and 15.0, respectively.

3.3. Effects of driver conditions on neck injury outcomes

In rear-end collisions, whiplash injuries with an injury scale of AIS 1 are the most frequent among neck injuries. The symptoms of whiplash-associated disorders (WAD) are classified into five grades from 0 to 4, in which the grades 0, 1, 2, 3, and 4 denote no complaint about the neck and no physical signs, neck complaint and musculoskeletal signs, neck complaint and neurological signs, and neck complaint and fracture or dislocation, respectively according to the Quebec WAD classification (16). The whiplash injury mechanism is not clearly elucidated yet; however, the MPS of
both velocities. In particular, the MPS in the sleeping driver was higher than those in the relaxed and braced drivers for impact velocities. In all the cases, the MPS of the facet joint capsule (more than 0.17) was larger than that of the ALL (less than 0.12). The MPS of the facet joint capsule in the sleeping driver was higher than those in the relaxed and braced drivers for both velocities. In particular, the MPS in the sleeping driver was 0.58 at 24 km/h.

Fig. 10 Maximum 1st principal strains predicted at facet joint capsule and ALL of sleeping, relaxed, and braced drivers for 16 km/h and 24 km/h.

4. Discussion

4.1. Assumption of three driver conditions

In this study, we assumed the three following driver conditions: without muscle activation to simulate the sleeping drivers, with posture control to simulate the relaxed drivers, and with total control of posture and force to simulate the braced drivers. Since the reflex reaction of muscles is depressed in Non REM (Rapid Eye Movement) sleep and abolished in REM sleep relative to that in wakefulness, the sleeping drivers that fell asleep could be modeled without muscle activation. However, actual human body has the digestive system (e.g. esophagus, stomach), respiratory system (e.g. trachea, lung), and circulatory system (aorta, heart). The reason why the simulation results for the sleeping drivers at 24 km/h (see Figures 6 to 10) showed the highest peak translational and rotational accelerations and the highest brain and neck injury predictors is because the THUMS Version 5 AM50 model does not include structures of the trachea and esophagus in the neck region (Figure 11 (a)(b)), although it has the structures of the trachea and esophagus below the neck. Therefore, the jaw of the sleeping drivers dug into the skin of the lower neck region, and therefore the neck region showed hyperflexion unlike an actual human body. This problem is due to the modeling of the neck region. Further studies are needed to model the trachea and esophagus, including pressures of the air or some gasses, to obtain more reasonable simulation results for the sleeping drivers. In the following discussion regarding the effects of driver conditions on injury outcomes, only comparisons of the relaxed drivers and the braced drivers were considered.

When the human body maintains wakefulness during driving, the reflex reaction of muscles would work. Thus, the relaxed drivers who do not detect the possibility of a rear-end crash could be modeled using the posture control that determines the activation level of each muscle to return to their initial postures as reflex reaction. On the other hand, the braced drivers who detect the possibility of a rear-end crash could be modeled using the total control of posture and force, because such drivers perform reflex reaction due to the wakefulness and protective response with voluntary muscular forces simultaneously.

A comparison of the maximum strains in the brains between the relaxed drivers and the braced drivers indicated that the braced drivers tend to have less strain in the brains than the relaxed drivers (see Figure 8). A comparison of the CSDM 10% and CSDM 15% also indicated that the braced drivers tend to have less brain injury than the relaxed drivers at 24 km/h (see Figure 9). There are significant differences in the CSDM 10% between the relaxed drivers and the braced drivers. According to the classification defined by Maxwell et al. (1997), axonal injuries can be categorized into two severities of primary axotomy and secondary axotomy. The primary axotomy indicates severe damage of an axon filament associated with strains of more than 20%. The secondary axotomy has several stages as follows: focal loss of axonal transport associated with 5-10% strain, axonal swelling associated with 10-15% strain, and axonal bulbs associated with 15-20% strain. The relaxed drivers have a tendency to increase the CSDM 10%, which could be related to the secondary axotomy, and then cause MTBI. As mentioned previously, the brain injury predictors of the MPS, the maximum shear strain, CSDM 10%, and CSDM 15% were strongly correlated with the RIC, which is one of the rotational head injury criteria and can be calculated from head rotational accelerations. The peak head rotational acceleration of the braced driver was half as high as that of the relaxed driver at 24 km/h. This suggests that the braced condition could reduce the head rotational acceleration during rear-end collisions after the AEB and result in reduced risk of the brain injury. However this is a kind of qualitative speculation; therefore, further studies using more detailed brain FE model are needed to understand the effect of muscle activations on the brain injury outcomes.

Comparisons of the MPS at facet joint capsule and ALL between the relaxed drivers and braced drivers indicated that the MPS of facet joint capsule was significantly larger than that of other cervical ligaments of ALL and PLL (see Figure 10). The failure threshold of the facet joint capsule was obtained from tensile test data using human surrogate specimens. Since subcatastrophic failure strains ranged from 14% to 93% (21)(22), the MPS predicted in the relaxed and braced drivers has a possibility to exceed the threshold of subcatastrophic failure strain. It is noted that the MPS of braced drivers tend to be lower than that of the relaxed drivers at the lower speed of 16 km/h. This is probably because the rotational displacement of the head-neck, relative to the torso, in a relaxed driver was larger than that in a braced driver at 16 km/h. These results suggest that the braced condition could reduce head-neck injury risks by decreasing the rotation of the head-neck relative to the torso during rear-end collisions after the AEB.
5. Conclusion

This study investigated the effect of muscle activations on head-neck injury risks of drivers rear-ended by the following car after an AEB, using a human body FE model (THUMS Version 5 AM50). Simulations using THUMS were performed with a vehicle sled FE model to reproduce rear-end collisions at 16 km/h and 24 km/h after an AEB with deceleration of 0.8G. Three types of muscle activations to simulate sleeping, relaxed, and braced drivers were predicted such as without muscle activation, with posture control, and with total control of posture and force, respectively. The simulation results demonstrated that muscle activations tend to change the driver’s kinematics and alter injury outcomes, and that relaxed drivers could sustain more brain injury and neck injury than braced drivers. The results suggest that vehicle systems that let drivers brace themselves at the onset of the AEB might be effective in reducing head-neck injury risks in the unexpected event of rear-end collisions.

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