Prediction of the Physical Motion of the Human Body based on Muscle Activity during Pre-Impact Braking

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ABSTRACT

The purpose of this study is to predict the physical motion of the human body under pre-crash conditions. Low speed front impact tests on human volunteers were conducted using a linear-motor sled with the rigid seat, steering wheel and seat-belt. During the experiments, the subject's physical motion, acceleration, and EMG signals were recorded. When the subject's muscles were initially relaxed, muscle responses were observed to start activation at around 130ms after the onset of acceleration. Furthermore, the head-neck-torso accelerations were strongly influenced by the muscle activity after the impact. The major body region such as head, neck and pelvis were constrained by the related muscle.

Key Words: Biomechanics, Occupants, Frontal Impacts, Volunteers.

OCCUPANT EVASIVE MANEUVERING is an important factor that can affect the resulting injury. **Fig. 1** shows statistical analysis data of evasive maneuvers. The data source consists of 271 cases of front impact collision (CDC: 11F-1F). According to accident analyses (ITARDA, 2007) driver injury incidence rates of chest after braking are lower than those of non-evasive maneuver when comparing travel speed estimated by the driver (speed before braking). One evasive maneuver that most drivers make just before the collision is braking, and it generally serves to reduce the impact speed and also the injury risk. However, the accident data suggests that the presence of braking (evasive maneuvers) is linked to higher injury incidence rates when considering equivalent barrier speed or delta-V. This contradictory phenomenon suggests that braking serves to reduce the vehicle speed, however the passenger travel forward during braking and further posture changes occur due to inertial forces. Therefore, the evasive maneuvers include additional factors such as posture changes and movement which must be considered when we discuss the accident analysis results. It is predicted that these differences in the driving posture and behavior of the occupant before such collisions will in turn affect the injuries sustained by the occupant. This study offers the possibility of quantifying the differences in injury mechanisms evaluated by individual occupant or posture differences.

In their review of prior works on muscle effects, the bracing effect was examined (Armstrong et al., 1968). They employed volunteers and demonstrated the large influence of leg bracing against the toe board. It was found that 55% of the subject kinematic energy absorbed was attributed to the restraint by the legs. Richard et al. (2005) studied how passengers "brace" and react during pre-impact vehicle maneuvers. This information is linked to real world occupant photographic studies (Bingley et al., 2005). Parkin et al. (1995) studied the effects of driving posture and passenger individuality. It was concluded that the configuration of initial posture how people sit in cars shows 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 not yet been quantified.

Authors have studied the posture changes caused by pre-braking through the use of the volunteer tests. The previous study in this series (Ejima et al., 2007) was conducted with only a simple lap-belt, and the effect of the lower extremities was neglected in order to focus on muscle activities in the upper part of the body. In this study, the posture of the driver at the moment of pre-braking just before the impact was examined with 3-point belt system and a steering wheel. At the same time, the basic data of posture changes and muscle activation were also measured, using a 3D motion capturing system, and muscle activation electromyography, respectively. Based on results from this experimental study, the prediction of the driver's posture and posture maintenance mechanisms were investigated. The final goal of this study is to establish an injury prediction approach 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 in accident situations.

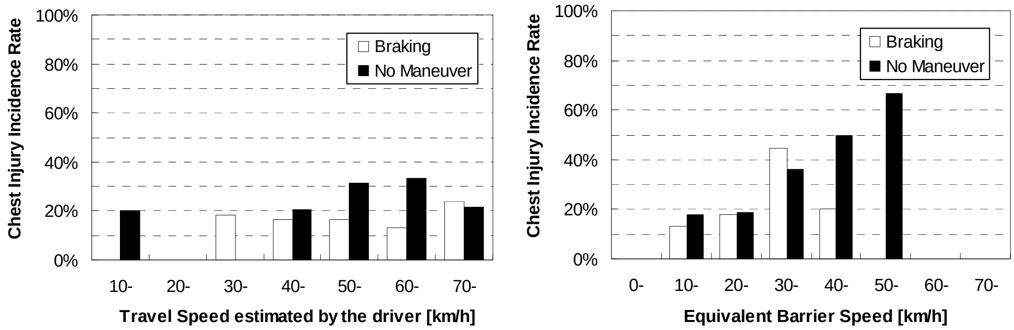


Fig. 1- Chest injury incidence rate by evasive maneuver

METHODS OF EXPERIMENT

VOLUNTEERS AND INFORMED CONSENT: Seven healthy 22 to 26 year-old volunteers (five males and two females) participated in the series of experiments. The protocol of the experiments was reviewed and approved by the Tsukuba University Ethics Committee, and all volunteers submitted their informed consent in a document according to the Helsinki Declaration. Basic physical data of these subjects are shown in **Table 1**.

SLED APPARATUS FOR SIMULATION OF LOW-LEVEL IMPACT: **Fig. 2** illustrates an initial view of the front-impact simulation sled system (hereafter referred to as "Mini Sled"). The front pre-impact sled was designed based on actual car pre-impact experiments, and a linear motor was installed in order to simulate the deceleration experienced when the driver brakes in an emergency situation. The length of the rail was 5.0 m, and the sled slides on the rails at a constant acceleration 8.0 m/s^2. The sled has a rigid seat (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, the reaction forces coming from the foot plate and steering wheel were measured by load cells. For modeling a real pre-crash condition, a 3-point seatbelt was used to constrain the hip and chest. In this case, the seatbelt was adjusted to the length of abdominal and chest region, thus these belts were not pre-tensioned at the initial stage. The muscles were conditioned to be relaxed or tensed, and several pre-tests were conducted to check the test repeatability under each condition (Siegmund et al., 2003).

Table 1. Subjects

Volunteer	Age (year)	Sex	Height (cm)	Weight (kg)	Sitting height (cm)
I	26	M	174.5	75.6	92.0
II	25	M	168.0	62.4	89.0
III	25	M	172.0	64.3	89.0
IV	23	M	173.0	70.0	85.0
V	25	M	178.0	65.0	91.0
VI	23	F	160.0	42.0	86.0
VII	22	F	157.0	53.0	80.0

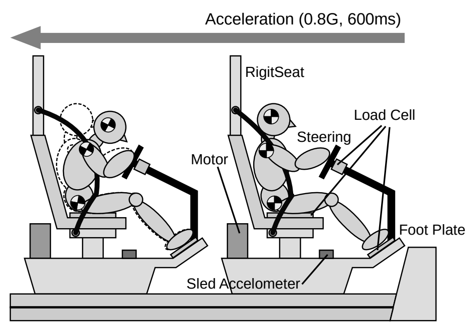


Fig. 2- Outlook of the front-impact simulation sled system

MOTION CAPTURING AND LANDMARKS: The subject's motion in this experiment is a three-dimensional movement within the X-Y-Z space. Thus, a three-dimensional motion-capturing device was used for the measurement of the body motion. The feature of this capturing system is that the position of each mark is extracted automatically from a video image caught with several cameras (Eagle Digital Camera) (NAC Inc.) and is translated into three-dimensional coordinates. The resolution of the camera is 1280 x 1024 pixels. The images were incorporated into EVaRT (NAC Inc.) and analyzed. This form of motion capturing technology commonly used in a variety of studies requiring the capture of human motions. In the experiment, eight sets of cameras were used and the landmarks were attached to defined body locations. The arrangement in this experiment 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 for the determination of the head, neck, torso, abdomen, hip and lower extremity locations.

ACCELERATION MEASUREMENT: An accelerometer was installed on the sled floor along the inclination of the rail. The sled velocity is calculated by integrating the acceleration of the sled. In order to monitor the motion of the volunteer at the time of impulse, accelerometers were placed 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 a mouthpiece, the first thoracic vertebra (T1), the twelve thoracic vertebra (T12) and the lumber vertebra (L3). The fixtures shown in **Fig. 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: The muscle activation levels of each subject in a relaxed state was measured preliminary to the tests in this experiment. 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. The electrodes had a diameter of 5mm, and were arranged as bi-polar electrodes with a distance of roughly 2cm between the electrode centers. The following muscles indicate the locations of the surface electrodes and "M." stands for 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 and two healthy females were selected as test subjects. In order to examine the effect of muscle activity on the physical motion, the experiments were conducted under two conditions: a relaxed state, in which the volunteers were subjected to the impact in a state of relaxed muscles, and a tensed state, in which volunteers intentionally tensed their muscles. Test subjects were instructed to assume each of these muscle configurations. During the test, the muscle activation was monitored to determine the extent to which 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. On the other hand, in the muscle tensed cases, the subjects were instructed to tense their all muscles intentionally. For the purpose of comparison, the subjects were asked to try to maintain their initial posture through the tensing of their muscles. Applying the acceleration to the sled with subjects assuming the same initial posture, the differences due to muscle activation could be seen clearly in the motion of upper torso.

Table 2. Test Matrix

	Impact acceleration	Direction	Muscle condition	Boundary condition
7 adult (5 male 2 female)	0.8G	Front	Relaxed	Shoulder Belt Lap Belt Steering
			Tensed	

ANALYSIS

DEFINITION OF JOINTS AND SEGMENT REGIONS ALONG THE FULL BODY: The physical motion of the human body and head-neck-torso kinematics at low-level impact accelerations were measured using the three-dimensional motion capturing system. From this motion data, a skelton image is generated based on segments determined by body surface landmarks. Each segment of the head, neck, torso, abdomen, thigh, and the lower legs are defined as follows and shown in **Fig. 3**. The definition of each segments are shown below Fig .3. 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. More precisely, 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, UpperLeg, LowerLeg, UpperArms, LowerArms) was calculated with the motion capturing software.

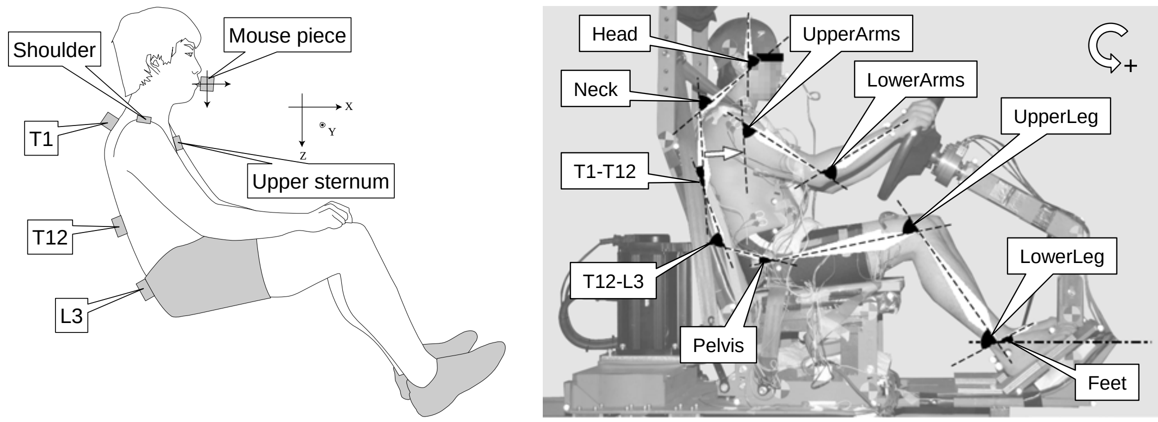


Fig. 3- Lateral view of the head/neck/torso/pelvis with mounted accelerometer and definition of the segment and rotational angle between each segment

- Head: the segment between a vertex and head C.G.
- Neck: the segment between head C.G. and the first thoracic vertebra
- Torso: the segment between the first thoracic vertebra and the 12th thoracic vertebra.
- Abdomen: the segment between the 12th thoracic vertebra and the 3rd lumber vertebra.
- Pelvis: the segment between the 3rd lumber vertebra and the trochanter
- Upper leg: the segment between the iliac crest and the knee joint
- Lower leg: the segment between the knee joint and the ankle
- Feet: the segment between the ankle joint and the toe
- Upper Arms: the segment between the shoulder joint and the elbow
- Lower Arms: the segment between the elbow joint and the wrist

HEAD ACCELERATION AND NECK LOAD: The location of the anatomic center of gravity of the head was determined using the methods reported by Ono et al (1997). The position of the head C.G. was located 5mm in front of the external auditory meatus and 20mm above the Frankfurt line which connects the lower orbital margin and the center of auditory meatus. Since the head motion was three-dimensional, 6-channel accelerometers comprising a combination of a tri-axial accelerometer and a tri-axial angular velocity meter attached to the mouth via a mouthpiece were used in the measurement. The shear and axial forces, and the bending moment acting against the upper region of the neck (occipital condyle) were measured using this acceleration data.

EMG SIGNAL PROCESSING: The major muscle activation during the impact was measured via surface electrodes and analyzed after systematic processing. At first, the raw EMG signals were filtered with a band pass filter (low pass filter: 500Hz, high pass filter 25Hz). Then, full wave rectification was applied to each signal. Finally, smoothing was applied and the average rectified value (ARV) was obtained. Each muscle response was normalized with their own maximum muscle activation value (ARV) in the relaxed case. The normalized ARV value is defined in each muscle. In this study, the muscle reflex time was defined based on the assumption that the criteria defining muscle activation is that the EMG signal after the onset of acceleration (time T=0) was more than 30% (i.e., more than 0.3 times the normalized ARV value). Additionally, the reflex time was always zero in the tensed muscle case, because the subjects had intentionally activated their muscles before the acceleration was applied.

RESULTS

INITIAL POSTURE: The initial driving posture is an important static factor when discussing the pre-crash conditions. Therefore the posture differences in the experiments were measured by the three-dimensional motion analysis system. During the test, volunteers were asked to assume their usual driving posture on the sled. **Fig. 4** indicate the measured angle at each joint (defined in the previous section) of the seven volunteers. For the purpose of comparison, the male and the female subjects are shown. When considering subject individuality, the variation of the pelvis and the lower arms angle are relatively large in the male subjects. On the other hand, the female subjects show most variation at head and lower leg positions. Particularly with the spine curvature, the female pelvis angle is smaller than male, but the thoracic (T12-L3) and head (Head) angle is larger than the male. As a result, the female is close to the steering wheel than the male in the experiment.

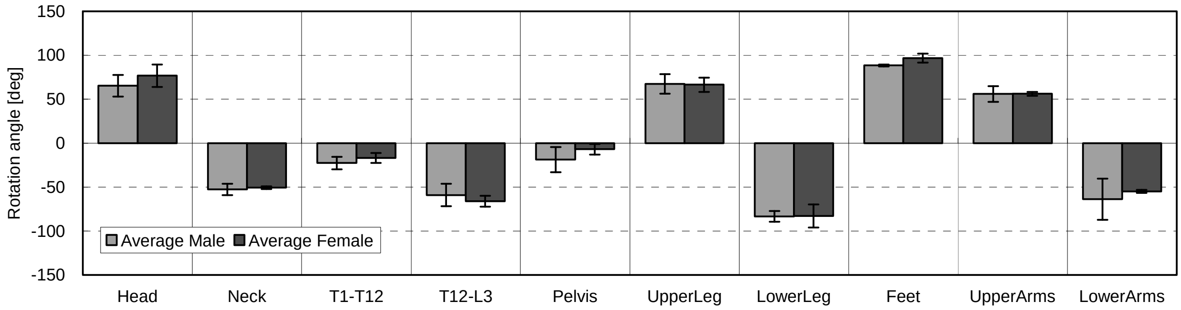


Fig. 4- Average and standard deviation of initial rotation angle at each joint

SUBJECT'S MOTION, ACCELERATION RESPONSE AND EMG: In order to investigate the effect of the impact level and muscle condition, a series of experiments were conducted on the seven volunteers in each case as shown in **Table 2**. 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. In this paper, only results of the experiments conducted with an impact acceleration of 8.0m/s^2 using a rigid seat are described. The following results were summarized according to each phase of pre-crash acceleration as sequential changes with time: 1) the motions observed by sequential picture images and three-dimensional movement analysis system respect to the sled, 2) acceleration at each region of the subject and loads, and 3) the electromyographic responses. Further explanation of pre-crash conditioning under two different muscle conditions are described in **Fig. 5-8**. **Fig. 5** shows the sequential images of a subject's motion taken by the high-speed camera and the 3D motion capturing system. In addition, subject's motion, response to the acceleration and EMG are divided into four phases as presented in **Fig. 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 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.

Phase 1 (0 - 100ms, Initial Response Phase): No significant motion was identified in this phase although the 8.0 m/s^2 acceleration was applied to the sled during this phase (**Fig. 5, 6**). This acceleration appeared with the acceleration of L3 and T12 in relax and tensed case (L3-R, T12-R: **Fig. 7, 8**) which was close to the sled. Then, no major acceleration was shown in this phase except for the L3 and T12. The major muscle activation was not detected in the monitored muscles.

Phase 2 (100 - 200ms, Muscle Active Phase): Because of the inertial force from the acceleration of the sled, the subject’s upper torso started to move forward. The lumber acceleration (L3-R: **Fig. 7**) reached maximum in this phase, and this acceleration was transferred to the T12 and T1-head one by one. According to the angular velocity of Head and the T1 (HeadCG-RY, T1-RY: **Fig. 7**), Head and T1 joint started to rotate around 100ms. The T12 and L3 also started to rotate at 100ms (T12-RY, L3-RY: **Fig. 7**). The angular velocity indicates an oscillation mode because of the shoulder-belt and lap-belt (Lap-Belt, Shoulder-RY: **Fig. 7**) started to react from the forward motion of the torso. In relation to the neck link motions, discharge of M. paravertebralis started around 100 ms (PVM: **Fig. 7**). Moreover, the position of the body trunk moved forward, and every muscle normalized ARV value are dramatically increase except M. sternocleidomastoideus and M. rectus abdominis (PVM, VMRA-R: **Fig. 7**).

Compare to the muscle relaxed case, the upper torso motion is constrained by the muscle activation in the muscle tensed case (**Fig. 6**). Therefore, the maximum value of acceleration, angular velocity and the seatbelt force are decreased (**Fig. 8**). Regarding the muscle response, the most of muscles are discharged before the impact (0ms) except M. sternocleidomastoideus and M. paravertebralis (SCM, PVM: **Fig. 8**).

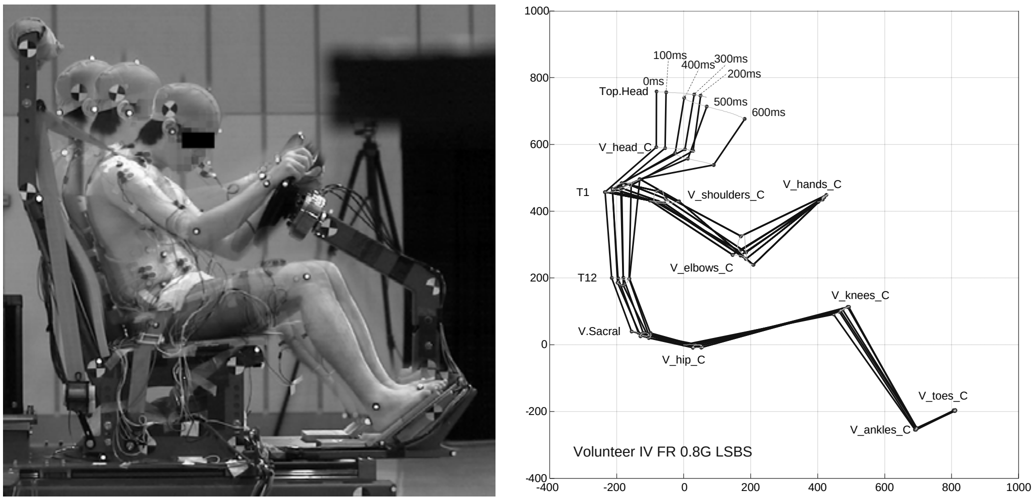


Fig. 5- Physical motions from the 3D motion capturing system (Male, 0.8G: Relaxed)

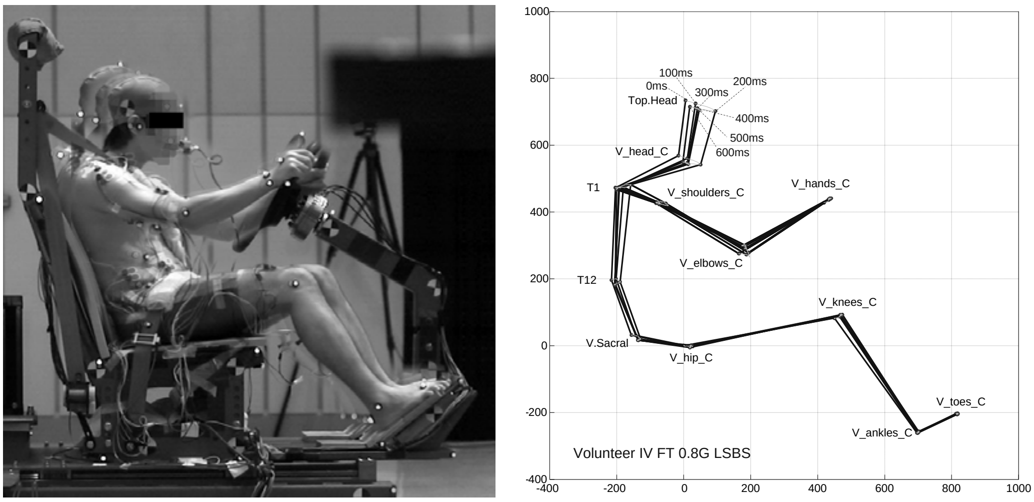


Fig. 6- Physical motions from the 3D motion capturing system (Male, 0.8G: Tensed)

Phase 3 (200 - 400ms, Forward Motion Phase): With the subject's the body trunk restrained to the seat with a belt, the arched rotation of the upper torso started to fold back at around 200 ms (**Fig. 5**). Simultaneously, the head-neck also returned in this phase. T1 movement was synchronized with the acceleration (T1-R: **Fig. 7**) and electromyogram (PVM: **Fig. 7**). Then, the maximum value was indicated around 200ms. In addition, the flexional angular velocity of the head and neck (HeadCG-RY, T1-RY: **Fig. 7**) reached maximum at around 200ms, when the spine deformation fold back. The magnitude of these acceleration decreased due to the activation of the muscles after 200 ms. On the other hand, the hip angular velocity (L3-RY: **Fig. 7**) converged to zero, and the magnitude of T1 and head angular velocity (T1-RY, Head-RY: **Fig. 7**) reached a positive value (Extension) during this phase. The muscles discharge of every location decreased at around 300 ms. The magnitude of foot plate and steering load (Footplate(R), Steering-FX, FZ: **Fig. 7**)) 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.

In the muscle tensed case, the body trunk restrained to the seat due to the muscle activation (**Fig. 6**). Therefore, the acceleration started to decrease and angular velocity of each body region converged to zero in this phase (**Fig. 8**). On the other hands, the muscle activation is discharged continuously and the posture is maintained by the resistance force from the footplate and steering (Footplate(R), Steering-FX, FZ: **Fig. 8**)).

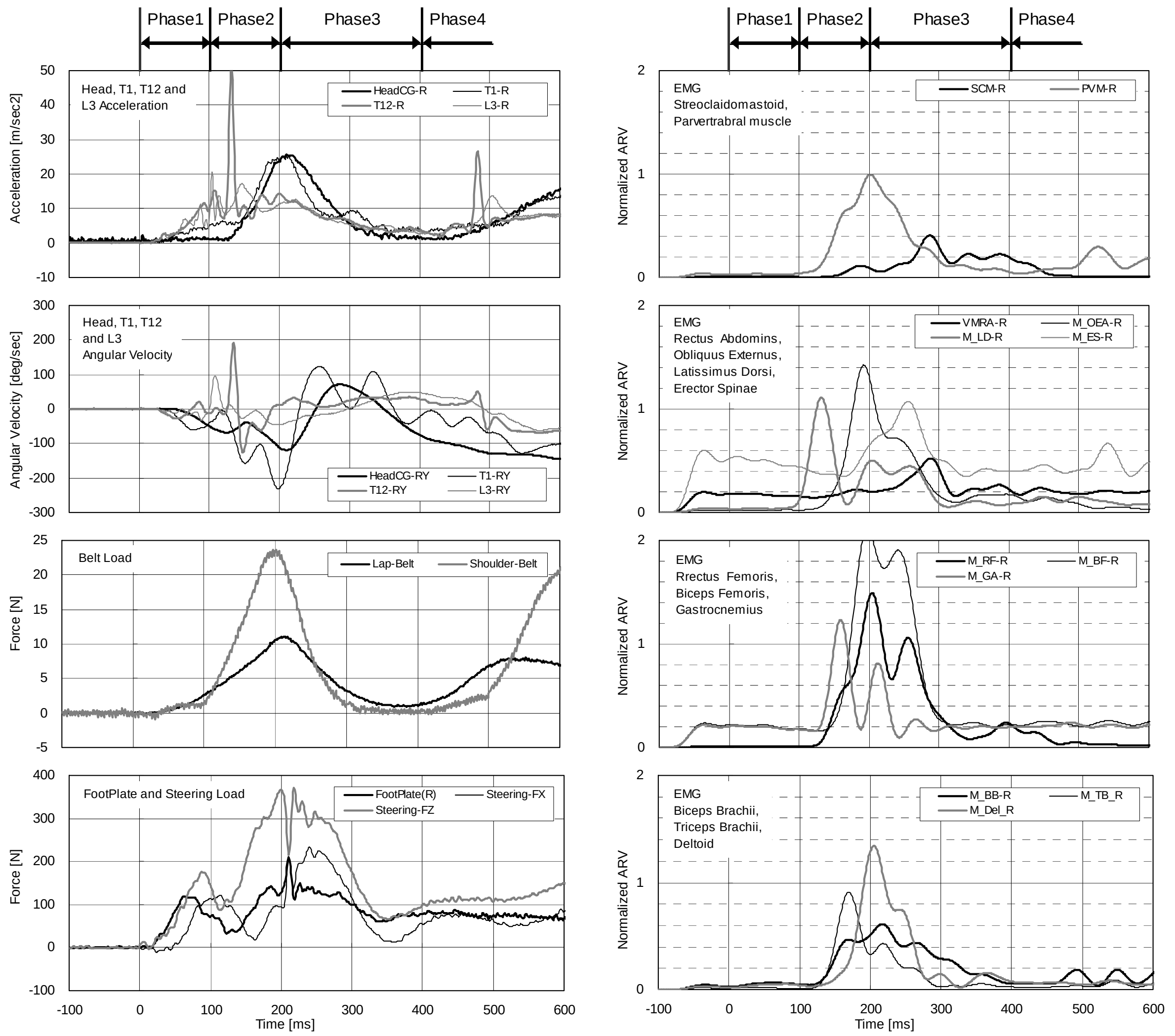


Fig. 7- Time histories of resultant acceleration, angular velocity, restraints load and EMG (Male, 0.8G: Relaxed)

Phase 4 (400ms - End, Final Phase): Because of the muscle relaxed condition, the upper torso started to forward again and the accelerations of head, neck, and torso increased (HeadCG-R, T1-R: **Fig. 7**). In the mean time, the angular velocity of each region shows negative value (Flexion, HeadCG-RY, T1-RY: **Fig. 7**), however, the muscular discharge of the neck, torso, and leg disappeared.

The reaction force of footplate and steering is converged to the 100 N by discharging the lower and upper extremity muscle, the subject found an appropriate balance between the muscle activation and the inertia effect (**Fig. 6**). Therefore, the acceleration and angular velocity converged to zero (**Fig. 8**). The muscle activation level falls by around 30-40% in the period of Phase 3 through Phase 4.

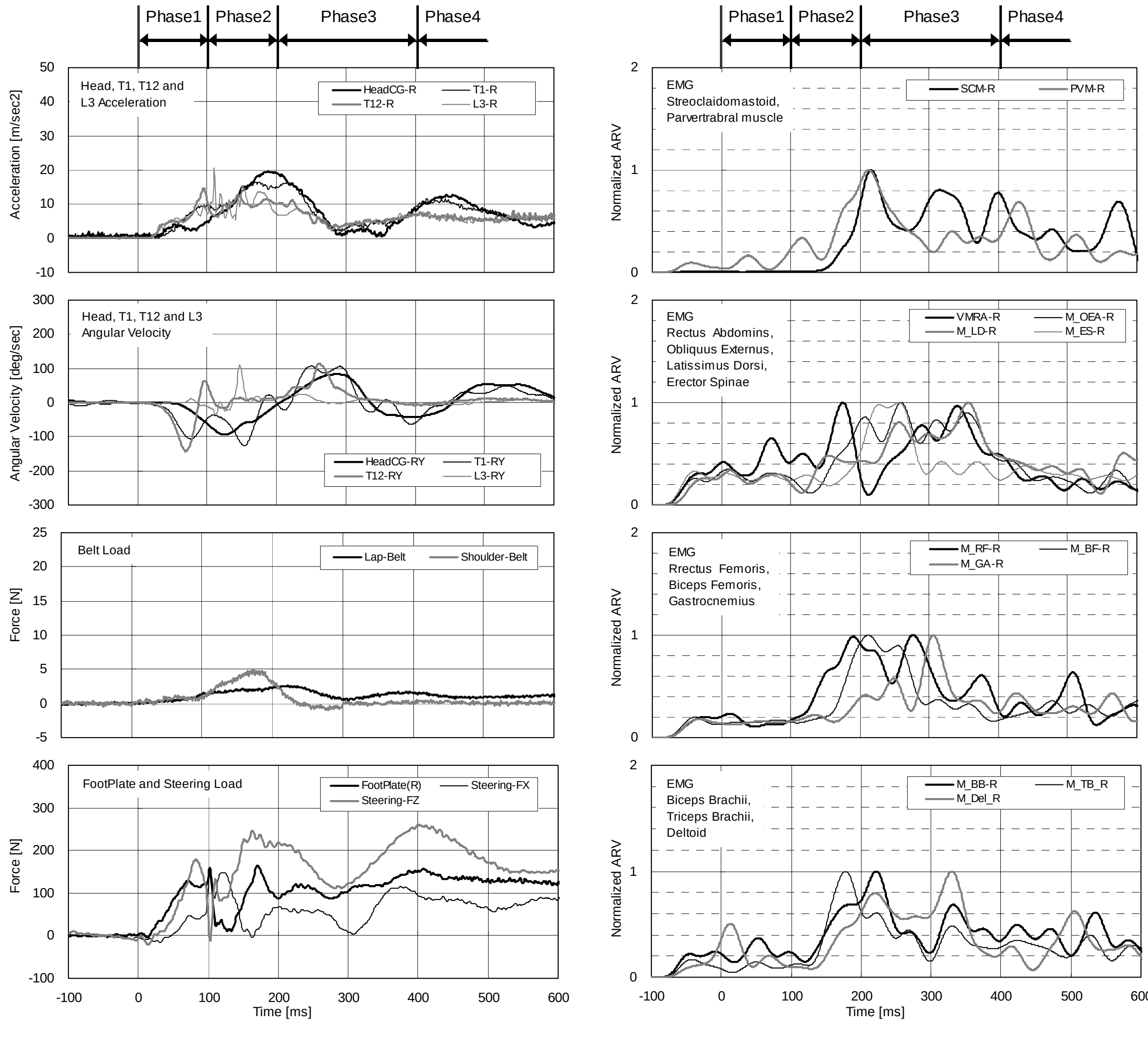


Fig. 8- Time histories of resultant acceleration, angular velocity, restraints load and EMG (Male, 0.8G: Tensed)

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 from the 3D motion capturing system. **Fig. 9** shows the average value of the maximum flexional and extensional angle at the joint. The average value was calculated from the five male volunteers and two female volunteer. 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 indicates flexion. Because of the muscle activities, the major angle difference of lower leg and feet could not be seen in the experiment. In the muscle relaxed case of male and female (**Fig. 9-(a), (c)**), the flexional motion of the neck (Neck) and upper extremity (Lower Arms) are dominated in the body trunk. The extensional motion of the head (Head) and lower extremity (Upper Leg) are shown in the muscle relaxed condition. On the other hand, in the case of tensed condition (**Fig. 9-(b), (d)**), the major flexional motion was detected in the neck (Neck), thorax (T1-T12) and upper extremity (Upper Arms) area, and the head (Head), pelvis (Pelvis), lower extremity (Upper Leg) and upper extremity (Lower Arms) showed extensional motion.

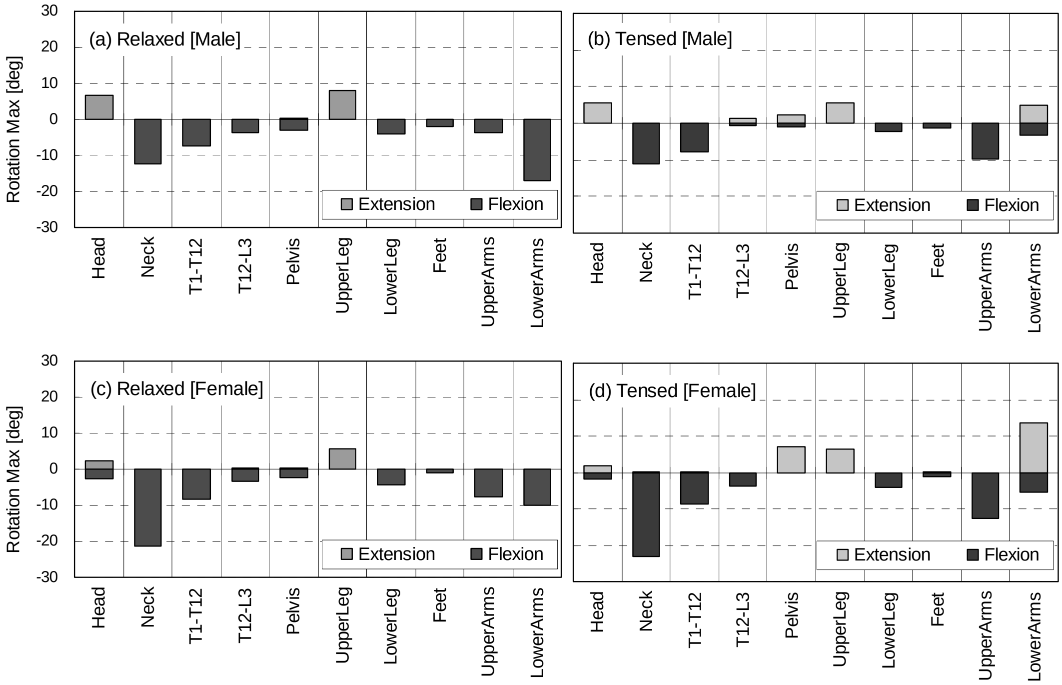


Fig. 9- Maximum flexion and extension angle of each joint

MUSCLE ACTIVITIES DURING PRE-BRAKING: The time history of the major muscle activation in relaxed case is shown in **Fig. 7**. This figure indicates not only the normalized ARV value for defining the muscle activation level of each volunteer during the acceleration but also the reflex time of muscles which affects the kinematics of occupant. According to the kinematic, the subject’s upper torso and head-neck starts to move forward and the torso motion is restrained by the shoulder-belt and steering in Phase 2. As a result of this phenomenon, the neck that is the link between the head and the torso starts to bend. In relation to these neck link motions, the discharge of M. paravertebralis (PVM: **Fig. 7**) starts around 130 ms. Simultaneously the position of the body center of gravity moves forward and the lower extremity such as the M. rectus femoris (M-RF: **Fig. 7**) activate to generate the resistance force to the footplate. In addition, M. obliquus externus abdominis and M. erector spinae (M_OEA, M_ES: **Fig. 7**) are mainly responds from 150ms - 200ms. Based on the volunteer test, the muscle activities during pre-braking was predicted, and the back, neck and lower limbs muscles were mainly working against the forward motion. **Fig. 10** organized the muscle reflex time of both five male and two female data. The listed muscles are mainly working against the forward motion. The activation level of 30% (normalized ARV) is assumed to define the muscle reflex time. Compare to the male subject, the female tends to delay at the lower extremity muscle. 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. These muscle activations should be taken into account when predicting this pre-braking phenomenon with the computer model.

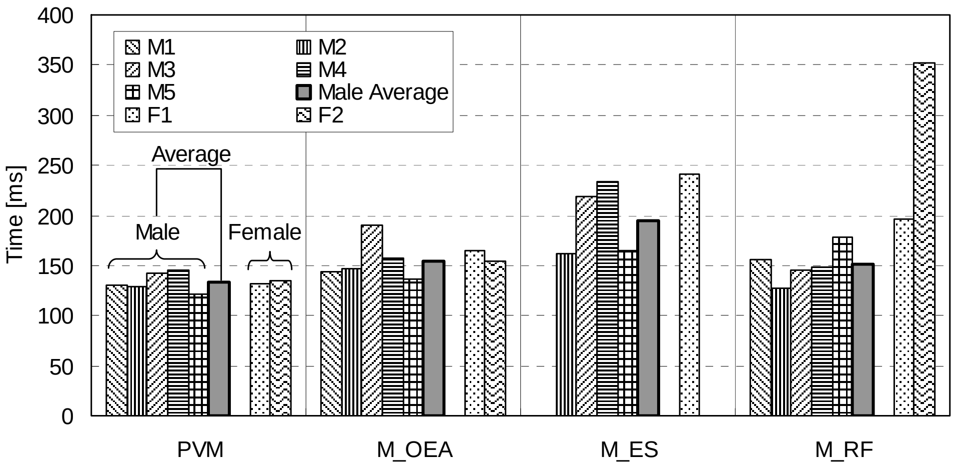


Fig. 10- Muscle reflex time of the major muscles (Right side; M. Paravertebralis (PVM), M. Obliquus Externus Abdominis (M_OEA), M. Erector Spinae (M_ES), M. Rectus Femoris (M_RF))

PHYSICAL MOTIONS WITH RESPECT TO RELAXED AND TENSED MUSCLE BETWEEN MALE AND FEMALE: In order to compare the movement of posture changes, the trajectory of head-CG, shoulder, T1, and elbow was measured with the motion capturing system. **Fig. 11** shows the trajectory of the subjects (five males and two females) under different muscle conditions in 8.0m/s^2. The trajectory is shown with the narrow solid line. Because of the subject's individuality, the location of each landmark at time zero (initial posture) was different from each other. Therefore, each landmark at time zero is set to the individual posture. The arched rotation of the upper torso was observed in each case. The tendency of the curvatures was similar in male subjects even though the maximum forward posture is different (**Fig. 11(a)**). Compared to the relaxed muscle case, the motion of upper torso was controlled by the muscles in the tensed case (**Fig. 11(b)**). On the other hand, the female subjects showed the different trajectory from the males, (**Fig. 11(c), (d)**). The female head trajectory shows a downward tendency in both tensed and relaxed cases.

DISCUSSION

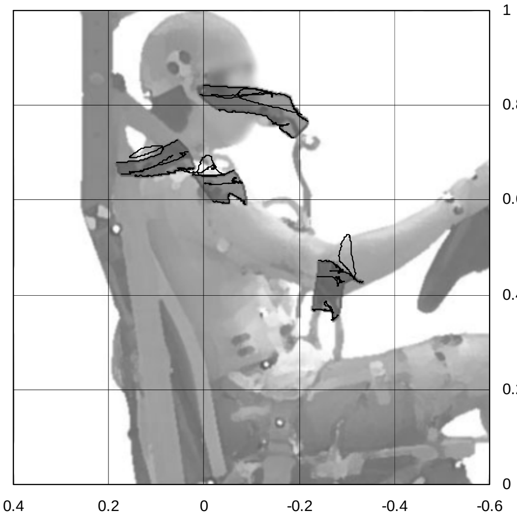
MECHANISMS OF POSTURE CHANGES DURING PRE-CRASH: There exists an adequate correlation between the discharge of muscle force and the acceleration of each body part, as estimated from the results of measurement system. For example, the head-neck-torso acceleration (HeadCG, T1, T12, L3: **Fig. 7**) increases in Phase 2, however decreases in Phase 3. The muscle tensed effect (**Fig. 8**) is clearly seen in the magnitude of acceleration and angular velocity compare to the relaxed muscle condition. This not only due to the back and abdominal muscles, but also to the upper and lower extremity muscles being discharged from the impact (0 ms). In other words, the upper torso was subjected to posture-control provided by these pre-tensed muscle condition which activation level is around 20-40 % of maximum muscle. Following the timing of the muscle activation with upper and lower extremity, the steering and the footplate shows the reaction force continuously. Thus, the subject found the appropriate balance to control the upper body motion by using the reaction force from steering and 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' is activated from Phase 2 and this muscle activation temporarily holds back the upper torso to control excessive motion in each body part. This protective mechanism works more effectively when the steering is installed in the system.

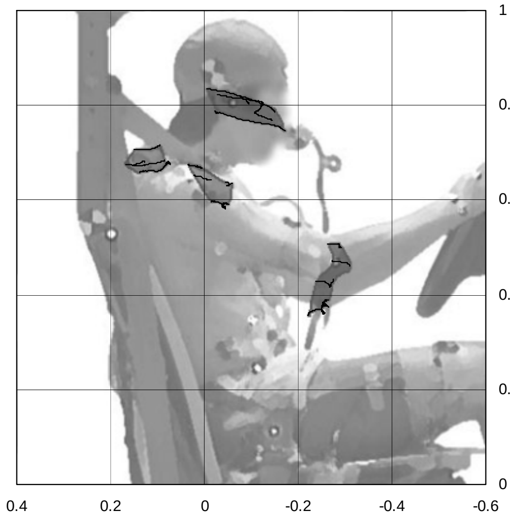
EFFECT OF MUSCULAR TENSION WITH A 3 POINT BELT SYSTEM: In this study, a 3-point seatbelt was installed to constrain the hip and chest in order to model real pre-crash conditions. It was identified that the pre-acceleration tension of muscles affected the physical motions when compared to the relaxed case, however this affect was greatly reduced from that seen in the cases of lap belt only (Ejima et al., 2007). This is because of the 3 point belt system with the steering. When comparing rotational angles between tensed and relaxed cases per body region from pelvis to head, ante-flexional motion due to muscle tension was slightly visible at the neck region in the male subject

(Neck: **Fig. 9**). In this region, the posture-control effect of the rotational angle due to muscle tension was around 10%. On the other hand, the hip region (Pelvis: **Fig. 9**) shows slight flexional rotational motion in tensed and relaxed case. According to a previous study with cases involving a lap-belt without steering wheel, the hip region showed the largest flexional motion. Consequently, the rotational angle of hip region is seen to be strongly affected by the upper torso motion which is restrained by the 3 point belt system and the steering in the front impact case. Therefore, the restrains effect is important when discussing the stability of the posture under low speed acceleration.

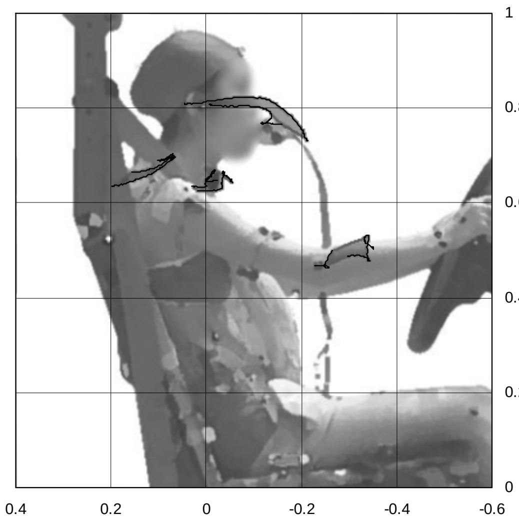
DRIVING POSTURES IN PRE-CRASH SITUATIONS: The solid lines in **Fig. 11** indicate the trajectory (0ms – 600ms) of each volunteer. In this figure, each of the muscle conditions (relaxed and tensed) is shown separately. Also, compared to the tensed muscle case, each landmark in the relaxed cases is displaced a large distance and the horizontal movement is seen at the initial stage of head motion in the relaxed case (**Fig. 11(a), (c)**). This figure indicates the difference between the male and female volunteers respectively at the head, T1 and shoulder markers. Compared to females, the males tend to constrain their posture more significantly with muscle activation. The reason for this difference between males and females is considered to be the relative volume of the muscles.



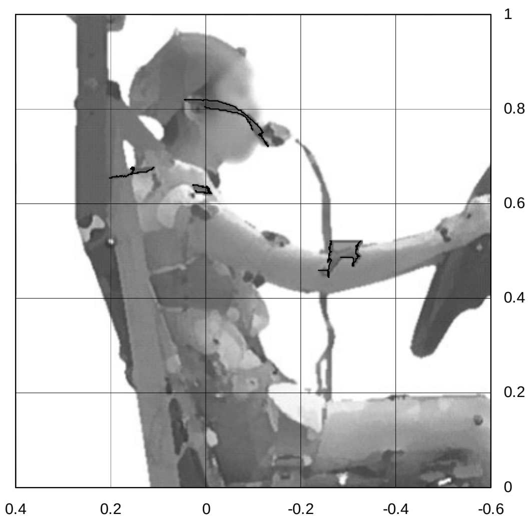
(a) Male (Relax)



(b) Male (Tensed)



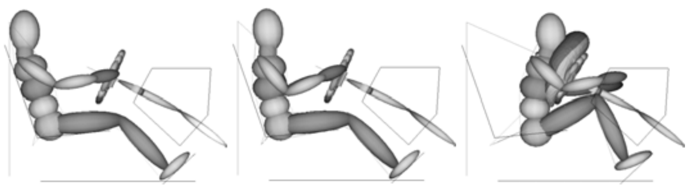
(c) Female (Relax)



(d) Female (Tensed)

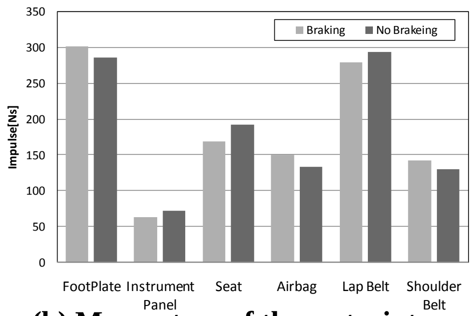
Fig. 11- Trajectory of the subjects (five males and two females)

PREDICTION OF IMPACT POSTURE CHANGES DUE TO PRE-CRASH CONDITIONS: According to the experimental result, the physical motion of the human body under pre-crash condition is predicted. Therefore, the effect of the posture change caused by pre-crash condition was calculated by using a computer human model. This computer model is based on a rigid body model, with muscle effects incorporated by using the hill-type muscle model. The EMG data measured from the braking acceleration experiments were directly applied to the major muscles in order to simulate this acceleration's effect on the human body. **Fig. 12(a)** shows the configuration of posture changes which simulate the actual car motion when the driver brakes in an emergency situation. In this figure, after the recognition of danger, the driver made an emergency braking. Therefore, the driver moved forward before the vehicle crashed into a rigid wall with delta-V 50km/h. Occupant component momentum was calculated as the integration of the various occupant contact forces with respect to time as shown in **Fig. 12(b)**. Compared to the "non-braking" case, the momentum measured at the foot plate, airbag, and shoulder belt is larger than that seen in the braking case. The reason for this difference is that the braking causes not only a posture change, but also a change in body velocity due to inertial forces. Therefore the contact force of between the chest and the restraints increases. From this calculation, the effect of pre-crash braking is predicted using the computer human model. For a more detailed understating the mechanisms, further study is needed to distinguish the factors which are present in real accident cases.



Hazard Perception (-630ms) **Crash (0ms)** **Maximum Force (69ms)**

(a) Configuration of posture changes



(b) Momentum of the restraints

Fig. 12- Front impact simulation with the pre-braking condition by using the human model

LIMITATION OF THIS STUDY AND SUGGESTIONS FOR FURTHER RESEARCH: The number of the subjects in this study was limited especially for the female volunteers. Therefore, the female data were insufficient to discuss the difference between male and female. In addition, modeling of occupant motions of the body necessities 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 about muscle effect. This could be done in a co-operation between the testing and simulations.

CONCLUSION

The result of this study concluded that the effects of muscular tension on each body motions have been clarified and the physical motion is divided into four phases in the pre-crash condition. Furthermore, it was identified based on acceleration, EMG electrodes and the reaction forces that differences in muscle activity govern the motion of the body during each phase. Finally, it was found that the muscles that were most highly activated when the occupant made a pre-braking action were the abdominal muscles and back muscles such as M. erector spinae and M. paravertebralis. These parameters are important factors when discussing the subject's motion with the restraints system just before a collision. In addition, the restrains system maintain the driving posture and stabilize the pelvis motion compare to the cases involving a lap-belt only.

The present human body model adequately represents the general kinematics of the physical motion seen in pre-impact braking conditions. This, in turn, indicates that the EMG data of major muscles significantly influences the physical motion, since these input variables are directly taken from the volunteer tests. This model is currently in the improvement phase, with its practical application, and injury level prediction to be completed using a finite element model in the next stage.

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