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Research Report

Development of an Active Human FE Model with 3D Geometry of Muscles for Simulating Driver's Bracing and Evasive Maneuvers in Pre-crash

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ABSTRACT Accident data analyses conducted at the Institute for Traffic Accident Research and Data Analysis (ITARDA) in Japan showed that around 50% of drivers made evasive maneuvers on braking and/or steering in each accident type such as frontal, side, rear impacts. Muscle activities of drivers in the evasive maneuvers might brace their bodies and change their postures just collisions, which 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 and simulated driver's bracing and evasive maneuvers in pre-crash. Parametric simulations were performed to investigate effect of muscle tense on driver's behaviors and injuries in frontal collisions. The simulation results under a condition with rigid seats and a 3-point seat belt and without any airbags demonstrate that braced drivers could constrain their upper body and reduce the head and thoracic injury risks, which is obviously different from drivers without muscle activity. Therefore, the muscle activity is critical for better understanding of occupant injury mechanisms in automotive collisions. The developed model is a unique and practical tool for the detailed investigation on effects of muscle activation in pre-crash for occupant behaviors and injuries under various impact situations.

KEYWORDS Finite Element, Human Model, Muscle, Activation, Bracing, Pre-crash, Post-crash, Injury Mechanism

1. Introduction

Accident data analyses conducted at the Institute for Traffic Accident Research and Data Analysis (ITARDA) in Japan showed that around 50% of drivers made evasive maneuvers on braking and/or steering in each accident type such as frontal, side, rear impacts.[(1)] 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 using human volunteers have been performed to investigate effect of muscle activity on injuries under the assumed impact situations (Tennyson and King, 1976,[(2)] Begeman et al., 1980,[(3)] Funk et al., 2001,[(4)] Levine et al., 1978[(5)]). The studies indicate that 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,[(6)] Vezin et al., 2005,[(7)] Ruan et al., 2005[(8)]). Recently, Shigeta et al. (2009)[(9)] 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 and injury outcomes (Choi et al., 2005,[10] Chang et al., 2008,[11] Behr et al., 2006,[12] Hedenstierna et al., 2008[13]). These models showed that muscle activity could affect the injury outcomes during impact situations. However, these models cannot represent muscular stiffness change according to the activity, which is dominant in bracing conditions, and are not available for simulations of evasive maneuvers on braking and/or steering.

In this study, we developed an active human finite element (FE) model with 3D geometry of individual muscles. The model can reproduce muscular stiffness change according to the activity, and simulate braking and/or steering motions. The active human FE model was validated against cadaver test data on thoracic and lower extremity responses in frontal impacts (Iwamoto et al., 2012[14]). A bracing situation was selected to investigate effect of muscle activity on human body kinematics and injury outcomes in frontal impacts. A volunteer test was conducted to obtain Electromyography (EMG) data of muscles in the upper and lower extremity in the bracing situation. The active human FE model reproduced the bracing situation in pre-impact and then sustained frontal impacts with 50 km/h. Simulation results of the human FE model with muscle activity were compared with those of the human FE model without muscle activity. We discussed effects of muscle activity in pre-impact on human body kinematics and injury outcomes. All simulations in this paper were performed using an explicit finite element code LS-DYNA (LSTC, USA).

2. Model Development and Validations

Muscular FE models of a human whole body were developed and integrated with our previously developed human body FE model called THUMS (<u>T</u>otal <u>HU</u>man <u>M</u>odel for <u>S</u>afety, Iwamoto et al., 2002[6]) whose size was similar to that of AM50 with a height of 175 cm and a weight of 77 kg. **Figure 1** shows a developed active human body FE model in a driving posture. In this figure, the skin was removed to see muscles clearly. The model includes 282 muscles of lower extremities, upper extremities, trunk, and neck such as the Sternocleidmastoid, 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 texts such as (Agur et al., 2005[15]). Then each muscle was modeled with hexahedron meshes by using HyperMesh (Altair Engineering, USA).

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. The non-linear passive properties were given using tensile properties of muscles obtained from Yamada (1970).[16] 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

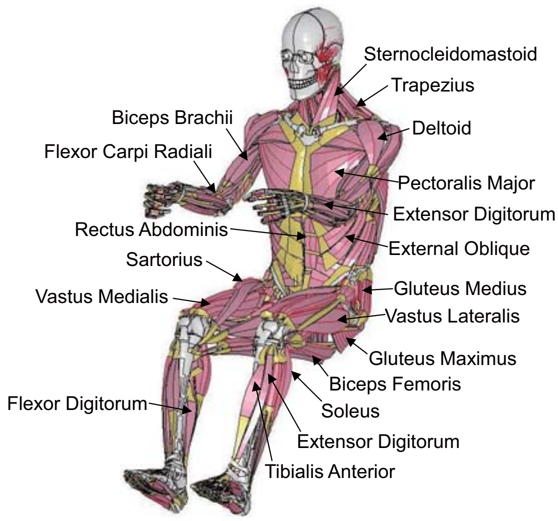


Fig. 1 Active human FE model with muscles.

which are in range from 0.0 to 1.0. Some material properties are needed for the Hill type muscle model. A maximum contraction force per unit cross-sectional area of 5.5 kgf/cm^2 and the physiological cross section area (PCSA) of each muscle were obtained from Gans (1982)[17] and Winters (1990),[18] respectively. The active force-length and active force-velocity were obtained from Thelen et al. (2003).[19] 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).[19] In addition, the hybrid muscle model was also validated against human volunteer test data on indentation for Biceps Brachii with and without the weight of 5 kg and reproduced increase of muscular stiffness with increase of muscle activation level as observed in the tests. **Figure 2** shows comparison of force-displacement curves of Biceps Brachii between simulation results and volunteer test data. The figure also includes the simulation setup, which was reproduced to be almost the same condition as the volunteer test. Muscle activation levels with and without the weight was assumed as constant values of 5% and 0.2%, based on measured EMG data, respectively. The predicted force-displacement curves well agreed with test data for both cases with and without the weight.

The human model allows each joint angle of whole body to be changed by inputting a time history curve of activation level from 0.0 to 1.0 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 an 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. In the simulation using the left arm FE model in Fig. 1, the EMG activity of each muscle was inputted to the corresponding muscle model. **Figure 3** shows comparison of elbow angle time history in arm flexion between simulation results and volunteer test data. Simulation result show good agreement with test data from 165 degrees to 115 degrees. The detail descriptions of these three validations are found in authors' publication (Iwamoto et al., 2009[20]).

The activation curves obtained from the EMG data were used to estimate activation levels of whole body muscles for various posture changes and motions. According to anatomical tests such as Agur et al. (2005),[15] we classified a role of each muscle for a unique motion, for example, flexion and extension of

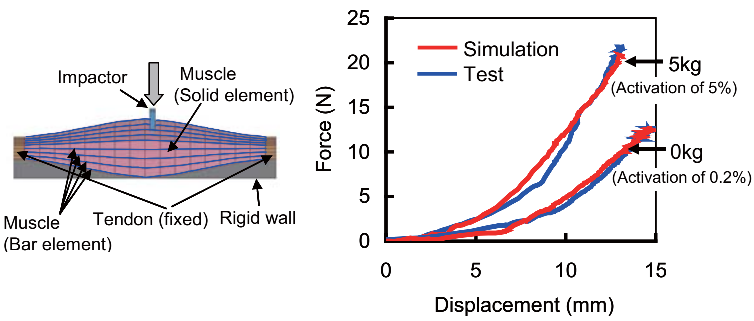


Fig. 2 Force-displacement curves of an arm muscle.

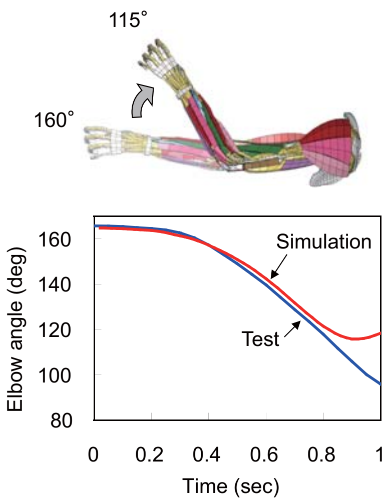


Fig. 3 Validation on arm flexion.

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. The detailed description on how to change posture and generate motions using EMG data can be found in the authors' publication (Iwamoto et al., 2009[20]). **Figure 4** shows simulation results of steering motions. The steering motions seem to be reasonable, but the activation levels used in this simulation should be validated by using EMG data measured in volunteer tests on steering motions.

3. Volunteer Test

Activity of each muscle is critical to simulate a bracing situation in pre-impact by using the developed active human 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. Based on volunteer tests with eighty subjects using a driving simulator performed by Audrey et al. (2009),[21] more than 67% of subjects moved their upper bodies backward with their right legs extended to a brake pedal and their both arms extended to a steering among various bracing situations to anticipate the crash. Therefore, we selected this bracing situation.

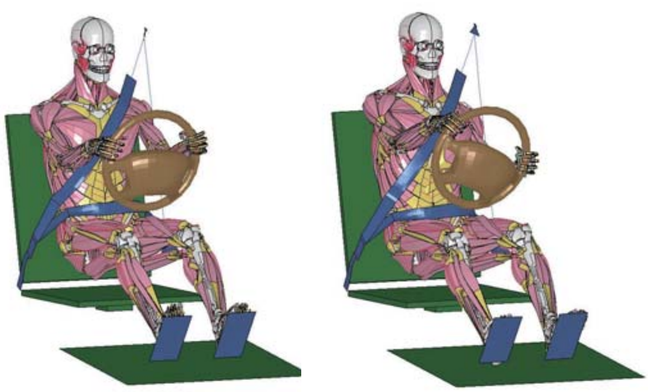


Fig. 4 Simulation results of steering motions.

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 5** shows a diagram of developed measuring system. Six data sets of (1) 3D motions of the subject, (2) 24 electromyography (EMG) from skeletal muscles of upper and lower extremities, (3) Pressure distributions on seats, (4) Pedal force, (5) Right and left separated steering forces, (6) Reaction force on seats, were obtained using the system. **Figure 6** shows time history of pedal force

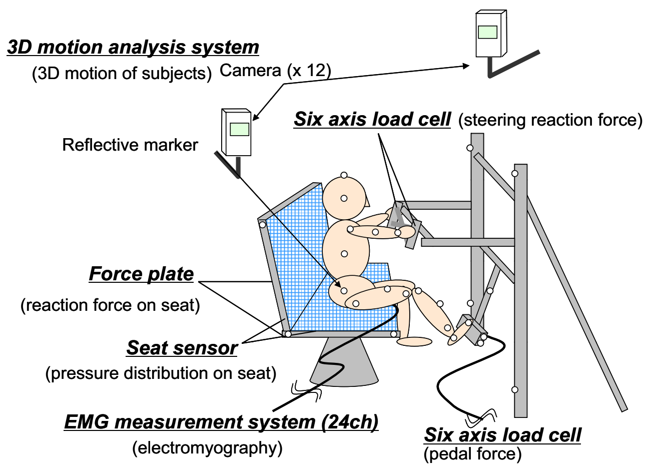


Fig. 5 Diagram of developed measuring system.

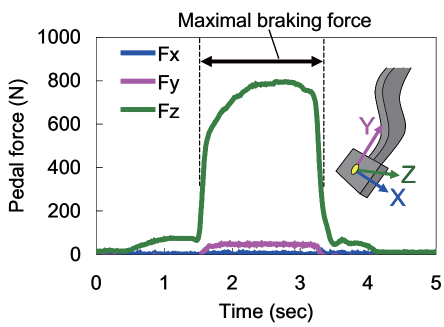


Fig. 6 Measured pedal force.

measured in the test. The subject's maximal braking force was reached to 750 N, which was comparable with that measured by Audrey et al., 2009.[21] **Figure 7** 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 the maximal EMG signal, which was obtained from other tests on maximal voluntary force conducted using the same subject in the same day. 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.

4. Frontal Impact Simulations

Based on the abovementioned studies conducted by Audrey et al. (2009),[21] 67% of drivers made an evasive maneuver of braking in which they moved their upper bodies backward with their right legs extended to a brake pedal and their both arms extended to a steering to anticipate the crash. Therefore, we performed simulations using the active human FE model under a crash situation in which an adult male driver made the evasive maneuver of braking with a deceleration of 0.7 G for 600 ms in pre-crash to reduce impact velocity and then he sustained a frontal impact with a speed of 50 km/h.

Figure 8 shows a simulation setup of crash situation. The active human FE model 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 (Fig. 8(a)). A 3-point belt model with a force-limiter of 4 kN and a pretension was also equipped with the simulation setup. No airbags were equipped in this simulation. Figure 8(b) shows an acceleration time history inputted for a sled model including rigid seats, a steering, a brake pedal, seatbelts, and a floor. Only an acceleration of gravity was given to have the human FE model sit on the seat from an onset of the simulation until 200 ms and after 200 ms a deceleration of 0.7 G was inputted to the sled model for a period of 600 ms. After 800 ms, an acceleration of 50 km/h was applied to the sled model in order to reproduce a frontal impact situation. The brake deceleration of 0.7 G was obtained from Ejima et al. (2010)[22] and the impact deceleration of 50 km/h was obtained from Vezin et al. (2001).[23]

In this study, two simulations under the abovementioned crash situation were performed to find out differences of an adult male driver's behaviors and injury outcomes in post-crash between an active human model and a cadaveric human model, which have not been estimated so far. For the cadaveric human model, less than 1% of activation levels were inputted to the muscle models to avoid the instability. For the active human model, activation level time history of each muscle were estimated with the maximum values of 20-60% based on the normalized

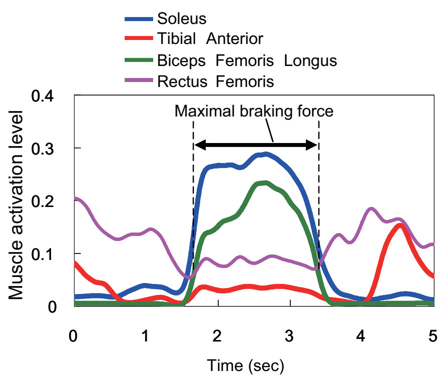
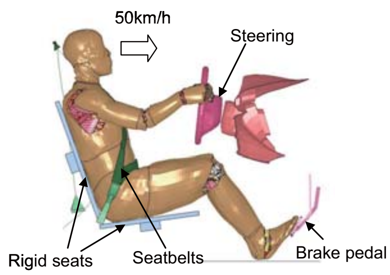
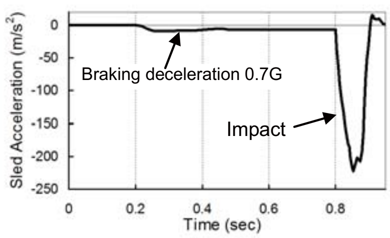


Fig. 7 Measured muscle activation levels of right lower extremity.



(a) Simulation condition



(b) Sled acceleration

Fig. 8 Simulation setup of frontal impacts.

EMG activity of 24 muscles in right lower extremity and right upper extremity obtained from the volunteer test conducted in a bracing condition. The method to estimate the activation levels of all muscles is almost the same as described above.

In pre-crash, driver's postures and reaction forces in the maximum braking force before impact were compared between simulation results using the active human model and volunteer test data. **Figure 9** shows a comparison of the driver's postures before a braking motion and before an 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 posture before the impact. This predicted braking motion was similar to that observed in the volunteer test conducted in our laboratory and the volunteer test reported by Audrey et al. (2009).[21] **Figure 10** 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 100 N in the test. This inconsistency is because the hip was completely apart from the seat cushion in braking motion of the simulation. This is because we regarded activity of each muscle in the left lower extremity as the same as that in the right lower extremity.

In post-crash, driver's behaviors and injury outcomes were compared between the active human model and the cadaveric human model. **Figure 11** compares Von Mises stress distribution of skeletal parts at 85 ms after impact (885 ms in total) between the cadaveric human model and the active human model. The active human model sustained more fracture risks at upper and lower extremities than the cadaveric human model. On the other hand, the active human model sustained less fracture risks at the skull and ribs than cadaveric human model. This is because the active human body constrained his upper body differently from the cadaveric human body. **Figure 12** shows comparison of the maximum rib deflection between the cadaveric human model and the active human model. The

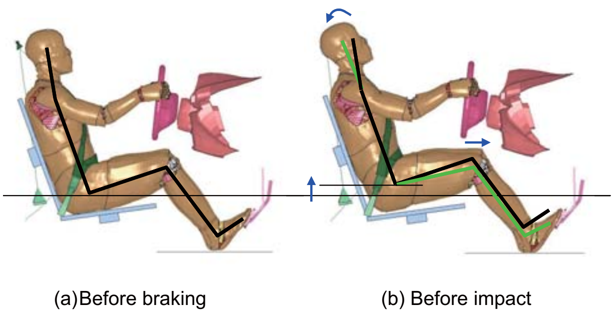


Fig. 9 Comparison of driver's postures before braking motion and before impact.

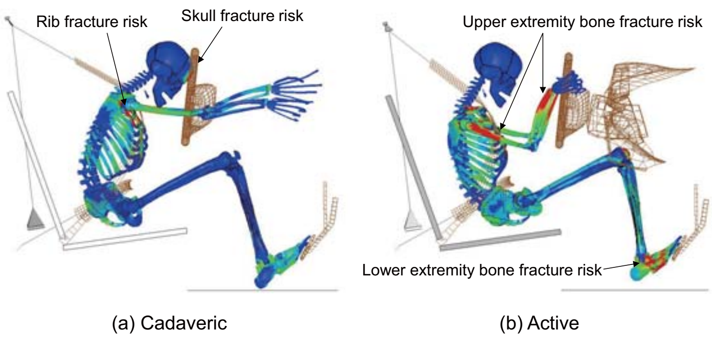


Fig. 11 Comparison of stress distributions between cadaveric human model and active human model.

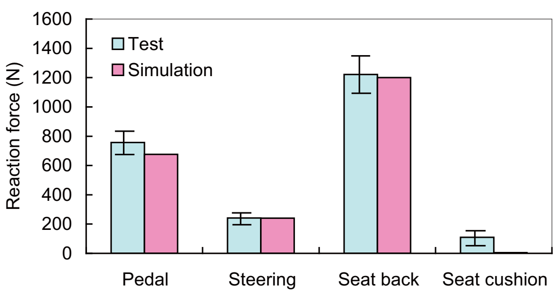


Fig. 10 Comparison of reaction forces between simulation results and volunteer test data.

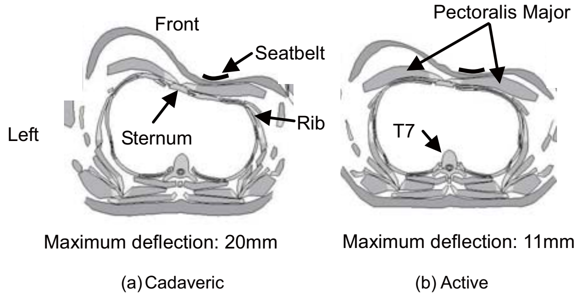


Fig. 12 Comparison of the maximum rib deflection between cadaveric human model and active human model.

maximum rib deflection predicted by the active human model was less than that predicted by the cadaveric human model. This is because Pectoralis Major muscle became stiffer with the muscle activation and caused less deflection by the seatbelt in the active human model differently from the cadaveric human model.

Kallieris et al. (1995)[(24)] 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 50 km/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 in the accident cases 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. Injury outcomes predicted by the cadaveric human model and the active human model were similar to those reported by Kallieris et al. (1995).[(24)] Although the simulations were performed under a condition with rigid seats and a 3-point seat belt and without any airbags in this study, the muscle activity is critical for better understanding of occupant injury mechanisms in automotive collisions.

5. Conclusions

An active human FE model with 3D geometry of muscles was developed to simulate driver's bracing and evasive maneuver in pre-crash and investigate muscular effects in pre-impact for human body kinematics and 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 muscle model also reproduced arm flexion motion and was used for simulating steering motions.

This study investigated the bracing effects in pre-impacts for human body kinematics and injury outcomes in frontal impacts by frontal impact simulations with pre- and post-impacts using the developed human FE 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 injury risks in upper and lower extremities. On the other hand, the muscle activity could reduce head and thoracic injury risks. These findings correspond to conclusions from comparison of injury outcomes between real-world accident data and cadaver test data with the same speed of 50 km/h. Although the simulations were performed under a condition with rigid seats and a 3-point seat belt and without any airbags, the model has possibility to make the detailed investigation of muscular effects in pre-impact for human body kinematics and injury outcomes. 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. However, the developed active human FE model would be a unique and useful tool for better understanding of unexplained injury mechanisms in real-world automotive accidents.

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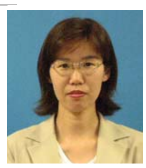

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Research Fields:
- Impact Biomechanics
- Computational Biomechanics
- Human Modeling
- Crash Safety

Academic Societies:
- The Japan Society of Mechanical Engineering
- Society of Automotive Engineers of Japan