

INVESTIGATION OF PRE-IMPACT BRACING EFFECTS FOR INJURY OUTCOME USING AN ACTIVE HUMAN FE MODEL WITH 3D GEOMETRY OF MUSCLES

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ABSTRACT

Accident data analyses conducted at the Institute for Traffic Accident Research and Data Analysis (ITARDA) in Japan reported that over 60% of drivers who faced unavoidable crash situations made evasive maneuvers on braking and steering in 2007. In such emergency cases, drivers also might brace their body with their muscle activity to prepare the upcoming impacts. Their muscle activity would not only generate muscular forces but also change muscular stiffness and mechanical properties of their articulated joints. Therefore, occupant behaviors during impacts could be different from those observed in dummy tests and cadaver tests.

In this study, we developed an active human finite element (FE) model with 3D geometry of muscles. The muscle was modeled as a hybrid model by combination of bar elements with active muscle properties and solid elements with passive muscle properties. The bar elements were modeled with a Hill type muscle model to generate muscular force according to inputted activation levels. The solid elements were modeled with a rubber-like material model to simulate 3D geometry of individual muscles and non-linear passive properties. This combined muscle model was validated against human volunteer test data and reproduced increase of muscular stiffness with increase of muscle activation level as observed in the tests.

A volunteer test with one healthy male subject was conducted to obtain physiological information in a bracing situation with braking under his informed consent based on the Helsinki Declaration. In this test, the subject was asked to push his right foot on a brake pedal and his hands on a steering with his maximal voluntary force in the test apparatus fixed on the laboratory. Besides three reaction forces of a brake pedal, a steering, and rigid flat seats, the posture, pressure distribution on the seats, and 24 surface EMG (Electromyography) signals during his braking motion were measured in this test. His maximal braking force was reached to 750N and was well matched to previously reported values for emergency braking situation.

We performed simulations using the active human model to reproduce the bracing condition. In the

simulations, the activation levels of 24 muscles obtained from the EMG data were directly inputted to the corresponding muscles of the active human model and those of the other muscles were estimated to reproduce the reaction forces. After reconstructing the reaction forces for the braced volunteer, we performed frontal impact simulations to compare occupant behavior and injury outcome in an active human body with those in a cadaveric human body. The simulation results showed significant differences between both human bodies. Different from the cadaveric human body, the active human body could have less injury risks in the thorax and more in extremities. These injury outcomes correspond to those previously reported in comparison between real-world accidents and laboratory cadaver sled tests. Although the active human model has some limitations on accuracy of estimation of muscular activation levels due to lack of consideration for muscular reflex and posture stabilization, it could have possibility to evaluate injury outcome in real-world accidents.

INTRODUCTION

Recent accident data analyses indicate that thorax injuries and lower extremity injuries are still important to mitigate occupant injuries in frontal crashes. Carroll et al. (2010)[10] conducted accident data analyses using the UK Cooperative Crash Injury Study (CCIS), the German In-Depth Accident data Survey (GIDAS), and the French GIE RE PR (Renault, and PSA Peugeot Citroen) database. They reported that older occupants are likely to sustain more torso injury. They also reported that occupants seated in the front passenger seat tended to sustain more torso injuries compared with the driver's seat. Brumbelow et al. (2009)[8] investigated impact and injury patterns in frontal crashes of vehicles based on the NASS-CDS crash data. They showed that occupants 60 or older more often received at least one serious chest injury than a serious head injury and the opposite was true for occupants younger than 30. Simamura et al. (2003)[35] conducted accident data analyses of a total 246 vehicle occupants using ITARDA data. They reported that elderly occupants frequently experienced rib fractures near the seatbelt

line even under lower impact severity while younger occupants appeared not to sustain rib fractures even in higher impact collisions. Their accident data analyses on thoracic injury indicate that occupant injury outcomes are different between the driver's seat and the front passenger seat as well as occupant injury outcomes are different between younger occupants and older occupants. However, the injury mechanisms are not well known.

As for the lower extremity injuries, Rudd (2009)[32] investigated lower limb injury risk and causation in the NASS-CDS crash database with mean age of about 38-year-old. They reported that foot and ankle injury prevalence has not decreased in newer model-year vehicles, and that injury risk to the foot and ankle has actually increased despite structural improvements aimed at reduced footwell deformation. They also reported that the majority of the foot and ankle injuries occur at lower crash severities with delta-V of less than 30km/h. This study show that the foot and ankle injuries occurred for even younger occupants and lower speed of impacts. However, the injury mechanisms are also not well known. Therefore, it is critical to elucidate the mechanisms for the thoracic injury and the lower extremity injury in order to mitigate occupant injuries in frontal impacts.

In addition, accident data analyses on frontal crashes conducted at ITARDA[1] in Japan reported that over 60% of drivers who faced unavoidable crash situations made evasive maneuvers on braking and steering in 2007. In such emergency cases, drivers also might brace their body with their muscle activity to prepare the upcoming impacts. Their muscle activity would not only generate muscular forces but also change muscular stiffness and mechanical properties of their articulated joints. Therefore, occupant behaviors with their muscle activity during impacts could be different from those observed in dummy tests and cadaver tests.

Several experimental studies have been performed to investigate the effect of muscle activity on injuries under the assumed impact situations. Tennyson and King (1976)[37] conducted a series of neck loading cadaveric tests and reported that muscle tense increased the neck injury in higher acceleration, while muscle tense contributed to the decrease in neck injury probability in low acceleration. This study suggested inherent performance tradeoffs in the role of muscle tense on the injury severities in impact loadings. Begeman et al. (1980)[6] conducted a series of sled test using human volunteers and cadaveric subjects and investigated the effects of muscle tense on kinematics of the lower extremities. As the results, they revealed that muscle tense stiffened up the human body rigidly and alter the overall kinematics of the human body during whole impact events. Funk et al. (2001)[17] performed

cadaveric axial impact tests for the foot and ankle complex under a condition to simulate entrapped knee. In their tests, a foot plate hit the foot axially with an initial velocity of 5 m/s while muscular tension forces of 0 kN, 1.7 kN or 2.6 kN were applied to the Achilles tendon for investigation of muscular effect to skeletal injuries of the lower leg. They found that the muscular tension force can increase axial compressive force and the possibility of bone fractures at the distal tibia. On the other hand, some benefit effects of muscle tense were also reported. Levine et al. (1978)[27] conducted frontal impact tests using cadavers restrained by a three-point belt system equipped with and without a knee brace which simulated muscular tense of quadriceps in thigh. They found that the knee brace played a role in the prevention of submarining in frontal collisions. Therefore, the muscle tense appears to have both aspects of advantage and disadvantage for occupant injuries. However, it is not fully understood how muscle tense affects the impact responses and injury severities.

Computational human models are effective tools to understand the injury mechanisms in automotive crashes. Several researchers developed human whole body FE models of which size is AM50 (American adult male 50%ile) and validated the models against impact responses obtained from existing cadaver test data (Iwamoto et al., 2002[21], Vezin et al., 2005[40], Ruan et al., 2005[31]). Recently, Shigeta et al. (2009)[33] developed much more detailed human FE model including internal organs whose total number of elements is 1.8 million and validated the model against impact responses obtained from several cadaver test data. These human FE model represented mechanical responses of human body during impacts and contributed to elucidate some injury mechanisms in automotive crashes. Since the purposes of developing these models were not to investigate effects of muscle activity on occupant injuries, these models did not include active muscles. Recently, some human FE models have been developed with active muscles to investigate the muscular effects for human body kinematics. Choi et al. (2005)[12] conducted both sled tests using eight volunteer subjects and computational analysis using a human FE model in bracing during frontal impacts. They used the EMG as an indicator of muscle activation levels and normalized it against that of maximal voluntary contraction. Reaction forces on steering wheel and brake pedal predicted by using their FE model with 16 muscles in the upper and lower extremities agreed with those of experimental data. However, muscles in deep layers were not considered in the impact analysis of the human body. Chang et al. (2008)[11] also developed a MADYMO FE model with 35 Hill-type muscles including muscles in deep layers for each lower extremity and

simulated knee-to-knee-bolster impact response in bracing during frontal impacts. Their simulation results with and without different levels of lower-extremity muscle activation for bracing suggested that muscle tension had the potential to decrease the externally applied force required to cause knee-thigh-hip fracture, and had the potential to increase the likelihood of femoral shaft fracture. Since their muscle models were developed using bar elements, their model did not represent the interaction forces between adjacent two muscles and the interaction forces between muscles and the adjacent bones. Behr et al. (2006)[7] developed a FE model of the lower limb with 20 independent muscle bundles in the superficial and deep layers and used to investigate the effect of muscle tense on the skeletal injuries of the lower limb. Each muscle was modeled using coupling of solid elements and actions of fiber elements. The interaction forces between muscles and bones were represented in the model. They validated the model for isolated muscle contribution in the direction of fibers and bracing conditions under an emergency braking. Their simulation results indicated that muscle activation in bracing during frontal impacts significantly increased the stress level on the tibial shaft. Hedenstierna et al. (2007[19], 2008[20]) represented a muscle using a combination of passive non-linear, viscoelastic solid elements and active Hill-type truss elements. They applied the muscle model to 22 separate pair of human neck muscles and conducted kinematical validation against volunteer experiments. They showed strain distribution in each neck muscle in frontal impact and rear-end impact for injury analysis. However, these two models were not validated for muscle stiffness change according to the activity, which is one of the essential characteristic features of muscles in considering muscular responses for impacts.

In this study, we developed a human body FE model with 3-D geometry of individual muscles to represent the interaction between two adjacent muscles and between muscles and the adjacent bones. We also provided the model with capability of muscle contraction to reproduce muscular stiffness change according to the activity. The developed human body FE model was validated against cadaver test data on thoracic and lower extremity responses in frontal impacts. The model was also validated against cadaveric frontal impact sled test data with 50km/h. A bracing situation was selected to investigate effect of muscle activity on injury outcome in frontal impacts. A volunteer test was conducted to obtain EMG data of muscles in the upper and lower

extremity in the bracing situation. The developed human body FE model with muscles reproduced the bracing situation in pre-impact and then sustained frontal impacts with 50km/h. Simulation results of the human body model with muscle activity were compared with those of the human body model without muscle activity. We discussed effects of muscle activity in pre-impact on injury outcomes. We also discussed if muscle activity can explain the unknown mechanisms for thoracic and lower extremity injury. All simulations in this paper were performed using an explicit finite element code LS-DYNA (LSTC, USA).

MODEL DEVELOPMENT AND VALIDATIONS

A human whole body FE model with muscles

Muscular FE models of a human whole body were developed and integrated with a human body FE model called THUMS (Total Human Model for Safety, Iwamoto et al., 2002[21]) whose size was similar to that of AM50 with a height of 175cm and a weight of 77kg. Figure 1 shows a developed human body FE model in a standing posture. In this figure, the skin was removed to see muscles clearly. The model includes 266 muscles of lower extremities, upper extremities, trunk, and neck such as the Sternocleidomastoid, Trapezius, Rectus Abdominis, Erector Spinae, Pectoralis Major, Deltoid, Biceps Brachii, Triceps, Extensor Digitorum, Flexor Carpi Radialis, Rectus Femoris, Gluteus Maximus, Vastus Medialis, Biceps Femoris, Vastus Lateralis, Tibialis Anterior, Gastrocnemius and so on. Total number of elements in the whole body model is about 250,000. Three dimensional surface geometry of each muscle was created based on MRI image data of a human male cadaver with a height of 180 cm and a weight of 90 kg (Visible Human Project Data; NIH, USA). Since the size of the cadaver was larger than that of THUMS, the geometry of each muscle was resized to fit THUMS by referring to configuration and individual size of muscles and bones depicted in cross-sectional image data obtained from anatomical tests such as (Agur et al., 2005[2]). Then each muscle was modeled with hexahedron meshes by using HyperMesh ver.8 (Altair Engineering, USA). The maximum aspect ratio and jacobian of solid elements for muscles were 8.95 and 0.41, respectively. The physiological cross section area (PCSA) of each muscle model was determined based on Winters (1990)[41].

Each muscle FE model was represented as a hybrid model by combination of solid elements with passive muscle properties and bar elements with active muscle properties. The solid elements were modeled with a rubber-like material model (LS-DYNA: #181, MAT_SIMPLIFIED_RUBBER) to simulate 3D geometry of individual muscles and non-linear passive properties. This material model is based on Ogden model and users can use the model by inputting a single uniaxial non-linear stress-strain curve. Poisson's ratio is automatically set to 0.495 (Du Bois, 2003)[14]. The non-linear passive properties were given using tensile properties of muscles obtained from Yamada (1970)[42]. The bar elements were modeled with a Hill type muscle model (LS-DYNA: #156, MAT_MUSCLE) to generate muscular force according to inputted activation levels which are in range from 0 to 1. Some material properties are needed for the Hill type muscle model. A maximum contraction force per unit cross-sectional area of 5.5 kgf/cm² and the PCSA of each muscle were obtained from Gans (1982)[18] and Winters (1990)[41], respectively. The active force-length and active force-velocity were obtained from Thelen et al. (2003)[38]. Although the passive force-length relations are needed in the Hill type model, they were not assigned to bar elements because the solid elements have the passive properties.

This hybrid muscle FE model was applied for a single muscle such as Biceps Brachii and was used to validate the mechanical responses against fundamental characteristic features of a single muscle, that is, the force-length curve and force-velocity curve shown by Thelen et al.(2003)[38]. In addition, the hybrid muscle model was also validated against

human volunteer test data and reproduced increase of muscular stiffness with increase of muscle activation level as observed in the tests. Figure 2 shows an experimental setup of the human volunteer tests. One healthy male volunteer of 33 years old with a weight of 75kg and a height of 176cm who was close to AM50 without any history of neurological or musculoskeletal disorders participated in this test. He gave his informed consent. All procedures were approved by the institutional ethics committee and conducted in accordance with the Declaration of Helsinki. The subject held his posture on supine body position and kept his elbow angle as 90 degrees with his muscular power while a load was given to his right wrist. Then, the subject pushed the circular head of the indentation machine with a diameter of 7mm

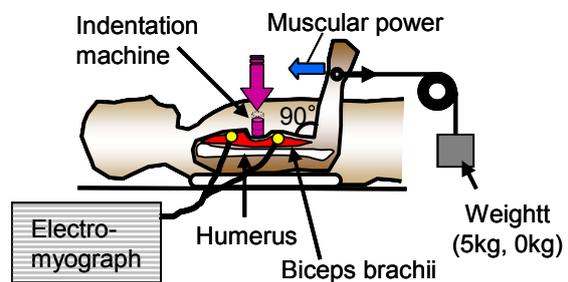


Figure 2. Experimental setup for measuring muscle stiffness and activation level

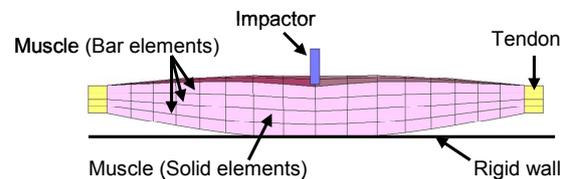


Figure 3. Simulation setup for validation of muscle stiffness.

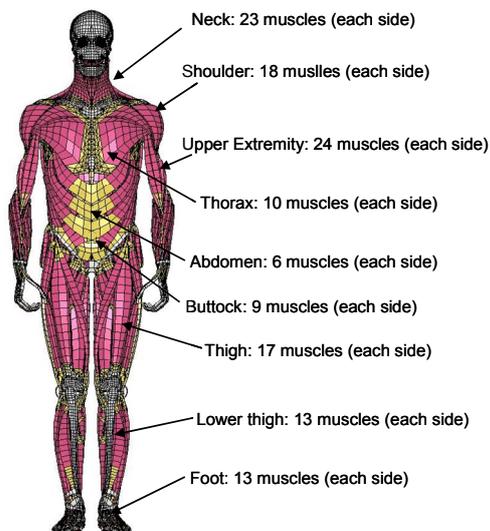


Figure 1. A human whole body FE model with muscles

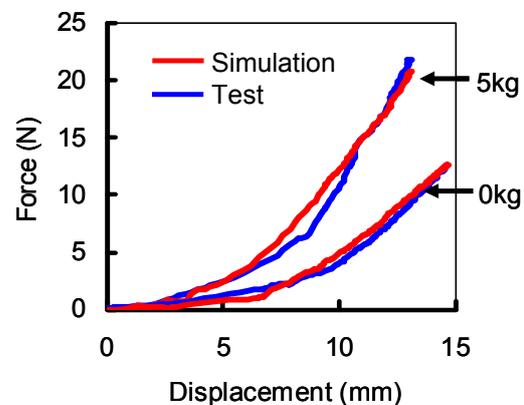


Figure 4. Comparison of muscle stiffness between experimental data and simulation results

into the most bulgy part of his biceps brachii in two cases with and without a weight of 5kg by himself. The EMG activity of the biceps brachii was measured. Figure 3 shows a simulation setup. In this simulation, the biceps brachii muscle was simplified and was pushed in the middle of the whole muscle while both ends of the muscle were fixed with a rigid wall. The rigid wall represented a bone to simulate the muscle pinched between the indentation head and the bone. Because the elbow joint angle little changed, we assumed the muscle length did not change and then we fixed tendons in both ends of the muscle. Displacement time history curves obtained from the tests with and without the weight were used for translating the head for the muscle. Muscle activation levels with and without the weight was assumed as constant values of 5% and 0.16%, respectively. These activation levels were the average values of the test data. Figure 4 shows a comparison between the model prediction and test data. The predicted force-displacement curves well agreed with test data for both cases with and without the weight. The detail descriptions of these validations are found in authors' publication (Iwamoto et al., 2009[22]).

According to anatomical text (Agur et al., 2005)[2], each muscle model was connected to the corresponding bone model through tendon models. The tendons were modeled by using shell and solid elements at both ends of muscles. Material properties of the tendons were obtained from the literature (Pioletti et al., 1998[30], Carlson et al., 1993[9]). Some sliding contacts were defined to produce the interaction between adjacent two muscles and the interaction between muscles and bones close to the muscles. The skin was modeled using shell elements with elastic material properties obtained from Yamada (1970)[42] while the fat was modeled using solid elements with the rubber-like material model. Sliding interfaces were also defined to produce interaction between the muscles and the skins.

To perform occupant injury analyses in frontal impacts, the human whole body FE model shown in Figure 1 must be changed to a sitting posture. The human model allows each joint angle of whole body to change by inputting a time history curve of activation level from 0 to 1 into each muscle. Although the model has possibility to change postures by activating each muscle, currently we do not have any enough muscle controllers for posture changes. Therefore, we determined activation level time history of each muscle based on EMG activity measured in volunteer tests. In this study, we conducted a series of volunteer tests on arm flexion from 165 to 90 degrees around right elbow joint while standing and obtained EMG activity of fourteen muscles of the right arm; the biceps brachii, brachialis, long head and medial head of triceps, extensor digitorum, flexor carpi ulnaris and so on

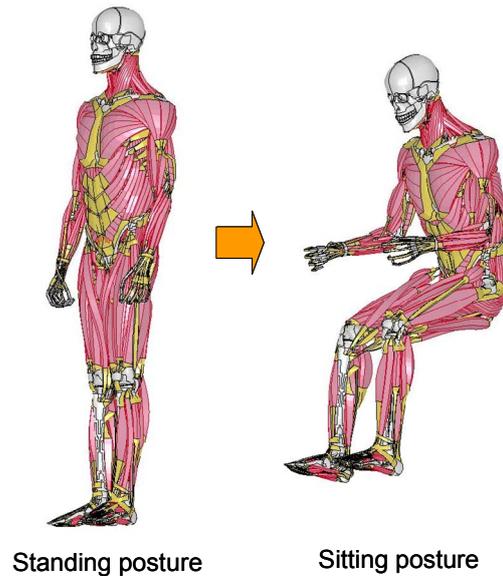


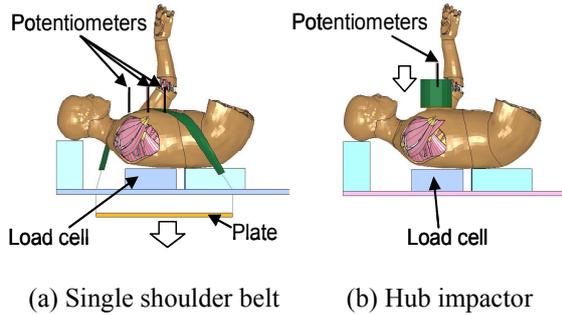
Figure 5. Posture change simulation results

(Iwamoto et al., 2009[22]). The activation curves obtained from the EMG data were used to estimate activation levels of whole body muscles for posture change from the standing posture to a sitting posture. According anatomical tests such as Agur et al.(2005)[2], we classified a role of each muscle for a unique motion, for example, flexion and extension of arm, leg, trunk, and neck as the agonists, synergists, and antagonists. Then, we hypothesized that the activation curves of agonists, synergists, and antagonists in whole body were similar to those of agonists, synergists, and antagonists in arm flexion obtained in the volunteer tests. Then, the absolute values of the activation levels were adjusted to achieve each target position for each motion. Consequently, a sitting posture was developed as shown in Figure 5. The detailed description of the posture change can be found in the authors' publication (Iwamoto et al., 2009[22]).

Model validation

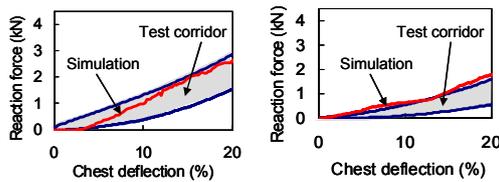
The developed human whole FE model with muscles was validated against two series of cadaver tests on thoracic responses and occupant behaviors in frontal impacts. In addition, the model was also validated against foot impact cadaver tests.

Thoracic responses in frontal impacts Kent et al. (2004)[25] presented thoracic response corridors developed using fifteen post-mortem human subjects (PMHS) subjected to single and double diagonal belt, distributed, and hub loading on the anterior thorax. Subjects were positioned supine on a table and a hydraulic master-slave cylinder arrangement was used with a high speed materials testing machine to provide controlled chest deflection



(a) Single shoulder belt (b) Hub impactor

Figure 6. Simulation setups for thoracic impact

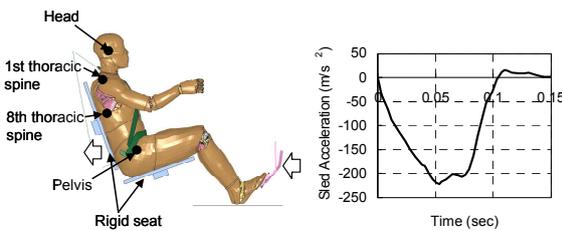


(a) Single shoulder belt (b) Hub impactor

Figure 7. Comparison of force-deflection responses between simulation results and test data

at a rate similar to that experienced by restrained PMHS in a 48-km/h sled test. Thoracic response was characterized using the deflection at the midline of the sternum and a load cell mounted between the subject and the loading table. Simulation setups using the human FE model carefully reproduced the abovementioned experimental setups. In this paper, only two simulation results with the single diagonal belt and hub loading were depicted. Figure 6 shows simulation setups for the two cases. Figure 7 shows simulation results of the posterior reaction forces and chest deflection compared with test corridors. Simulation results almost fell within test corridors in both single diagonal belt and hub loading.

Occupant behaviors in frontal impacts Vezin et al. (2001)[39] conducted a series of sled tests using four unembalmed cadavers to see head and thorax responses of occupants in frontal impact. The rigid flat seats with geometry close to that of a standard mid-size car were used in the tests. The feet of the



(a) Simulation model (b) Sled acceleration

Figure 8. A simulation setup for frontal impacts

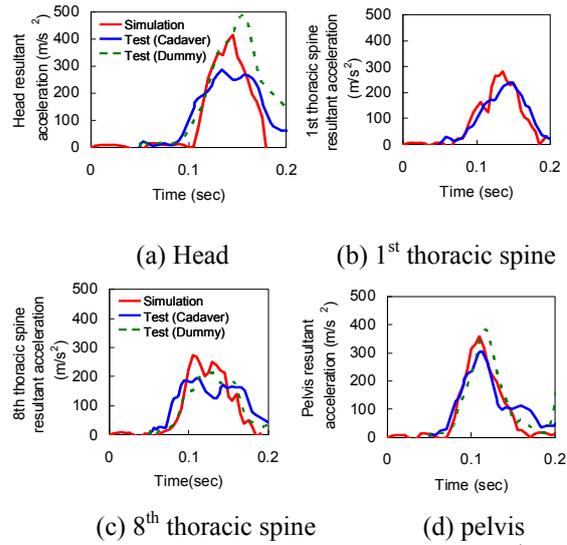


Figure 9. Resultant acceleration of pelvis, 1st and 8th thoracic spine, and head

cadavers were fixed on the footrest while the hands were maintained in the natural driver posture in the 10:10 o'clock position, with two nylon wires, which were released at the impact. The same device was used to maintain the head in a natural position just before the impact. The seat back was tilted at 20 degrees angle. The subjects were restrained by separate shoulder and static pelvis belts. The shoulder belt was equipped with a force-limiting system. Energy absorption by the retractor assembly was controlled through a torsion bar and the belt restraint was a standard production retractor system without a pre-tensioning device. The pre-tension was made manually before the crash. The nominal force limit was 4kN for the two first pairs of tests. Simulation setups with force limit of 4kN using the human FE model carefully reproduced abovementioned experimental setups. Figure 8 shows a simulation setup for frontal impact simulations. Figure 9 shows comparison of resultant accelerations of the pelvis, 1st and 8th thoracic spine, and head between simulation results and test data. Simulation results show good agreement with test data.

Lower leg responses in frontal impacts

Impact response of right lower leg was also validated against cadaveric test data with a preload simulating occupant bracing before foot impacts. Kitagawa et al. (2001)[26] conducted a series of impactor tests using four human cadaveric legs. All specimens were allowed to be sectioned above the knee at mid-femur to preserve the functional anatomy of the knee joint and leg musculature. Specimens were instrumented with an implanted tibia load cell to measure the tibial forces and moments. The specimen was mounted in a position simulating driver geometry. A rigid bar was attached to the femur and connected at the hip joint.

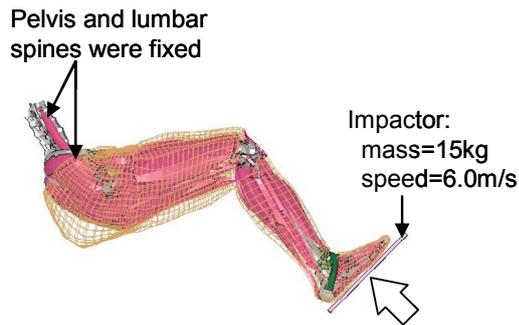


Figure 10. Simulation setup for foot impact

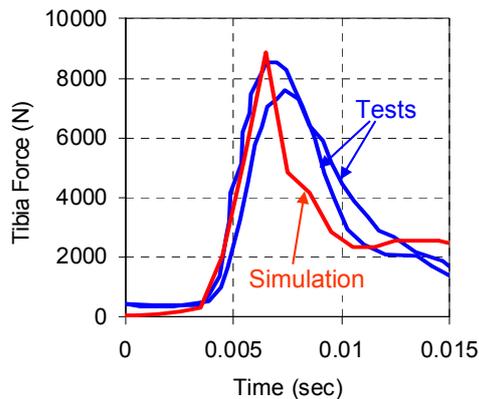


Figure 11. Comparison of tibial axial force between simulation result and test data

The femur was positioned and rotated to correct for the natural valgus angle at the knee such that the long axis of the tibia would be aligned with the direction of the impact direction when the foot was placed on the footplate. A 9.5mm thick piece of foam padding was placed between the foot and the footplate to damp out oscillation. The effect of occupant bracing was simulated externally with a harness placed over the knee which was attached to a spring via a pulley. Immediately before impact, the harness was tightened until the axial load in the specimen reached half of the specimen's body weight. Impacting energy was generated by a rigid pendulum with an effective mass of 15kg and the average impact speed of 6.0m/s (Crandall et al, 1996[13]).

Figure 10 shows a simulation setup using the human leg FE model with muscles for foot impact. A 9.5mm thick footplate including foam padding with a mass of 15kg was modeled as an impactor to reproduce abovementioned experimental setup. In this study, to represent occupant bracing, muscle models in the right lower extremity were activated by using EMG data of right lower extremity in braking motion, which were obtained from a volunteer test we conducted as shown below. The femur model was

fixed in braking motion before impact while pelvis and lumbar spine were fixed. After the preload predicted at the tibia reached a preload of about 300N measured at the tibia in the test, the footplate was impacted with the initial velocity of 6.0m/s and then the femur was released to reproduce cadaver's leg responses. Figure 11 shows comparison of tibial axial force between simulation result and test data. The simulation result shows good agreement with test data.

VOLUNTEER TEST

Activity of each muscle is critical to simulate a bracing situation in pre-impact by using a developed human body FE model with muscles. Since no data of muscle activity for bracing situations were found, we developed an experimental test apparatus in our laboratory to obtain muscle activity for a selected bracing situation. In real-world accidents, drivers show various types of bracing situations. Audrey et al. (2009)[3] conducted a series of volunteer test to analyze driver behavior during critical events using a driving simulator. Eighty subjects who are aged between 22 and 30 years old have participated to the test. They found that more than 67% of subjects moved backward with right leg extended to a brake pedal and arms extended to a steering to anticipate the crash. According to their findings, we selected a bracing situation in which a volunteer subject pushes his right foot on a brake pedal and his hands on a steering with maximal voluntary force.

In this study, a volunteer test with one healthy male subject of 33 years old whose height was 176.5 cm and weight was 75 kg, similar to AM50, was conducted to obtain physiological information in a bracing situation with braking under his informed consent based on the Helsinki Declaration. All procedures were approved by the institutional ethics committee. In this test, the subject was asked to push his right foot on a brake pedal and his hands on a steering with his maximal voluntary force in the developed test apparatus fixed on the laboratory. Figure 12 shows a diagram of developed measuring system. Six data sets were obtained using the system:

- (1) 3D motions of the subject
- (2) 24 electromyography (EMG) from skeletal muscles of upper and lower extremities (cf. Table 1)
- (3) Pressure distributions on seats
- (4) Pedal force
- (5) Right and left separated steering forces
- (6) Reaction force on seats.

The obtained volunteer test data were analyzed and each joint angle during braking motion was calculated from measured 3D motions of the subject. Figure 13 shows time history of pedal force measured in the test. The subject's maximal braking

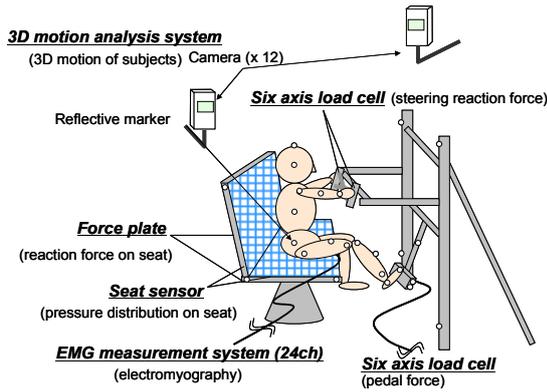


Figure 12. Diagram of developed measuring system

Table 1. Muscles measured in test

Muscles (Lower ex.)	Muscles (Upper ex.)
Soleus	Pectoralis Major
Tibialis Ant.	Deltoid Ant.
Gastrocnem Med.	Deltoid Lat.
Gastrocnem Lat.	Deltoid Post.
Vastus Med.	Infraspinatus
Vastus Lat.	Biceps Brachii
Rectus Femoris	Brachialis
Adductor Longus	Triceps Brachii
Biceps Femoris	Flexor Carpi Radialis
Semitendinosus	Flexor Carpi Ulnaris
Gluteus Maximus	Brachioradialis
	Extensor Carpi Radialis
	Extensor Carpi Ulnaris

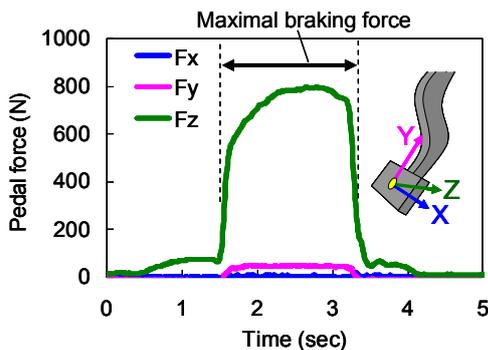


Figure 13. Measured pedal force

force was reached to 750N, which was comparable with previously reported values of 700-1000N for emergency braking situation (Audrey et al., 2009[3], Owen, C. et al., 1998[28], Palmertz C. et al., 1998[29]). Figure 14 shows activation levels of Soleus, Tibialis Anterior, Biceps Femoris (Long Head), and Rectus Femoris in right lower extremity. The activation levels were normalized by dividing EMG signal of each muscle measured in the test by

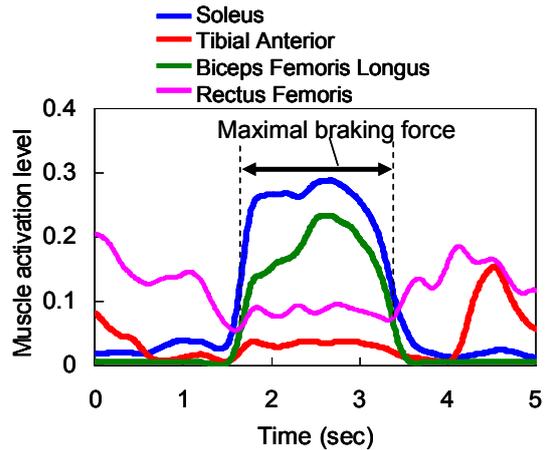


Figure 14. Measured muscle activation levels of right lower extremity

the maximal EMG signal, which was obtained from other tests on maximal voluntary force conducted using the same subject in the same day. The Soleus and Tibialis Anterior are extensor and flexor muscles of ankle joint, respectively. Biceps Femoris (Long Head) and Rectus Femoris are flexor and extensor muscles of knee joint as well as extensor and flexor muscles of hip joint, respectively. In the braking motion, activation levels of extensor muscles of ankle joint and hip joint were increased to 25-30% while those of flexor muscles of ankle joint and hip joint were less than 10%. The muscle activity suggests that right lower extremity was extended in the braking motion. Therefore, the selected braced situation was appropriately reproduced in this test.

FRONTAL IMPACT SIMULATIONS

In frontal automotive accidents, drivers made evasive maneuvers on braking and steering to reduce their vehicle speeds and avoid crashes. In such emergency cases, drivers also might brace their body with their muscle activity to prepare the upcoming impacts. In this study, a frontal crash situation was selected to find out differences of an adult male driver's behaviors and injury outcomes in post-crash between a living human body and a cadaveric human body, which have not been estimated so far. We selected a crash situation which an adult male driver made an evasive maneuver of braking with a deceleration of 0.7G for 600 ms in pre-crash and then he sustained a frontal impact with a speed of 50km/h. In pre-crash phase, he pushed his right foot on a brake pedal and his hands on a steering with his maximal voluntary force and simultaneously braced his body to reduce the impact speed and protect his body for the impact. In post-crash phase, he could not do anything for the impact, although he kept his muscle activity until 85ms after impact. We simulated this situation using

the developed human FE model with muscles as described in following sections.

In the selected crash situation, the driver made an emergency braking and reduced his vehicle speed. However, we do not have experimental volunteer test data on emergency braking using a vehicle or a simulated vehicle which should include a driver's motions, EMG data of some muscles, and reaction forces of a brake pedal, a steering, a seat cushion, a seat back and so on for reconstruction of a driver's kinematic and kinetic responses. These kinds of experimental volunteer tests are not easy to be conducted due to risks for volunteers. Therefore, this study adopted an alternative method to reconstruct the pre-crash situation. We simulated a driver's kinematic and kinetic responses in his emergency braking using a deceleration of 0.7G automatic braking obtained from the literature (Ejima et al. 2010[16]) and the volunteer test data on a bracing motion with a maximum voluntary force conducted in the laboratory static apparatus as mentioned previously.

As shown in Figure 15, in the simulation setup, the developed human FE model with muscles was set to a sitting position with rigid seats while the right foot was positioned on a brake pedal and the hands was positioned to get a grip on a steering in order to reproduce the volunteer test setup. A 3-point belt model with a force-limiter of 4kN and a pretension was also equipped with the simulation setup. The normalized EMG activity of 24 muscles measured in right lower extremity and right upper extremity was

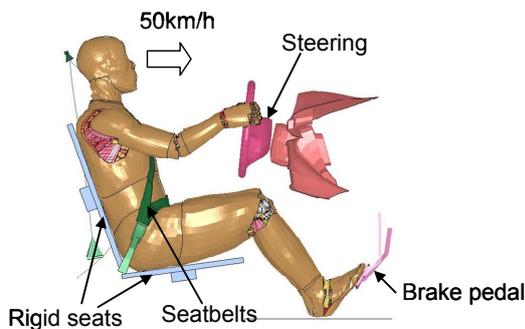


Figure 15. Simulation setup of frontal impacts

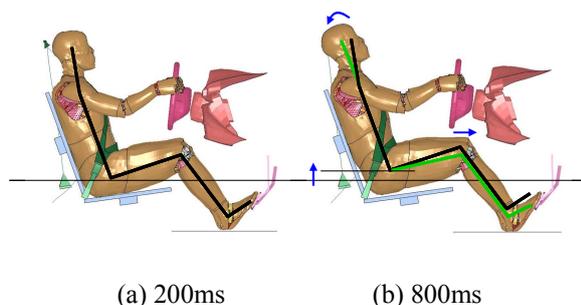


Figure 16. Comparison of occupant postures before a braking motion and before an impact

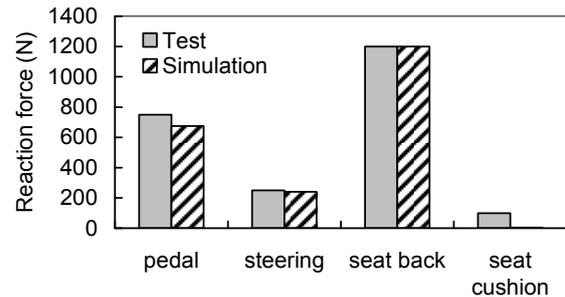


Figure 17. Comparison of reaction forces between simulation results and volunteer test data

directly inputted to the corresponding muscle model. Muscle activity of other muscles in the right lower extremity and the upper extremities which were not measured in the test was estimated to reproduce forces on a brake pedal and a steering, respectively. The method to estimate the activation levels of muscles is almost the same as described above. Since muscle activity of the left lower extremity was not measured in the test, we assumed the muscle activity as similar to that in the right lower extremity. In addition, muscle activity of the trunk and neck was also not measured in the test. Therefore, we assumed activation levels of most muscles in the neck and trunk as 10-20% because activity of some muscles such as Sternocleidomastoid in the neck and Rectus Abdominis in the trunk presented 10-20% in other experimental volunteer tests conducted in our laboratory. Activation levels of Longus Colli, Scalenus Anterior, and Sternohyoid associated with neck flexion were assumed as 50% to reproduce the volunteer's neck motion.

In the simulation, only an acceleration of gravity was given to have the human FE model sit on the seat from an onset of the simulation until 200ms and after 200ms a deceleration of 0.7G was inputted to a sled model including the rigid seats, the brake pedal, the steering, the seatbelt, and the floor for a period of 600ms. After 800ms, an acceleration of 50km/h shown in Figure 8 was applied to the sled model in order to reproduce a frontal impact situation. The activity of each muscle was given to the muscle model at 100ms after the onset of the simulation and was assumed to be kept as a constant value until the end of simulation after the pedal force reached the maximum. This is because we do not have EMG data of volunteers during brake deceleration and frontal impact sled deceleration.

Figure 16 shows a comparison of the driver's postures at 200ms before the braking motion and at 800ms before the impact. Comparing with the posture before the braking motion, the hip displaced upward and the right leg displaced forward and downward while the head rotated rearward in the

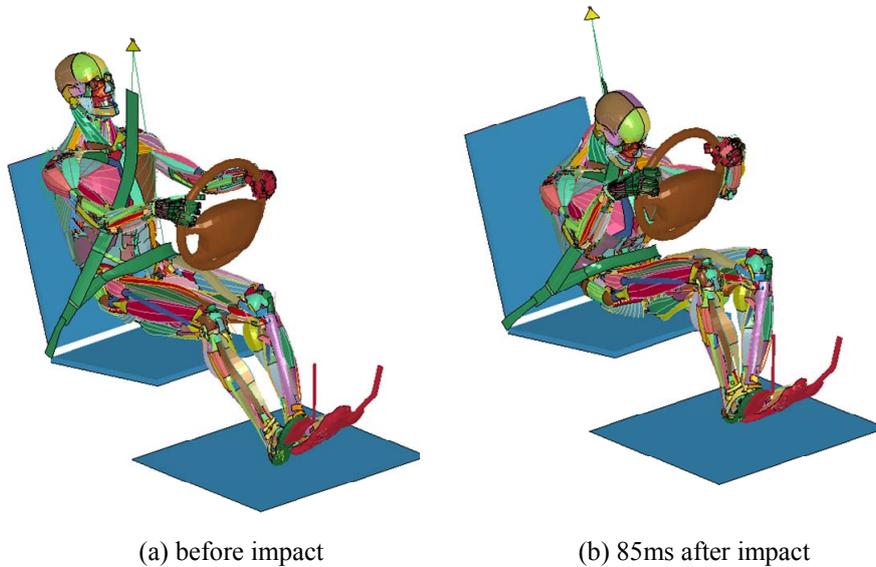


Figure 18. Occupant motions predicted by active human model (50km/h)

posture before the impact. This predicted braking motion was similar to that observed in the volunteer test. Figure 17 shows a comparison of reaction forces between simulation results and volunteer test data. Predicted forces of the pedal, the steering, and the seat back showed good agreement with test data. Predicted force of the seat cushion was zero while the force was 100N in the test. This inconsistency is because the hip was completely apart from the seat cushion in braking motion of the simulation.

In this study, we are interested in the difference of a driver's behaviors and injury outcomes between a living human body with muscle activity and a cadaveric human body without muscle activity. We are also interested in the rate dependency of muscular effects for the driver's behaviors and injury outcomes. Therefore, four parametric simulations on frontal impact situations were performed to find out the difference and the rate dependency. Case 1 represents a crash situation for an active human body with an impact speed of 50km/h. Case 2 represents that for a cadaveric human body with the same speed of 50km/h. Case 3 represents that for an active human body with an impact speed of 25km/h. Case 4 represents that for a cadaveric human body with the same speed of 25km/h. For the cadaveric human body, all muscles were assumed not to be activated. However, when the muscle model was given as zero of activation level, the model caused instability. Therefore, less than 1% of activation levels were inputted to the muscles of the cadaveric human body model. An acceleration time history for impact speed of 25km/h was determined using that of 50km/h. Figure 18 shows occupant motions predicted by the active human model with 50km/h. The skin models were removed to show muscle models clearly. From Figure 18, the active human model braced his body

to push his arms on a steering and right foot on the brake pedal before impact. After impact, the active human model continued to brace his body and kept his head from approaching the steering. Figure 19 shows the driver's postures at 85ms after impact for the four cases. The posture of the active human body

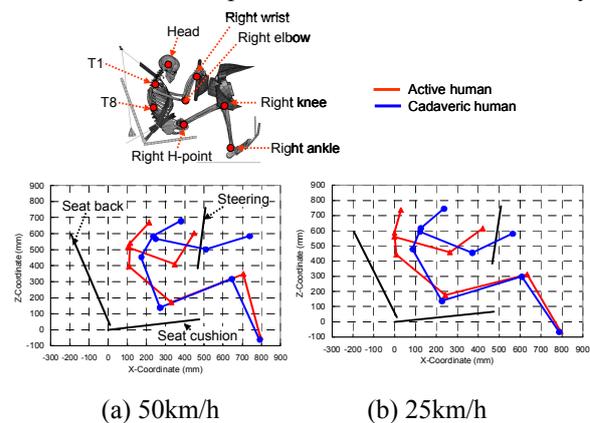


Figure 19. Occupant postures at 85ms after impact

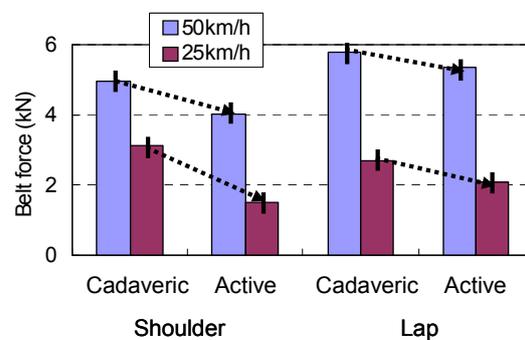


Figure 20. Comparison of maximum forces at shoulder and lap belts

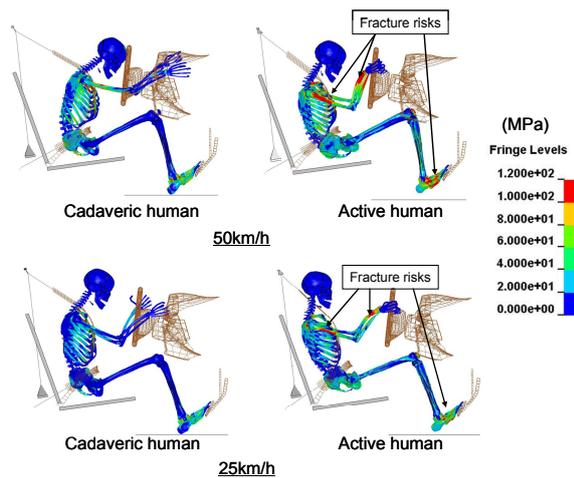
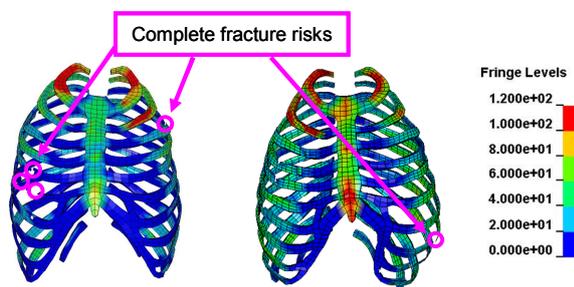


Figure 21. Comparison of stress distribution at 85ms after impact



(a) Cadaveric human (b) Active human

Figure 22. Comparison of rib fractures between cadaveric human body and active human body

was different from that of the cadaveric human body in both 50km/h and 25km/h. In comparison with the cadaveric human body, the knee and hip went forward while the upper body including upper extremities went backward in the active human body. In addition, difference of the driver's postures between the active human body and the cadaveric human body in lower speed of 25km/h was larger than that in higher speed of 50km/h. Figure 20 shows comparison of maximum forces at a shoulder belt and a lap belt for four cases. In both the shoulder and lap belts, maximum belt forces of the cadaveric human body were larger than those of the active human body at 50km/h. The similar trend was found at 25km/h. Figure 21 compares Von Mises stress distribution of skeletal parts at 85ms after impact for four cases. The active human body sustained more fracture risks at upper and lower extremities than the cadaveric human body. Figure 22 shows comparison of Von Mises stress distribution in the ribcage between the cadaveric human body and the active human body. The figure includes locations of complete fractures. The simulation result of cadaveric human body was obtained from the validation results using frontal sled cadaver test data with an impact speed of 50km/h conducted by Vezin

et al. (2001)[39] as shown previously. The simulation result of the active human body was obtained from the simulation results of frontal impact simulation for the active human body with 50km/h. The comparison shows that the cadaveric human body sustained more rib fractures than the active human body.

DISCUSSION

It is one of challenging issues to predict a driver's injuries and activate his muscles simultaneously during pre-impact and post-impact using human FE models. However, it is critical to understand how drivers sustain injuries in real-world accidents. In past decade, a lot of human body FE models have been developed and validated against existing cadaver test data (Iwamoto et al., 2002[21], Vezin et al., 2005[40], Ruan et al., 2005[31] etc.). However, these models could not be used to analyze muscular effects in pre-impacts for injury outcomes in post-impacts due to lack of active muscles in these models. On the other hand, some human multi-body models with active muscles have been developed (SIMM[34], AnyBody[4], etc.) and used for predicting muscular force and fatigue in human motions such as exercise, driving, ingress, and sports. In most of these models, skeletal parts, anatomical joints, and muscles were simplified to rigid bodies, mechanical joints, and line segments, respectively. Therefore, these models could not be used to analyze occupant injury risks in post-impacts. Some researchers tried to combine both benefits and analyze muscular effects in pre-impacts and post-impacts (Chang et al., 2008[11], Sugiyama et al., 2007[36]). They incorporated muscles modeled by bar elements into an existing human FE model and predict bone fracture risks with muscle activity. However, the muscle model represented by bar elements has less accuracy for injury analyses because of the following three reasons. Firstly, the model cannot represent exact interactions of muscle-to-muscle, muscle-to-bone, muscle-to-skin. Secondly, the model cannot represent stiffness changes in the transverse plain to a muscular action line according to activation levels. Finally, unrealistic stress concentrations could occur because the muscle bar elements have to be connected with rigid elements defined in skeletal parts. On the contrary, each muscle model developed in this study was represented as a hybrid model of solid elements with muscular 3D geometry and bar elements with active muscular properties. Therefore, the muscle model has more accuracy for injury analyses.

This study selected the most typical bracing situation among volunteer tests using eighty subjects and a driving simulator performed by Andrey et al. (2009)[3] to investigate bracing effects in pre-impacts. A male volunteer subject pushed his

right foot on a brake pedal and his hands on a steering with maximal voluntary force in static laboratory apparatus to reconstruct the bracing situation. Then, muscle activity in the upper and lower extremity was measured and inputted to muscles of the developed human model to simulate the bracing situation. However, the muscle activity measured in the test is that for a bracing situation in a static condition and might be different from that for a bracing situation in a dynamic condition such as deceleration of braking or impact. Recently, some volunteer tests have been conducted under deceleration of braking or impact (Ejima et al., 2009[15], Beeman et al., 2010[5]) and EMG data were measured in the tests. Unfortunately, they did not make clear the difference of muscle activity between a dynamic situation and a static situation. Postural control could change the muscle activity under the deceleration and a driver's mental state could affect the muscle activity in cases of panic braking. However, this study focused on investigating the difference of a driver's behaviors and injury outcomes between a living human body with muscle activity and a cadaveric human body without muscle activity. Although further studies are needed to investigate effects of a driver's postural control and mental state on muscle activity of whole body, the muscle activity used in this study is good enough to investigate the difference between a living human body and a cadaveric human body.

The simulation results using the developed human body FE model with muscles demonstrated that an active human body kept the position of the upper body backward and also kept the position of the lower extremities forward for the braking deceleration comparing with a cadaveric human body as shown in Figure 19. These occupant behaviors are similar to those observed in comparison between tensed volunteers and relaxed volunteers conducted under a braking deceleration of 0.7G (Ejima et al., 2010[16]). From the difference on occupant behaviors, forces of the shoulder and lap belts in the active human body were a little bit smaller than those in the cadaveric human body, although shoulder belt force of the active human body was much smaller than that of the cadaveric human body at the lower speed of 25km/h (Figure 20). Therefore, the active human body had less rib fracture risks than the cadaveric human body at 25km/h. Since muscular forces in upper and lower extremities of the active human body increased and the lower extremities had more forward positions, the active human body had more bone fracture risks in the upper and lower extremities comparing with the cadaveric human body (Figure 21). The difference of injury outcomes between the active human body and the cadaveric human body appeared more remarkably at the lower speed of 25km/h.

Kallieris et al. (1995)[23] compared 29 sled tests with belted cadavers and 24 accident cases with 24 belted drivers and 6 belted front passengers at the configuration of the frontal collision with impact speeds of about 50km/h. They found fractures of the radius in the upper extremities as result of reinforcement against the steering wheel during the collision phase in the accident cases while no injuries were observed in the cadaver tests. They also found some leg injuries including fractures at the femur, tibia, fibula, foot, and ankle joint while no injuries were observed in the cadaver tests. Additionally they reported that the cadaver tests showed a rib fracture frequency twice as high as for the accident cases. Since the cadaver tests conducted by Kallieris et al.[23] did not include braking deceleration, injury outcomes in the cadaver sled tests might correspond to those for cadaveric human body simulated in the validation using cadaver test data conducted by Vezin et al. (2001)[39]. On the other hand, injury outcomes in accident cases might correspond to those for active human body with a braking deceleration of 0.7G and an impact speed of 50km/h. In the simulations, the active human body had more fracture risks in the upper and lower extremities while the cadaveric human body had no fracture risks. In addition, the active human body had less fracture risks in the ribcages than the cadaveric human body as shown in Figure 22. Injury outcomes predicted by the developed human body FE model with muscles show good agreement with those reported by Kallieris et al.[23]. Therefore, the model is a useful tool to investigate the bracing effects in pre-impacts of real-world accidents on injury outcomes.

A lot of researchers have investigated traffic accident data and have tried to find injury patterns and the mechanisms. Some injury mechanisms are still unknown. However, if we consider muscular effects of occupants in pre-impacts and post-impacts, we might be able to elucidate such injury mechanisms. Carroll et al. (2010)[10] reported that occupants seated in the front passenger seat tended to sustain more torso injuries compared with the driver's seat. This mechanism can be explained from the simulation results (see Figure 22). Different from a driver, an occupant seated in the front passenger seat does not push his foot on a pedal and his hands on a steering. If an occupant in the front passenger and a driver are regarded as the cadaveric human body and an active human body with respect to the impact responses, respectively, the occupant is likely to sustain more rib fracture risks than the driver. Rudd (2009)[32] reported that the majority of the foot and ankle injuries occur at lower crash severities with delta-V of less than 30km/h. This mechanism can be also explained from the simulation result that the bone fracture risks in the lower extremities were predicted even in 25km/h for the active human body

(see Figure 21). In addition, as Kent et al. (2003)[24] reported, elderly occupants sustain more thoracic injuries in frontal impacts due to weakness of their skeletal parts. Different from younger occupants, elderly occupants have less muscular power and less muscle activity, besides of the weakness of bone properties. If an elderly occupant and a younger occupant are regarded as the cadaveric human body and the active human body with respect to the impact responses, the elderly occupant could sustain more rib fracture risks than the younger occupant (see Figure 22). Therefore, the active human body FE model has potential for better understanding of unclear injury mechanisms of occupants in automotive crashes.

LIMITATIONS

This study has the following limitations.

- Validation of muscle stiffness was performed for only Biceps Brachii because the volunteer test data were not obtained for other muscles such as Rectus femoris in the leg. Although the hybrid model has possibility to reproduce muscular stiffness change, more data are needed for complete validation of muscle stiffness.
- EMG data of trunk and neck muscles were not measured in the bracing situation of this study because it was not so easy to measure EMG activity of muscles in trunk and neck regions. Therefore, we estimate the activity to reproduce measured reaction forces. Further studies are needed to measure activity of trunk and neck muscles and activity of inner muscles.
- Data of muscle activity under dynamic situations such as brake deceleration and sled deceleration were not obtained in this study. Further study is needed to obtain muscle activity from volunteer test data conducted under such dynamics situations.
- This study selected a bracing situation. However, in real-world accidents, there are some bracing situations including panic condition. Therefore, more investigation is needed for understanding of drivers' behaviors in pre-impacts.
- Muscle activity used in this study was estimated based on EMG data from a volunteer test. Therefore, the activity did not include effect of muscular reflex and posture stabilization. Further study on muscle controller based on neural science is necessary.
- No airbags were included in the frontal impact simulations performed in this study. This indicates that injury outcomes predicted in this study cannot be applied for current commercial vehicles.

CONCLUSIONS

An active human body FE model with 3D geometry of muscles was developed to investigate muscular effects in pre-impact for injury outcomes. The muscle was modeled as a hybrid model of solid elements with passive properties and bar elements with active properties. The muscle model reproduced muscular stiffness change according to muscle activation levels. The developed human body FE model, which had originally a standing posture, was changed to a driver's sitting posture by activating each muscle of whole body. The model was validated against cadaver test data on frontal impacts for the thorax using a 3-point seatbelt and hub impactor. Force-displacement responses predicted by the model fell within test corridors. The model was also validated against cadaver test data on frontal sled impacts using occupants equipped with seatbelts. Acceleration of the head, thoracic spines, and pelvis predicted by the model showed good agreement with those obtained from test data. In addition, the model was used for foot impact simulations with a preload representing braking effect and was compared with cadaver test data obtained from the literature.

This study investigated the bracing effects in pre-impacts for injury outcomes in frontal impacts by frontal impact simulations with pre and post impacts using the developed human body model. A volunteer test was conducted to reproduce a bracing condition, which could occur in real-world accidents, using static laboratory apparatus with rigid seats, a steering, and a brake pedal. Muscle activity obtained from the test was inputted to the muscle models. The model reproduced the bracing condition because predicted reaction forces of the pedal, steering, and seat back agreed well with those of test data. Comparisons between an active human model and a cadaveric human model indicate that muscle activity with the bracing condition could constrain upper body for frontal impacts and cause more bone fracture risks in upper and lower extremities. From frontal impact simulations performed at the impact speed of 50km/h, the cadaveric human model could sustain more rib fracture risks than the active human model. These findings correspond to conclusions from comparison of injury outcomes between real-world accident data and cadaver test data with the same speed of 50km/h. Therefore, the model has possibility to make the detailed investigation of muscular effects in pre-impact for injury outcomes. Although further studies are needed to model the muscular reflex and posture stability control as well as to obtain muscle activity under dynamic situations of brake deceleration and sled deceleration, the active human body FE model would be a useful tool for better understanding of unexplained injury mechanisms in real-world automotive accidents.

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