PREDICTION OF PRE-IMPACT OCCUPANT KINEMATIC BEHAVIOR BASED ON THE MUSCLE ACTIVITY DURING FRONTAL COLLISION

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ABSTRACT

The objective of this study is to predict the behaviors of the human body in pre-crash conditions based on the experiment with active human models. In order to simulate the actual pre-crash condition of a car that occurs when the drivers brakes or pre-crash safety system activates in an emergency situation, low speed front impact tests on human volunteers were conducted using a sled-mounted rigid seat, on which each subject sat, sliding backwards on the rails. It was observed that when the subject's muscles were initially relaxed, muscle responses started activation at around 100ms after the onset of acceleration and reached its maximum value at around 200ms. During this time period, most of the individual body region acceleration responses and restraint system reaction forces also peaked. Furthermore, the head-neck-torso kinematics was strongly influenced by the muscle activity. This experiment indicates that muscles can react quickly enough to control the driver's behavior significantly during the low-speed impact, relating to the driver's posture just before the collision. Thus, the active human model with the Hill-type multi-bar muscle was employed to estimate the possible driving posture in an emergency. From the result of this experiment, pre- and post- crash occupant behavior was predicted. For a more detailed understating, a parametric study was conducted that distinguishes the factors presented in real accident cases.

INTRODUCTION

In the discussion of automobile crash safety, it is usual that occupant safety is discussed with a 50% adult male (AM50) or Anthropomorphic Test Dummy (ATD) in a normal sitting posture. However, the posture of the driver's seat occupant varies due to age, gender and physiques. Moreover, further posture changes will occur just before the collision due to occupant evasive maneuvers; thus, it is difficult to keep a normal position just before the collision. **Figure 1** shows the accident type and evasive maneuvers obtained from the Institute for Traffic Research and Data Analysis (ITARDA) in Japan (1993-2004)[1]. This data source consists of 860 of front impact collision (CDC:11F-1F) cases. The accident analysis show that around 50% of the drivers made evasive maneuvers in each accident type; namely, most drivers made evasive maneuvers just before the collision. In addition, driver injury incidence rates of chest with evasive maneuver are relatively larger than those with non-evasive maneuver when compared in each delta-V [2]. Therefore, evasive maneuvers include additional factors such as posture changes and body movement, which must be taken into consideration in discussing the accident analysis results. However, these differences in the driving posture and behavior of the occupant before such collisions is unreadable information from the accident data; thus, it is difficult to quantify the differences in injury mechanisms only from the accident data.

Armstrong et al.[3] examined the bracing effect in their review of prior works on muscle effects. They employed volunteers and demonstrated the strong influence of leg bracing against the toe board. It was found that 55% of the subject's kinematic energy absorption was attributed to the restraint by legs. Cross et al.[4] studied how passengers "brace" and react during pre-impact vehicle maneuvers. This information is related to the real world occupant photographic studies (Bingley et al.[5]). Parkin et al.

[6] studied the effects of driving posture and passenger individuality. It was concluded that the configuration of initial posture, i.e., how people sit in cars, showed marked differences between male and female. According to these previous studies, the causation between these differences and the injuries seen in the accident data has not been clearly explained. Nor has the relationship between evasive maneuver and the amount of posture change or muscle response been quantified yet.

The authors have studied the posture changes caused by pre-braking by means of volunteer tests. The previous study in this series [7][8] was conducted. In this study, the posture of the driver at the moment of pre-braking just before the impact was examined in two different muscle conditions. In one condition, muscle is fully relaxed, and in the other muscle is fully tensed. At the same time, the basic data of posture changes and muscle activation were also measured by using a 3D motion capturing system, and muscle activation electromyography, respectively. Based on the results of this experimental study, the prediction of the driver's posture and posture maintenance mechanisms were investigated. Moreover, this experimental study was applied to the injury prediction approach by using a computer human model to verify the influence of human body posture changes on the occupant injuries in a traffic accident. This in turn leads to further improvement of the effectiveness of occupant crash protection measures such as smart restrain system or adaptive restrain system in the accident. The aim of the current study is to build an injury prediction approach to verify the influence of human body posture changes on the occupant injuries in a traffic accident.

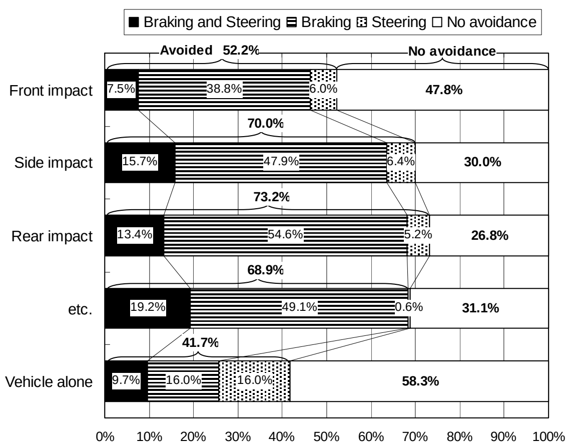

Figure 1. Accident type and evasive maneuver

VOLUNTEER EXPERIMENT

Volunteers and informed consent

Five 22 to 26 year-old male volunteers in good health participated in the series of experiments. The average and standard deviation of the 5 volunteer's anthropometric data is shown in **Table 1**. The protocol of the experiments was reviewed and approved by the Tsukuba University Ethics Committee, and all the volunteers submitted their informed consent in a document complied with the Helsinki Declaration.

Table 1.
Volunteer data

	Age (year)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Sitting height (cm)
Mean	24.1	168.9	61.8	21.5	87.4
±SD	1.5	7.8	11.1	2.6	4.1

Sled apparatus for simulation of low-level impact

Figure 2 illustrates the initial view of the front-impact simulation sled system (hereafter referred to as "Mini Sled"). The front pre-impact sled was designed based on the actual car pre-impact condition in order to simulate the deceleration experienced when the driver brakes or pre-crash safety barking system activates in an emergency situation. The sled is equipped with rigid seat made of steel (hereafter referred to as "R-seat"). Low-level frontal pre-impact was applied to the volunteer by accelerating the sled. The R-seat made of steel was mounted on the sled. In this experiment, a foot plate and a removable steering is equipped on the sled and the reaction forces coming from the plate and wheel were measured by load cells.

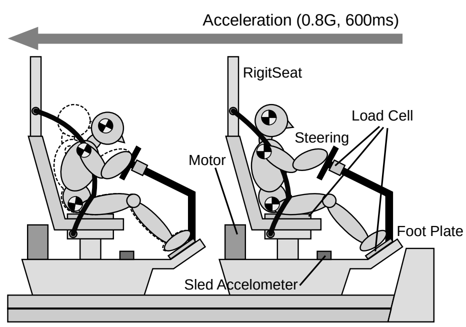

Figure 2. Outlook of the front-impact simulation sled system

Motion analysis and definition of joints and segment region of the full body

The physical motion of the human body and head-neck-torso kinematics at low-level impact accelerations were measured by means of the three-dimensional motion capturing system[9]. The features of this capturing system are that the position of each marker is automatically extracted from a video image caught with several cameras, and the skeleton image which builds up with these markers is translated into three-dimensional coordinates. In the experiment, eight sets of cameras were employed and the landmarks were attached to the defined body locations. The arrangement in this experiment

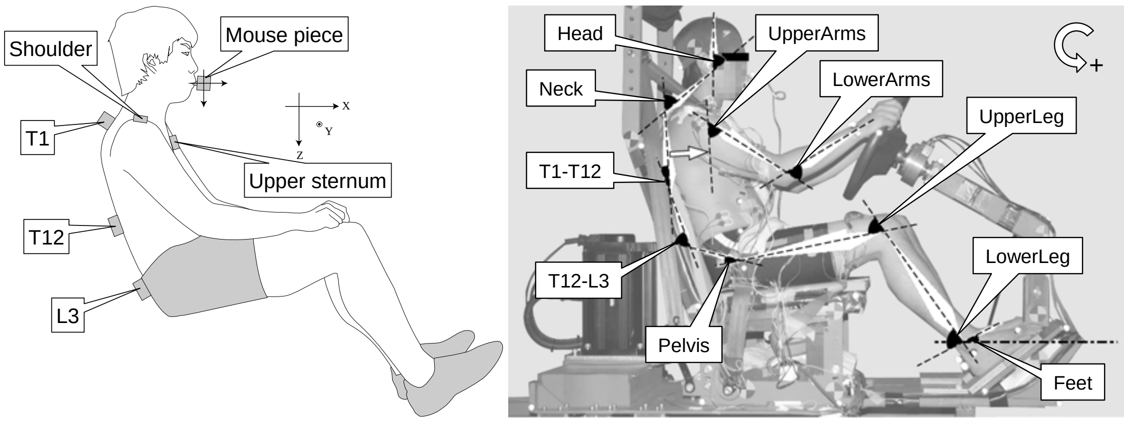

Figure 3. Lateral view of the head/neck/torso/pelvis with mounted accelerometer and definition of the segment and rotational angle between each segment

included landmarks attached to the head (Parietal, Auditory Meatus), shoulder (Acromion), chest (Sternum), back (T1, T11), lumbar (L3), arm (Elbow), hand (Wrist, Back), and leg (Knee). These markers were used as the reference points in order to determine of the head, neck, torso, abdomen, hip and lower extremity locations. From this motion data, a skeleton image is generated based on segments determined by body surface landmarks. **Figure 3** shows the definition of each segment of the head, neck, torso, abdomen, thigh, and the lower legs. With these motion segments, the rotational angle at the joint was recorded, and the differences between subsequent rotational angles were calculated. In order to represent the hip motion separately, a virtual marker was created based on the skin surface marker. To be more precise, the upper torso was separated into five segments (Head, Neck, Chest, Abdomen, Pelvis), and the joint angle at each connection point (Head, Neck, T1-T12, T12-L3, Pelvis, Upper Leg, Lower Leg, Upper Arms, Lower Arms) was calculated with the motion capturing software.

Acceleration measurement

In order to monitor the motion of the volunteer at the time of impulse, accelerometers were placed both on the body surface and the sled. Since the head motion was three-dimensional, tri-axial accelerometers and a tri-axial angular velocity meter were attached to the mouth via the mouthpiece, the first thoracic vertebra (T1), the twelve thoracic vertebra (T12) and the lumber vertebra (L3). The fixtures shown in **Figure 3** were fabricated for the installation of accelerometers on the body of each subject. The acceleration of the shoulder and chest were measured by the tri-accelerometer attached to the surface of the acromion and the front chest around the sternum region with a surgical tape, over which double-coated tape was adhered.

Electoromyography

Muscle activity was measured by means of surface electromyogram, the timing of which was synchronized with the three-dimensional movement data. EMG electrodes were attached to the skin over the major muscles of the subject. **Table 2** indicates the locations of the surface electrodes, and "M." stands for muscle. The measured muscle activation during the impact was analyzed by systematic processing, and the average rectified value (ARV)[7] was obtained. Each muscle response was normalized by its own maximum muscle activation value (ARV) in the tensed case.

Table 2.
Location of the muscle

Neck	M. Sternocleidomastoideus M. Paravertebralis
Torso	M. Latissimus Dorsi M. Erector Spinae
Abdomen	M. Rectus Abdominis M. Obliquus Externus Abdominis
Lower Extremity	M. Biceps Femoris M.Rectus femoris M. Gastrocnemius
Upper Extremity	M. Biceps Brachii M. Triceps Brachii M. Deltoideus

Experimental conditions

Five healthy males were selected as test subjects. In order to examine the effect of muscle activity on the physical motion in the pre-crash situation, the experiments were conducted in two conditions: a relaxed state, in which the volunteers were subjected to the impact in the state of relaxed muscles, and a tensed state, in which volunteers intentionally tensed their muscles. Test subjects were instructed so that they could assume each of these muscle configurations. During the test, the muscle activation was monitored to determine to what extent the subjects were relaxed or tensed. In the relaxed case, the subjects were required to be fully relaxed until the body motion was naturally stopped. For the safety of the subjects, a seatbelt was used to immobilize the waist of each subject. On the other hand, in the muscle-tensed cases, the subjects were instructed to tense their all muscles intentionally. In this case, a steering was installed to reconstruct the pre-crash condition in which the driver tenses their muscle to hold the steering wheel for bracing. The 3-point seatbelt was also attached to the shoulder to prevent the contact between the upper torso and the steering. The seatbelt was adjusted to the length of the abdominal and chest regions; thus, these belts were not pre-tensioned at the initial stage. For the purpose of comparison, the subjects were solicited to try to maintain their initial posture by tensing their muscles. Applying the acceleration to the sled while the subjects were assuming the same initial posture, the differences due to muscle activation could be clearly seen in the motion of their upper torso.

EXPERIMENTAL RESULTS

Subject's motion, acceleration response and EMG

In order to investigate the effect of the muscle condition, a series of experiments were conducted on the five volunteers in each case as shown in **Table 3**. The impact phenomenon seen in the typical frontal collision case can be described by the motions observed by three-dimensional movement analysis system, the acceleration at each region of the subject, and the electromyographic response. A subject's motion, acceleration response, and EMG are divided into four phases.

Table 3.
Test Matrix

5 adults

Impact acceleration	Direction	Muscle condition	Boundary condition
0.8G	Front	Tensed	Shoulder belt Lap belt Steering
		Relaxed	Lap belt

In this section, the results of the experiments conducted with an impact acceleration of 8.0 m/s^2 using a rigid seat are described. The following results were summarized according to the two different muscle conditions of pre-crash acceleration as time sequential changes. Further explanation of pre-crash conditioning in the two different muscle conditions are described in **Figure 4-7**. **Figure 4** and **Figure 5** show the sequential images of a subject's motion taken both by the high-speed camera and by the 3D motion capturing system. In addition, subject's motion, response to the acceleration and EMG are divided into four phases as illustrated in **Figure 7**. This figure shows the time histories of resultant acceleration and angular velocity of the head, T1, T12, and L3. In addition to the acceleration, the time histories of reaction forces with the belt, footplate, and steering wheel are shown. Moreover, the time histories of EMG response of each muscle of the subject are indicated. Finally, the time of acceleration onset is set at zero (0ms) in the time history diagram.

Motion observed by sequential picture images and 3D movement analysis system (Forward motion kinematic) - Because of the inertial force generated by the acceleration of the sled, the subject's upper torso started to move forward, while the head was moving backward relative to the first thoracic spine (T1) in the relaxed case. As a result of this phenomenon, the neck that links the head and torso started to extend. Moreover, T10 and T1 showed a ramp-up motion. With the subject's the body trunk restrained to the seat by a lap belt, the arched rotation of the upper torso started at around 200 ms. Simultaneously, the neck also started to rotate (flexion). Compare with the muscle-relaxed case, the upper torso motion is constrained by the muscle activation in the muscle-tensed case. With the subject's body trunk restrained to the seat with the reaction force from the steering wheel through the arm, the forward motion of the upper torso started to fold back at around 200 ms.

Acceleration at each region of the subject and loads - The constant acceleration of 8.0(m/s^2) was applied to the sled, and this acceleration appeared with the acceleration of L3 close to the sled in initial stage. Then, the lumber acceleration was transferred to the head of T12 and T1 one by one. The magnitude of L3 and T12 acceleration indicated the maximum value at around 150 ms, and the head and T1 acceleration gradually increased due to the forward motion of the upper body. According to the angular velocities of the L3, T12, T1, and Head, each portion started to rotate at around 100ms. Then, the L3, T12 angular velocity was getting decreased, and the magnitude of T1 and head C.G. angular velocity reached the maximum value at around 300ms. Compared to the muscle-relaxed case, the upper torso motion is constrained by the muscle activation in the muscle-tensed case. Therefore, the maximum value of acceleration, angular velocity, and the seatbelt force decreased. The maximum value of T1 and head were indicated at around 200ms. In addition, the flexional angular velocity of the head and neck reached maximum at around 150 ms, when the head-neck deformation started to fold back. The magnitude of this acceleration decreased due to the activation of the muscles after 200 ms. On the other hand, L3 angular velocity converged to zero, and the magnitude of T1 and head angular velocity reached a positive value (Extension). The angular velocity indicates a miner oscillation mode because the shoulder-belt and lap-belt started to react with the forward motion of the torso. The magnitude of foot plate and steering load indicated the maximum value at around 200 ms, and these reaction forces are strongly correlated to the muscle activation at lower extremity and upper extremity. The reaction force of footplate and steering is converged to 100 N by discharging the lower and upper extremity muscle, when the subject found an appropriate balance between the muscle activation and the inertia effect (**Figure 5**). Therefore, the acceleration and angular velocity converged to zero.

Electromyographic response- Because of the muscle-relaxed condition, the upper torso started to move forward, and the major muscle activation was not detected except M. Erector Spinae, M. Biceps Femoris, and M. Gastrocnemius. In relation to the neck link motions, discharge of M. paravertebralis slightly activated at around 100 ms. Moreover, the position of the body trunk moved forward, and M. Biceps Femoris and M. Gastrocnemius normalized ARV value are rapidly increased at around 150ms. Then, the muscular discharge of the neck, torso, and leg disappeared.
In the muscle-tensed case, the body trunk was restrained to the seat due to the muscle activation. Therefore, the acceleration started to decrease and the angular velocity of each body region converged to zero. The muscle activation is discharged continuously, and the posture is maintained by the resistance force from the footplate and the steering wheel. Regarding the muscle response, most of the muscles are discharged before the impact (0ms) except M. sternocleidomastoideus and M. paravertebralis .

Differences in HEAD, NECK, and TORSO Motions related to the muscle responses and restrain effect

It has been detected that the pre-impact tension of muscle affects the physical motion at low-level impact, and this muscle effect is mostly related to the rotational angle of head, neck, and torso. Therefore, the rotational motion of the upper torso was analyzed based on the trajectory of each landmark measured by the 3D motion capturing system. **Figure 8** shows the average values of the maximum flexional and extensional angles at the joint. The average value was calculated from the data of the five male volunteers. For the purpose of comparison, the tensed and relaxed muscle cases are shown in these figures. As for the rotational angle of each joint, the primary value was set as zero (0). The plus (+) direction indicates extension, while the minus (-) direction

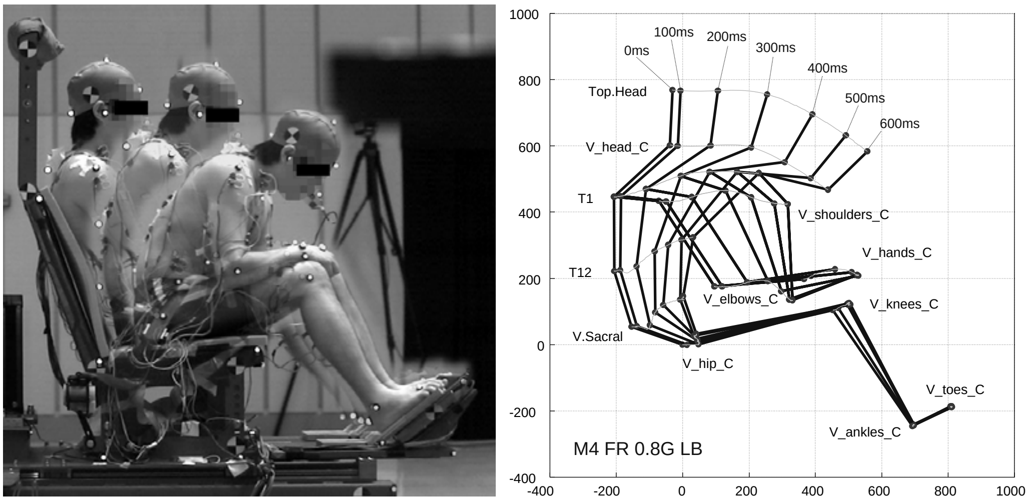

Figure 4. Physical motions from the 3D motion capturing system (Male, 0.8G: Relaxed)

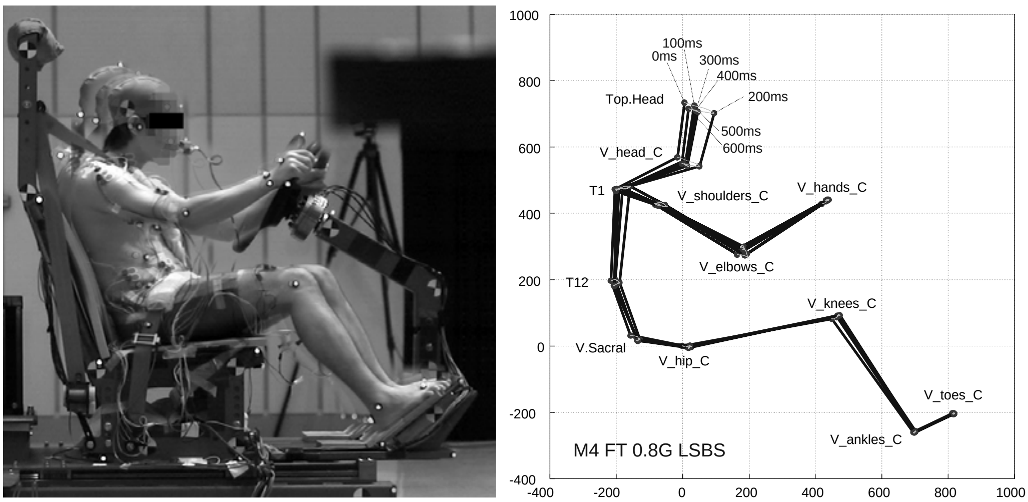

Figure 5. Physical motions from the 3D motion capturing system (Male, 0.8G: Tensed)

indicates flexion. Because of the muscle activities, the major angle difference between the lower leg and the feet could not be identified in the experiment. In the muscle- relaxed case, the flexional motion of the pelvis (Pelvis) is dominated in the body trunk. The extensional motion of the head (Head) and lower extremity (Upper Leg) were identified in the muscle-relaxed condition.

On the other hand, in the case of tensed condition, the major flexional motion was detected in the neck (Neck), thorax (T1-T12), and upper extremity (Upper Arms) area. The head (Head), pelvis (Pelvis), lower extremity (Upper Leg) and upper extremity (Lower Arms) showed extensional motion.

Muscle activities during pre-braking

According to the kinematic and muscle activity in the experimental results in **Figure 4-7**, the subject's upper torso and head-neck motion is constrained by the muscle activation. In addition, the torso motion is restrained by the resistance force from the upper extremity which is connected to the steering wheel. As a result of this phenomenon, upper torso motion was strongly affected by the muscle and the boundary condition in the pre-crash situation. **Figure 9** indicates the average value of the integrated normalized ARV to define the each muscle activation during the impact. This value is calculated from the integration of average time history of normalized ARV value with five volunteers in both relaxed and tensed case. The interval of integration is between 0ms to 600ms when the sled is moving in the constant acceleration. In the muscle-tensed case, most of the muscles become larger than the relaxed case except M. Rectus Femoris and M. Biceps Femorise. Particularly, M. Sternocleidmastoideus (Neck), M. Rectus Abdominis (Abdomen), M. Gastrocnemius (Lower Extremity) and Upper Extremity have a significant difference in the amount of integrated normalized ARV between muscle-relaxed case and tensed case.

The result of the integrated value indicates the muscle condition working against the forward motion during the impact. These muscle activities were strongly related to the motion of the upper torso. The limitation of the posture-control with muscle activation was identified based on the activation level.

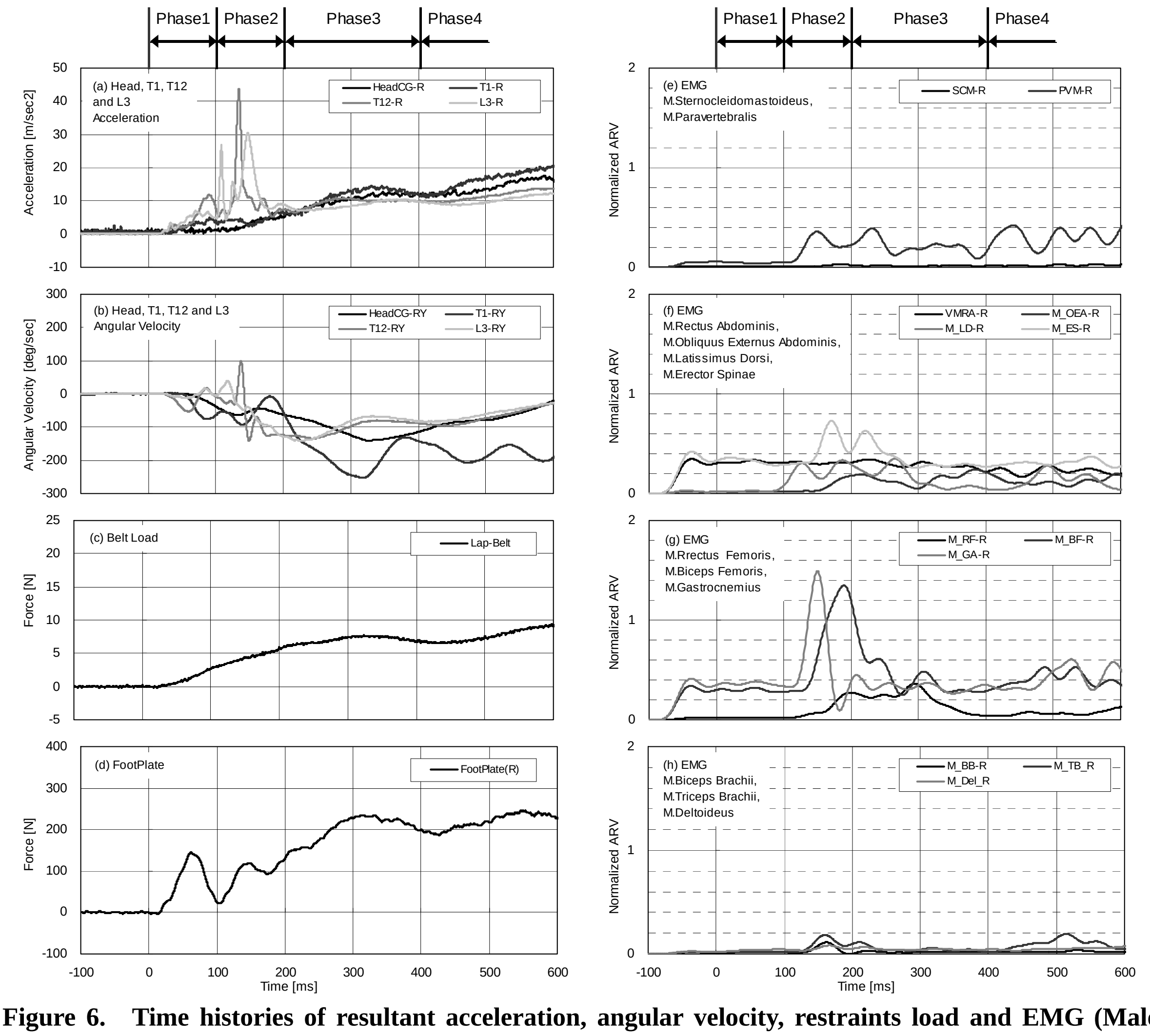

Figure 6. Time histories of resultant acceleration, angular velocity, restraints load and EMG (Male, 0.8G: Relaxed)

These muscle activations should be taken into account in predicting this pre-crash phenomenon with the computer model.

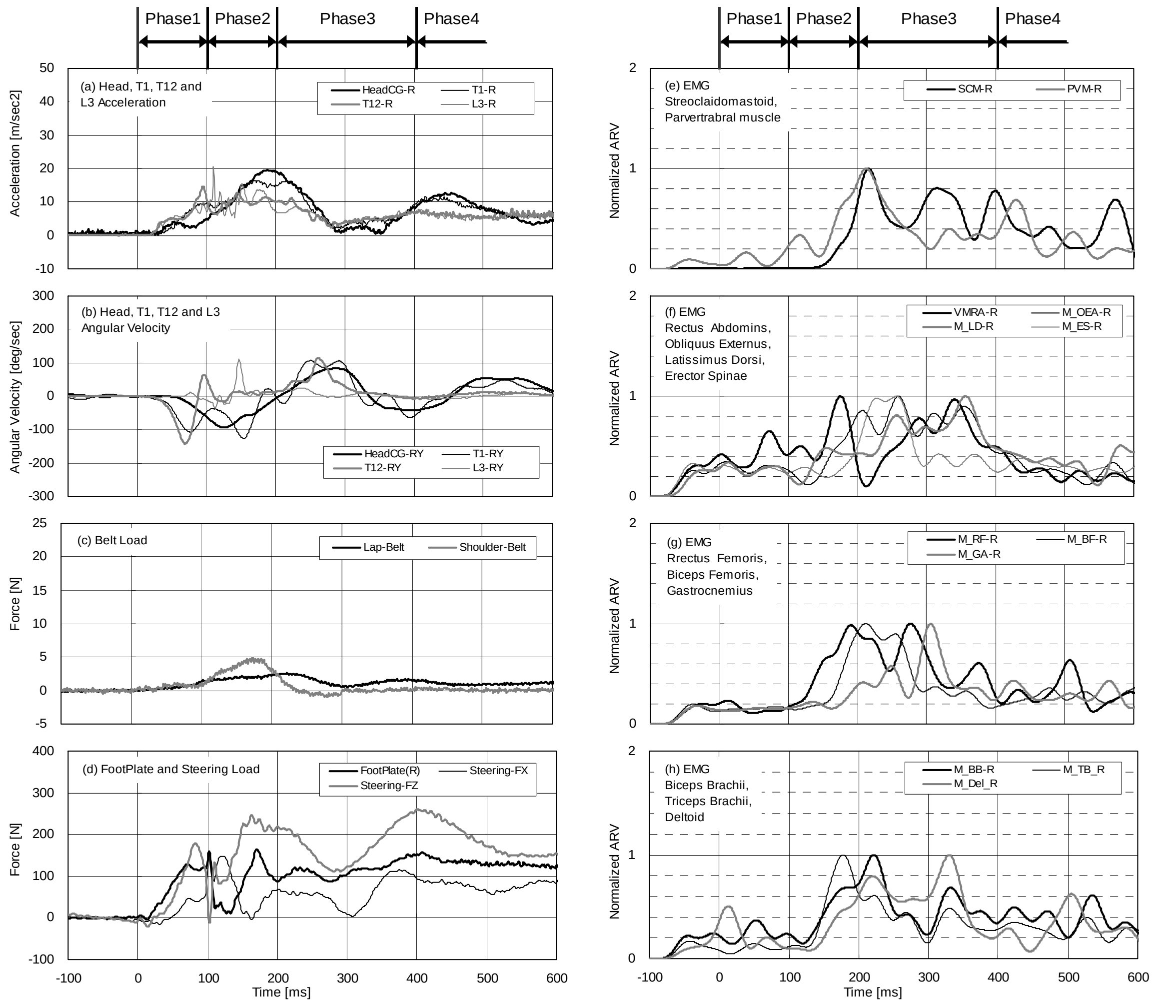

Figure 7. Time histories of resultant acceleration, angular velocity, restraints load and EMG (Male, 0.8G: Tensed)

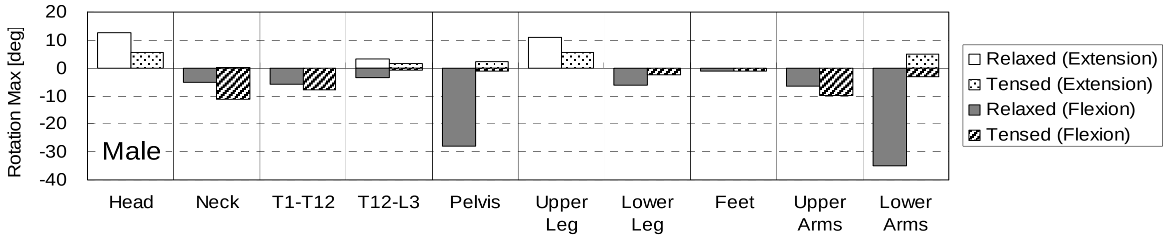

Figure 8. Maximum flexion and extension angle of each joint

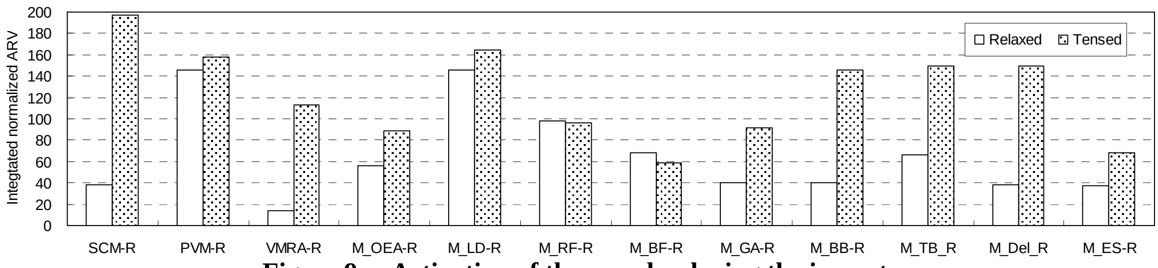

Figure 9. Activation of the muscles during the impact

NUMERICAL STUDY

Prediction of impact posture changes due to pre-crash condition

Active human model –It can be said that the experimental result indicate that the physical motion of the human body under pre-crash condition can be predicted. Therefore, the effect of the posture change caused by pre-crash condition was estimated by using MADYMO TM [10] with a multi-body adult male as the occupant model (**Figure 10**). This computer model is based on a rigid body model that incorporates muscle effect by using the hill-type muscle model. The muscle repose measured from the acceleration experiments was directly applied to the major muscles in order to simulate this acceleration's effect on the human body. To populate the design space of occupant anthropometric properties, the scaled rigid body model is developed by using GEBOD [10] which is constructed by the human anthropometric data. To examine the reliability of the MADYMO occupant model, the current human model was validated against the results of the experiments [11]. The validation was done against the impact tests at 8.0 m/s^2 in muscle relaxed and tensed cases which are described in the previous section, and it included the kinematics of the head-neck complex and whole body motion. Therefore, the validation of the modeling of these impacts was done only in terms of the posture change during the low-speed impact.

Baseline model -The baseline model used in this study was a multi-body representation of the driver's side interior compartment of a mid-size sedan car. The vehicle interior consists of the standard three-point belt system, a steering wheel, a knee bolster, and a footplate (**Figure 10**). The mechanical property of each component is validated with the experimental study. In this study, the accident scenario was reconstructed to evaluate the effect of posture change in the pre-crash condition. Therefore, the pre-crash phase which was defined in the volunteer test (constant deceleration 8.0 m/s^2: Duration: 0-600ms) is considered just before the collision and the deceleration pulse taken from the barrier test with mid-size sedan car (delta-V of 50 km/h) is applied to the model.

Sensitivity analysis

Design parameter and injury criteria -Three design parameters are mainly related to posture change in the pre-crash condition. In the previous study on the volunteer test [7], the small female subject showed the different trajectory from the male. Therefore, the anthropometric size should be one of the parameter in sensitivity analysis. Bose et al. [12] proposed the several kinds of initial posture based on the computer model, and it showed large difference in the injury outcome from the computer simulation. For this reason, the muscle response, anthropometric size, and initial posture that are considered as deign parameters in this simulation, and this study calculate the contribution of each parameter to the injury outcome. The head injury criterion (hereafter referred to as HIC_{36}) and chest acceleration for 3ms (hereafter referred to as C_{3ms}) were employed for the evaluation of injury. The variation of each of the design parameters is listed in **Table 4**. The combination of the muscle response, anthropometric size, and initial posture are carried out, and the total number of simulations is eighteen cases. The variation of each parameter is described as follow. The muscle response is defined as two condition from the experimental result (Tense and relax). The anthropometric size is based on the Japanese anthropometric male database and defined as three types (AM05, AM50, AM95). The initial posture is referenced to the literature [5][6][11] and three kinds of posture is defined (STD: Standard, UPR: upper body upright, FOW: upper body forward to steering column) .

Sensitivity Evaluation

The sensitivity information of each parameter is calculated by the analysis respect to the HIC_{36} and C_{3ms}. **Figure 11** shows the effects on individual injury values by changes of design parameters from the HIC_{36} and C_{3ms}. The HIC_{36} is sensitive both to the anthropometric size and to the muscle response. **Figure 12** indicates the eighteen cases of HIC_{36} respect to the anthropometric size (AM05, AM50, and AM95) in muscle relaxed and tensed case. In this figure, HIC value is normalized by the output value from the standard condition (Muscle Tense, AM50, STD). Compared with the muscle-tensed case, the variation of HIC_{36} is larger than that in the muscle-relaxed case. The reason for this difference is that the pre-crash phase defined in this simulation causes not only a posture change but also a change in body velocity due to inertial forces. Therefore, the contact speed between the head and the steering wheel changes in the muscle-relaxed case. On the other hand, the C_{3ms} is very sensitive to the initial posture. **Figure 13** indicates the eighteen cases of C_{3ms} respect to the initial posture (STD: Standard, UPR: upper body upright, FOW: upper body foward to steering column) in muscle relaxed and tensed case. The value of C_{3ms} is also normalized by the output value from the standard condition (Muscle Tense, AM50, STD). Compared with the muscle tensed case, large variation C_{3ms} can be detected in the muscle relaxed case. This phenomenon is also related to the pre-crash phase defined in this simulation, and the difference of initial posture affects the relative chest velocity to the steering. Therefore, the 3ms criterion value of the chest deceleration variation is increases. From this calculation, the effect of pre-crash phase is predicted by using the computer human model. For a more detailed understating of the mechanisms, further study will be needed to distinguish the parameters that are present in real accident cases.

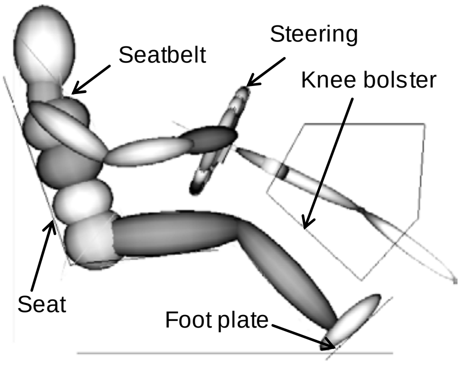

Figure 10. Computer human model with the driver's side interior compartment of mid-size car

Table 4.
Variation of design parameter

	Muscle response	Anthropometric size	Initial posture
18 case	Relaxed	AM05	STD
		AM50	UPR
	Tensed	AM95	FOW

DISCUSSION

Mechanisms of posture changes during pre-crash condition

The estimation from the results of measurement system indicates a significant correlation between the discharge of muscle force and the kinematic of each body part. For example, the head-neck-torso acceleration (HeadCG, T1, T12, and L3) increases, but decreases when the volunteer intentionally tensed their muscle. The muscle-tensed effect is clearly detected in the magnitude of acceleration and angular velocity compared to the muscle-relaxed condition. This is not only due to the back and abdominal muscles, but also due to the upper and lower extremity muscles discharged from the impact (0 ms). In other words, the upper torso was subjected to posture-control provided by these pre-tensed muscle condition in which activation level is around 20-40 % of maximum muscle. Following the timing of the muscle activation with upper and lower extremity, the steering wheel and the footplate show the reaction force continuously. Thus, the subject found the appropriate balance to control the upper body motion by using the reaction force from the steering and the footplate.
In the relaxed case, the subjects were required to be fully relaxed until the body motion was naturally stopped. However, a natural muscle 'stretch receptor' activated and this muscle activation temporarily seen in the back and lower extremity to control excessive motion. This protective mechanism works more effectively when the steering is installed in the system.

Effect of muscular Tension

In this study, a steering was installed to constrain the hip and chest in order to simulate real pre-crash conditions. It was identified that the pre-acceleration tension of muscles exerted influence on the physical motions compared to the relaxed case; however, this effect greatly reduced from that detected in the cases of relaxed cases. In comparing rotational angles between tensed and relaxed cases per body region from pelvis to head, ante-extensional motion due to muscle tension was detected at the neck region (Head: **Figure 8**). In this region, the posture-control effect of the rotational angle due to muscle tension was around 60%. On the other hand, the hip region (Pelvis: **Figure 9**) shows slight flexional rotational motion in tensed cases. In the relaxed cases, the hip region showed the largest flexional motion. Consequently, it was detected that the rotational angle of hip region was strongly affected by the upper torso motion restrained by the steering in the front impact case. Therefore, the boundary condition effect is important in discussing the stability of the posture under low speed acceleration.

Prediction of the effect of posture change for the injury

From the result of sensitivity analysis of the muscle response, anthropometric size and initial posture were selected as design variable. Injury values such as HIC_{36} and C_{3ms} were sensitive to the anthropometric size and initial posture respectively. However, these results strongly affected by the muscle condition and the different tendency are shown in **Figure 12** and **Figure 13**. As for the head injury criteria, the HIC_{36} value increases when the anthropometric size gets larger in the muscle-tensed case, and this is because of the inertia effect caused by the occupant's mass. On the other hands, the relaxed effect is clearly detected in the muscle relaxed case. For example, when the initial posture is set as FOW (upper body forward to steering column), HIC_{36} value decreased even though the occupants mass increased. This is because of the distance between the head and the steering wheel. In the pre-crash phase, the occupant upper torso inclines to the steering, and the head is almost attaching to the steering wheel when the crash deceleration is applied. Therefore, the contact speed when the occupant hits their head to the steering is almost close to zero.

As for the 3ms criterion value of the chest deceleration (C_{3ms}) value tends to increases when the initial posture (STD: Standard, UPR: upper body upright, FOW: upper body forward to steering column) is close to the steering in the muscle-tensed case and this is because of the contact between the chest and the steering wheel during the impact. In the muscle-relaxed case, because of the difference of the initial posture, the relative speed between the chest and the wheel is changed during the impact. This speed depends on the balance between the belt force and the inertia force generated by occupant's mass. The sensitivity analysis provides a guideline about the effects on injury levels by changing the design parameters. This is the preliminary study of the effect of posture change with active human model. For more detail analysis, several parameter studies are needed to understand the mechanisms of posture change during the impact.

Limitation of this study and suggestion for further research

The number of the subjects in this study was limited. Therefore, the data were insufficient to discuss the difference between tensed and relaxed muscle condition. In addition, modeling of occupant motions of the body is necessary to solve the muscle cooperation problem and the solution of this problem is to activate the muscle model. For reliable qualitative validation of the model, it is necessary to analyze the relationship between the kinematic and muscle activation in detail in order to obtain the information of muscle effect. This could be done in a co-operation between the tests and simulations.

CONCLUSION

The result of this study concludes that the effects of muscular tension on each body motions have been clarified, and the physical motion of the driver side occupant is predicted in the pre-crash condition. Furthermore, it has been identified based on acceleration, EMG electrodes and the reaction forces that differences in muscle activity govern the motion of the body in each phase. Finally, it has been found that the muscles that most highly activated when the occupant made a pre-braking action were the neck and abdominal muscles. These parameters are important factors in discussing the subject's motion with the restraints system just before the collision. In addition, the steering also supports the driving posture and stabilizes the pelvis motion. The present human body model adequately represents the general kinematics of the physical motion detected in

Figure 11. Effect of design parameter changes on injury value

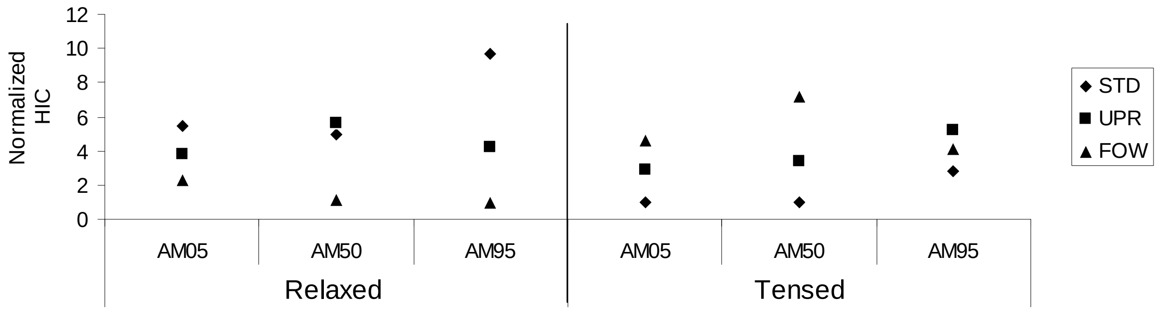

Figure 12. Distribution of normalized HIC_{36} respect to the anthropometric size

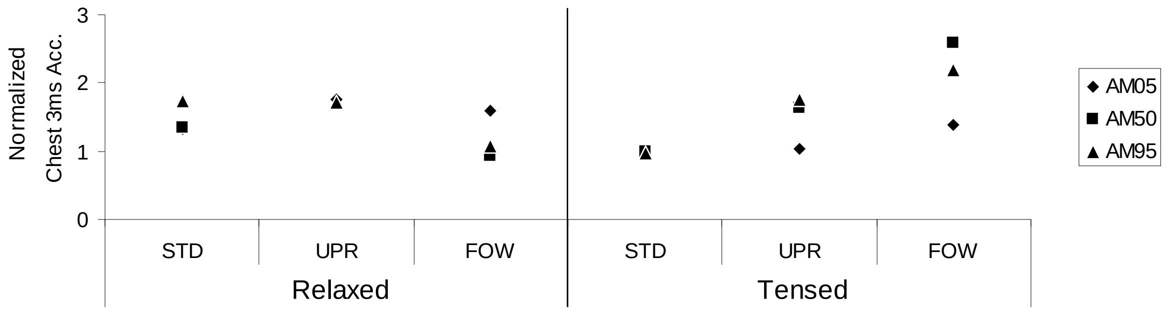

Figure 13. Distribution of normalized C_{3ms} respect to the initial posture

pre-impact braking conditions. This, in turn, indicates that the EMG data of major muscles significantly influences the physical motion, because these input variables are directly taken from the volunteer tests. This model is currently in the improvement phase, and its practical application and injury level prediction will be completed by using a finite element model in the next stage.

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