Effects of Pre-impact Swerving/Steering on Physical Motion of the Volunteer in the Low-Speed Side-impact Sled Test

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Abstract In order to simulate the actual pre-crash condition of a car that occurs when the drivers avoid an accident in an emergency situation, low-speed lateral sled tests on human volunteers were conducted using a sled-mounted rigid seat. It was observed that when the subject's muscles were initially relaxed, Sternoclaidomastoid and Rectus Abdominis started activation at around 100ms after the onset of acceleration and reached its maximum value at around 200-400ms. During this time period, most of the individual body region acceleration responses and restraint system reaction forces also peaked. Furthermore, the lateral flexion of head-neck-torso kinematics was strongly influenced by the muscle activity in the muscle tensed case and the posture-control effect of the lateral flexion due to the muscle tension was around 20-40%. This experiment indicates that muscles can control the occupant's behavior significantly during the low-speed impact, relating to the occupant's posture at the pre-crash phase.

Keywords Kinematics, Muscle, Occupants, Pre-Crash, Side Impacts, Volunteers

I. INTRODUCTION

In the studies on car crash safety, post-crash driver behavior and injury mechanism are often discussed, and in such discussions, it is assumed that at the pre-crash phase (just before the collision), the driver maintains standard posture of the Anthropomorphic Test Dummy (ATD). However, not only does the driver posture vary according to the age, gender and physique, but in the real accident, the posture also changes during the driving experienced due to harsh braking and/or swerving/steering. Thus, it can be supposed that it is difficult for the driver to maintain the standard posture just before the collision. Fig.1 shows the accident type and evasive maneuvers obtained from the Institute for Traffic Research and Data Analysis (ITARDA) in Japan (1993-2004). According to the accident analysis[1], 60% of drivers took accident avoidance measures such as braking, swerving/steering, or both of them[2]. This means that most drivers take accident avoidance measures of some kind at the pre-crash phase. Moreover, the accident data analyses suggest that the change of the driver behavior at the pre-crash phase exerts influence on the injury site as shown in Fig. 2. The data source consists of 271 cases of frontal impact collision (CDC: 11F-1F) with belted driver. The distribution of injury incidence rates of body region changes depend on the type of evasive maneuvers. Therefore, the examination of driver behavior at the pre-crash phase is an important theme in searching for more effective safety measures and damage reduction.

Many research studies have been done on the influence of drivers' physique and of their posture change at the pre-crash phase. For example, Armstrong et al. [3] examined the influence of the muscle response to the crash in the volunteer tests with a sled in the high-speed range, and they point out the shock-absorbing effect of the lower limbs. Cross et al. [4] investigated drivers' postures at the pre-crash phase in volunteer tests with a real car. They discuss the distinction between the live human body and the ATD by comparing the amount of their posture changes that are caused by their driving postures and accident prevention behaviors such as harsh braking or lane change. Behr et al.[5] also conducted volunteer experiments in order to define the pre-braking condition, both on a driving simulator and on a real car. Bingley et al. [6] and Parkin et al. [7] recorded the drivers' drive-by postures as image data and categorized them based on the physique and gender of the drivers. They report that male drivers obviously take different postures from female drivers. In addition, research studies on the influence of posture change on human injury have been done by using a human computer model [8]-[11] in order to reflect muscle response features to the computer model and to optimize the occupant

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protection system by using occupant's physique and initial posture as parameters [12][13]. Recently, on the other hand, motorized seatbelts have been equipped in some commercially available vehicles as an occupant protection device to control the influence of the pre-crash posture change. This device, besides the pretentioner and the force limiter, has the function to enhance restraint on the occupant posture at the pre-crash phase by automatically furling the belt with the electric motor in response to the damage-reducing braking [14]. Good et al. [15]-[16] investigated the basic features of the restraint effect of the motorized seatbelt based on the data from the volunteer tests and ATD tests, and they defined the appropriate posture changes from the numerical model that takes the effect of the motorized seatbelt into consideration. These previous studies suggest that driver's posture change in the pre-crash phase exerts influence on occupant injury risk. Therefore, it is important to make quantitaive analyses on the relationship between accident avoidance maneuvers and the amount of the posture change.

In this paper, the effect of pre-impact swerving/steering on physical motion is evaluated by tests with volunteers applying the pre-crash conditions. The authors have studied the posture changes caused by pre-impact braking through the use of volunteer tests. The previous study in this series [17]-[19] was conducted only in the frontal direction. In this study, the posture of the driver at the moment of swerving/steering just before the impact was examined. 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. In addition, the effects of muscular tension on each body motions were important factors in discussing the subject's motion just before collision. 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 should lead to further improvement of the effectiveness of occupant crash protection measures in accident situations.

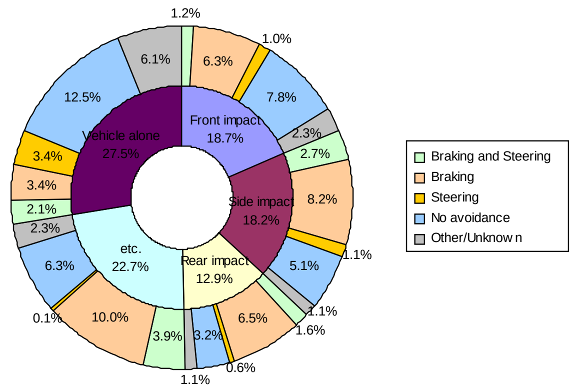


Fig. 1- Accident type and evasive maneuver in Japan.

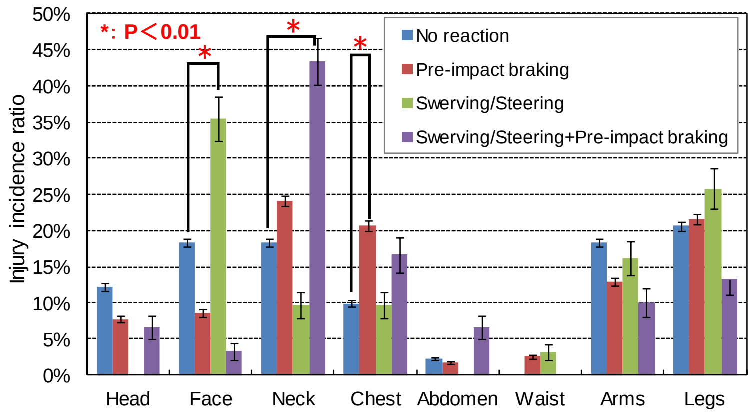


Fig. 2- Distribution of injury incidence ratio of each body region in the frontal collision (271 cases belted driver)

II. METHODS

Volunteer and Informed Consent

Three healthy 22 to 28 year-old volunteers (three males) 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 the Low-Speed Impact

Fig. 3 illustrates the side-impact simulation sled system (hereafter referred to as "Mini Sled"). The side pre-impact sled was designed in order to simulate the acceleration experienced when the driver was swerving/steering in an emergency situation. The sled has a rigid seat (hereafter referred to as "R-seat"). Low-level side pre-impact was applied to the volunteer by accelerating the sled. The acceleration of 6.0m/s^2 is the limit of side-impact with this Mini Sled. Therefore, two possible acceleration (4.0m/s^2 and 6.0m/s^2) were applied to reconstruct the posture change of passenger at the accident avoidance by swerving/steering as shown in Fig. 4. For the purpose of comparison, the duration of acceleration is set 600ms which was defined in the frontal impact test[18].

The R-seat without wings/armrest made of steel was mounted on the sled. The seat surface has 500mm square area and sitting height is 350mm. In this experiment, the reaction forces coming from the foot plate and the seat were measured by load cells and pressure sensor, respectively. A belt was used to constrain the hip for safety purpose. In this case, the belt was adjusted to the length of the hip region and thus was 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[20].

TABLE 1. Subjects

Volunteer	Age (year)	Sex	Height (cm)	Weight (kg)	Sitting height (cm)
I	26	M	176.2	68.6	94.0
II	25	M	165.5	57.7	86.3
III	28	M	178.0	67.2	92.0

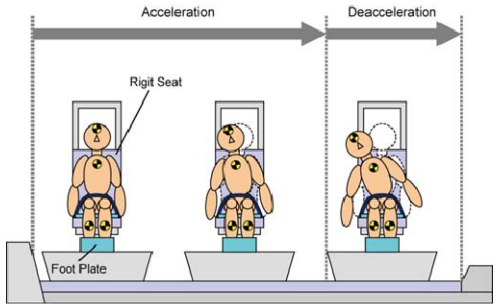
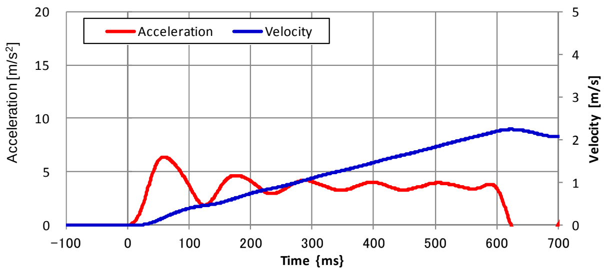
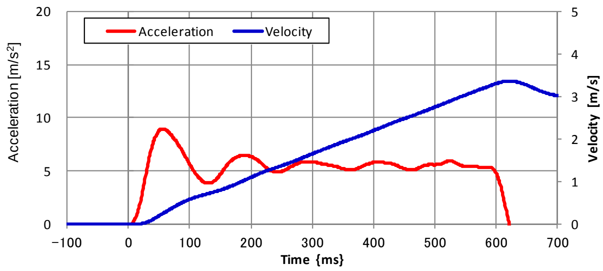


Fig. 3- Outlook of the side-impact sled system



(a) 4.0m/s^2



(b) 6.0m/s^2

Fig. 4- Sled acceleration and Velocity

Experimental Condition

Three healthy males 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 all their muscles intentionally. The muscle activation during the impact was evaluated with EMG signal measured via surface electrodes and the initial activation is compared with the relaxed case at the normal condition to ensure that the volunteers were relaxed or tensed in every test. 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 the upper torso. The alignment of head-neck-torso was adjusted to the upright posture and head-neck region was referenced to the Frankfurt line which was defined in the previous paper[17].

TABLE 2 TEST MATRIX

	Impact acceleration	Direction	Muscle condition	Boundary condition
3 adult (male)	0.4G	Side	Relaxed	Lap Belt
	0.6G		Tensed	

III. ANALYSIS

Measurement

The measurement methodology of the acceleration at each region of the subject and the electromyographic response were referenced in the previous reports [18][19] and the measurement items were listed in Fig. 5 and Fig. 6. In order to monitor the motion of the volunteer at the time of impulse, accelerometers were placed on the body surface. 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. 5 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.

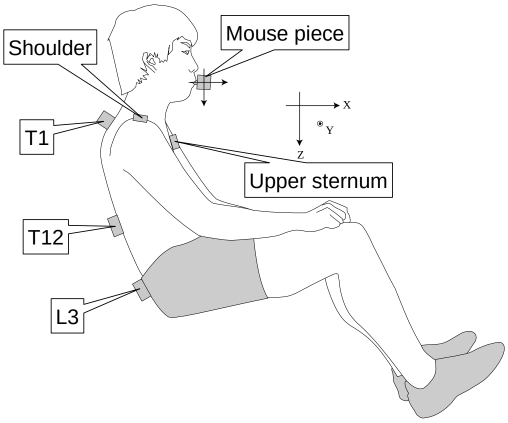
The location of the anatomic center of gravity of the head was determined using the method applied in the frontal impact test[17]. The position of the head CG was located 5mm in front of the external auditory meatus and 20mm above the Frankfurt line which connects the lower orbital margin and the center 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 selected muscles as shown in Fig. 6 were related to each segment motion and relatively closed to the skin surface that can detect the muscle activity via surface electrodes. In addition, the relationship between the physical motion and the muscle activity was confirmed by the orthopedic specialist by changing the joint angle of each segment at the preparation.

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. 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. From these motion data, a skeleton image for the side-impact is generated based on segments determined by body surface landmarks. Each segment of the head, neck, torso, abdomen, hip, thigh and the lower legs are defined as shown in Fig. 7. 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, Hip) and the joint angle at each connection point (Head, Neck, T1-T12, T12-L3, Hip, Upper Leg, Lower Leg, Upper Arms, Lower Arms) was calculated with the motion capturing software. The maximum measuring error estimated from 3D motion capturing system by using body surface landmarks was around 1.6mm in this environment.

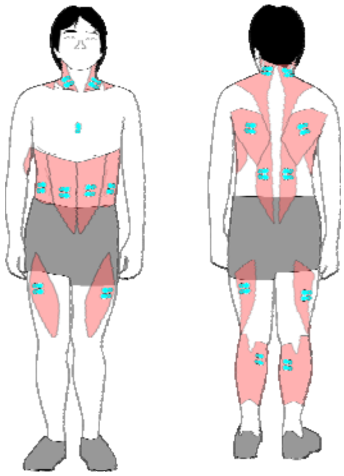
EMG Signal Processing

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). The normalized ARV value is defined in each muscle.



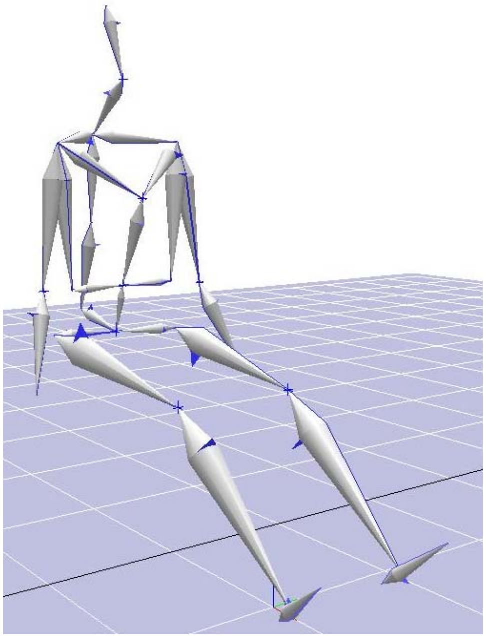
- **Head (via a mouthpiece) :**
  tri-axial accelerometers and angular velocity meter
- **First thoracic vertebra (T1):**
  tri-axial accelerometers and angular velocity meter
- **Twelve thoracic vertebra (T12):**
  tri-axial accelerometers and angular velocity meter
- **Third lumber vertebra (L3):**
  tri-axial accelerometers and angular velocity meter
- **Shoulder:**
  tri-accelerometer
- **Sternum:**
  tri-accelerometer

Fig. 5- Lateral view of the head/neck/torso/pelvis with mounted accelerometer



Neck:
M. Sternocleidomastoideus (SCM)
M. Paravertebralis (PVM)
Torso:
M. Latissimus Dorsi (M.LD)
M. Erector Spinae (M.ES)
Abdomen:
M. Rectus Abdominis (M.RA)
M. Obliquus Externus Abdominis (M.OEA)
Lower Extremity:
M. Biceps Femoris (M.BF)
M. Rectus femoris (M.RF)
M. Gastrocnemius (M.GA)

Fig. 6- Location of the electromyography



- **Head:** vertex - head C.G.
- **Neck:** head C.G.- 1st thoracic vertebra
- **Shoulder(R&L):** 1st thoracic vertebra - shoulder joint
- **Upper Arms(R&L):** shoulder joint - elbow
- **Lower Arms(R&L):** elbow joint - wrist
- **Torso:** 1st thoracic vertebra - 12th thoracic vertebra
  shoulder joint - sternum
  shoulder joint- lower-rib
  sternum - rib center
  lower-rib - rib center
- **Abdomen:** 12th thoracic vertebra - 3rd lumber vertebra
  rib center - trochanter center
- **Hip:** 3rd lumber vertebra - trochanter center
  trochanter- trochanter center
- **Upper leg(R&L):** trochanter- knee join
- **Lower leg(R&L):** knee joint - ankle joint
- **Feet(R&L):** ankle joint- toe

note : R&L means right and left

Fig. 7- Definition of the segment region along the whole body

IV. RESULTS

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 three volunteers in each case as shown in Table 2. The results of the experiments conducted with an impact acceleration of 4.0m/s^2 and 6.0m/s^2 using a rigid seat are described. The motions observed by three-dimensional movement analysis system was illustrated 2D projection of the head, neck, spine and pelvis on a global YZ plane at 100ms interval until the approximate time of peak displacement (500ms). A subject's motion, response to the acceleration and the angular velocity are divided into four phases. It is because the subject momentarily changes their posture during the low-speed impact. Therefore, the categorization of lateral motion helps to understand the mechanisms of posture change in each phase. The length of each phase was defined by the timing of muscle activation and the kinematic motion.

A subject's acceleration and angular velocity response were defined as the average time histories of resultant acceleration(-R) and X-axis angular velocity(-RX) of the head, T1, T12 and L3 from the three volunteers. In addition to the acceleration, the time histories of reaction forces with belt and footplate are shown in the same way. Moreover, the average time histories of EMG response of each muscle of the three subjects are indicated. Finally, the time of acceleration onset is set at zero (0ms) in the time history diagram.

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 movement analysis system with respect to the sled, 2) acceleration and angular velocity 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.

<u>Phase 1 (0 - 100ms, Initial Response Phase)</u>: No significant motion was identified in this phase although the acceleration was applied to the sled during this phase. This acceleration appeared with the acceleration of L3 in both relaxed and tensed case (L3-R: Fig. 10, Fig. 15) which was close to the sled. Then, no major acceleration was shown in this phase except for the L3. The major muscle activation was not detected in the monitored muscles in the relaxed case. Meanwhile, initial muscle activation was activated around 20-40% relatively in the back and lower extremity muscle in the tensed case.

<u>Phase 2 (100 - 200ms, Muscle Active Phase)</u>: Because of the inertial force from the acceleration of the sled, the subject's upper torso started to move sideward. The lumber acceleration and angular velocity (L3-R, L3-RX: Fig. 10, Fig. 15) reached maximum value in this phase, and this acceleration was transferred to the T12 and T1-head one by one. According to the angular velocity of T1 and T12 (T1-RX, T12-RX: Fig. 10, Fig. 15), T1 and T12 jointly started to rotate around 100ms. The angular velocity L3-RX reached the maximum value because the belt (Belt Force: Fig. 11, Fig. 16) started to react from the sideward motion of the hip. In relation to the neck link motions, discharge of M. Sternocleidomastoideus started around 100 ms (SCM: Fig. 12, Fig. 17). Moreover, the position of the body trunk moved sideward, and every muscle normalized ARV value was increased. There were no major difference in posture between the muscle relaxed and tensed case at the 200ms from the 2D projection of the skeleton image with target marker. Therefore, the time history of acceleration, angular velocity and the belt force were not much different between the muscle relaxed and tensed case in this phase. Regarding the muscle response, most of the muscles are discharged in this phase.

<u>Phase 3 (200 - 400ms, Sideward Motion Phase)</u>: With the subject's body trunk restrained in the seat with a belt, the arched rotation of the upper torso started. T1 and T12 movements were synchronized with the acceleration (T1-R, T12-R: Fig. 10, Fig. 15) and electromyogram (M.ES, M.LD: Fig. 12, Fig. 17), and the maximum value was indicated around 250ms. In addition, the angular velocity of the head and neck (HeadCG-RX, T1-RX: Fig. 10, Fig. 15) reached maximum at around 300ms. The magnitude of this acceleration decreased due to the activation of the muscles after 300 ms. On the other hand, the hip angular velocity (L3-RX: Fig. 10, Fig. 15) converged to almost zero. The magnitude of belt load (Belt Force: Fig. 11, Fig. 16) indicated the maximum value at around 400 ms. In the muscle tensed case, the body trunk was restrained due to the muscle activation. Therefore, the angular velocity started to decrease and acceleration of each body region converged to a certain value in this phase. On the other hand, the muscle activation was discharged continuously and the posture was maintained by the resistance force from the footplate and the belt.

<u>Phase 4 (400ms - End, Final Phase)</u>: Because of the muscle relaxed condition, the upper torso maintained the sideward motion and the accelerations of head and neck closed to the impact acceleration (HeadCG-R, T1-R, T12-R, L3-R: Fig. 10, Fig. 15). In the mean time, the angular velocity of each region showed convergence to almost zero (Flexion, HeadCG-RX, T1-RX T-12-RX). The muscular discharge of the neck, torso and leg kept a certain amount of level even though the subjects were intended to relax their muscle. The reaction force of the belt converged to 300 N for the right side and 500N for the left side in the muscle relaxed case in 6.0m/s^2. During this time period, the subject found an appropriate balance between the muscle activation and the inertia

effect. Therefore, the angular velocity converged to zero and the muscle activation level fell in the period of Phase 3 through Phase 4.

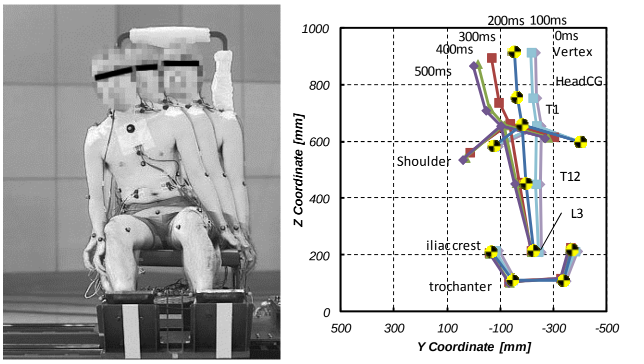


Fig. 8- Physical motions with 2D projection of the head, neck, spine and pelvis (Male, 0.4G: Relaxed)

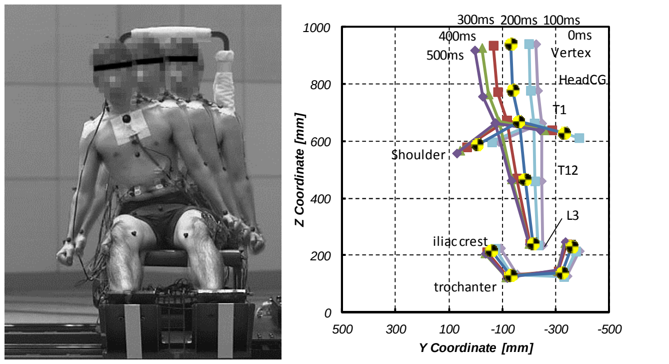


Fig. 9- Physical motions with 2D projection of the head, neck, spine and pelvis (Male, 0.4G: Tensed)

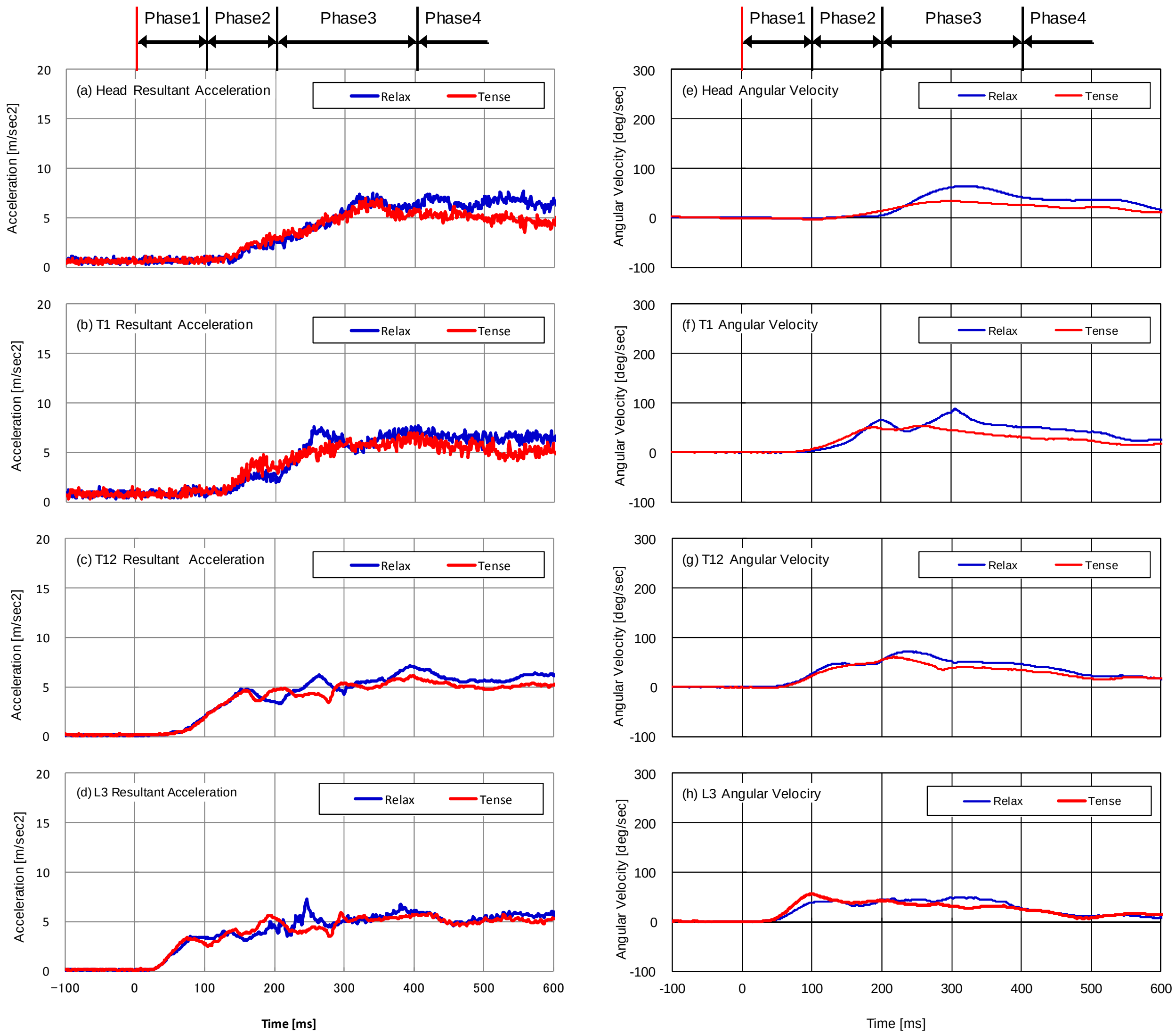


Fig. 10- Time histories of resultant acceleration, angular velocity (0.4G: Ave. Relaxed & Ave. Tensed)

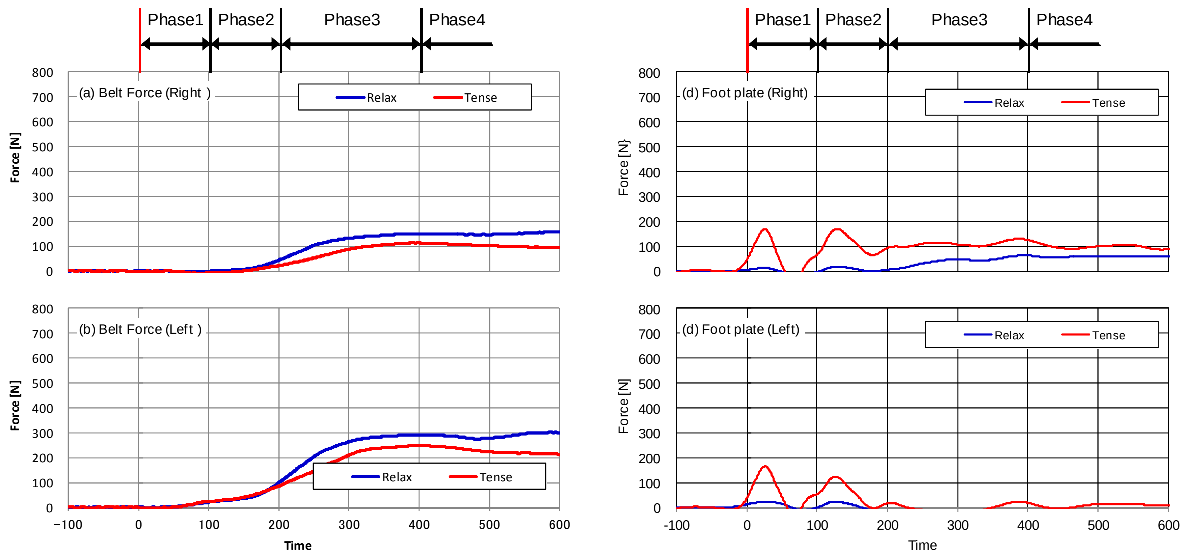


Fig. 11- Time histories of belt and footplate force (0.4G: Ave. Relaxed & Ave. Tensed)

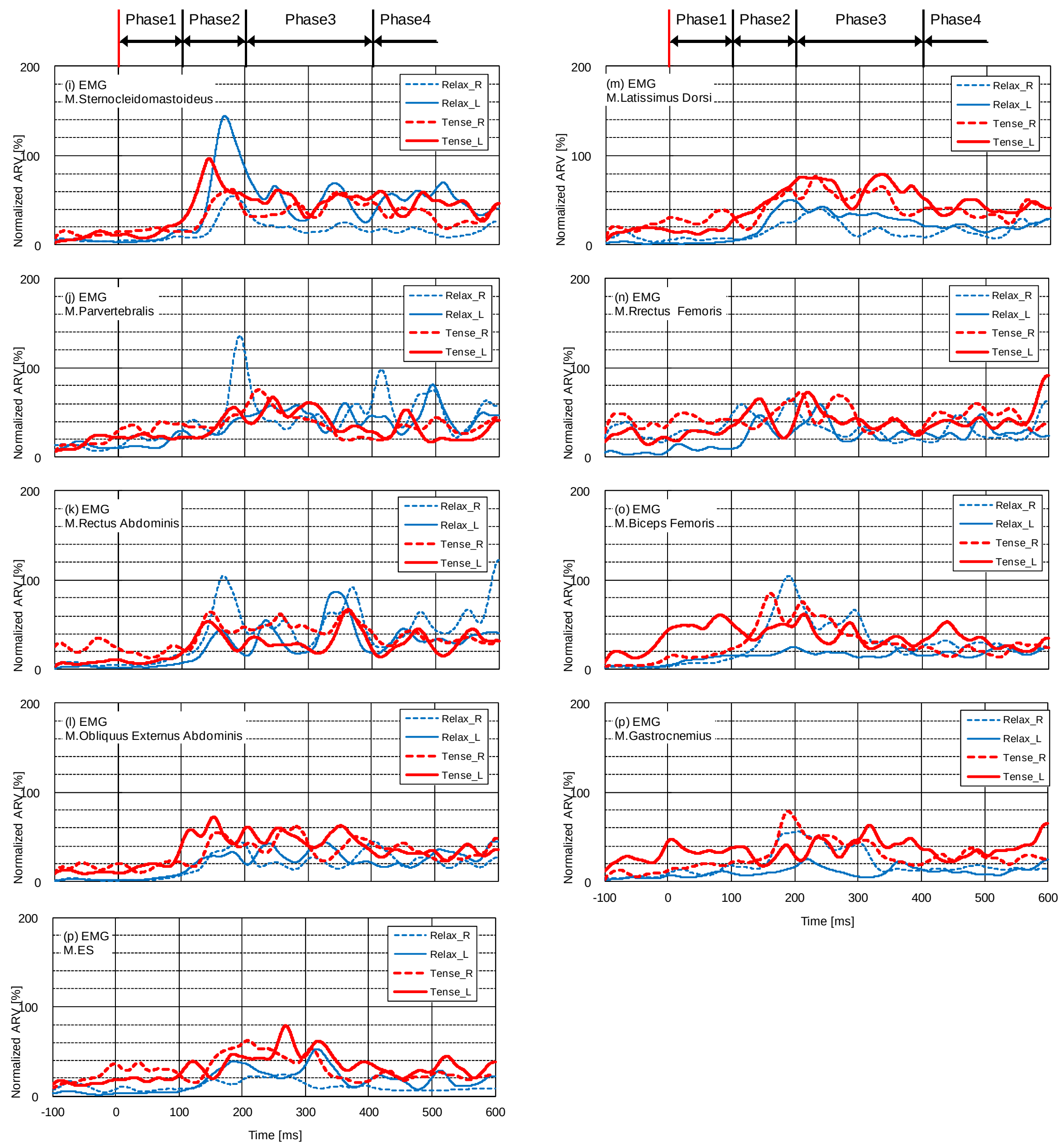


Fig.12- Time histories of ARV (0.4G: Relaxed & Tensed)

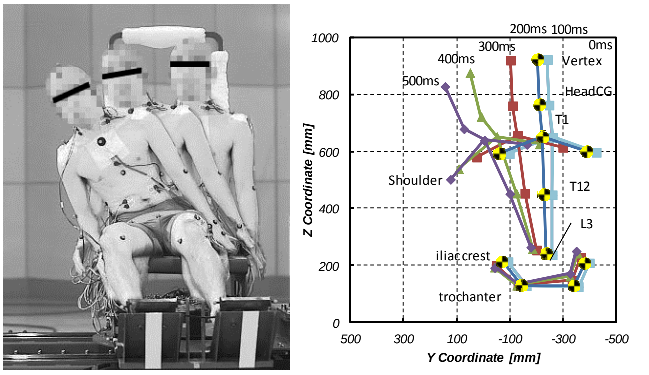


Fig. 13- Physical motions with 2D projection of the head, neck, spine and pelvis (Male, 0.6G: Relaxed)

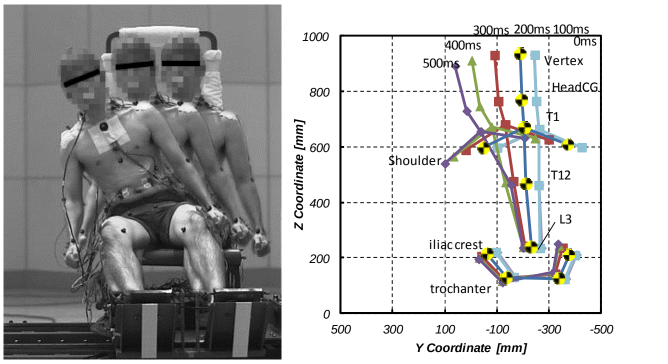


Fig. 14- Physical motions with 2D projection of the head, neck, spine and pelvis (Male, 0.6G: Tensed)

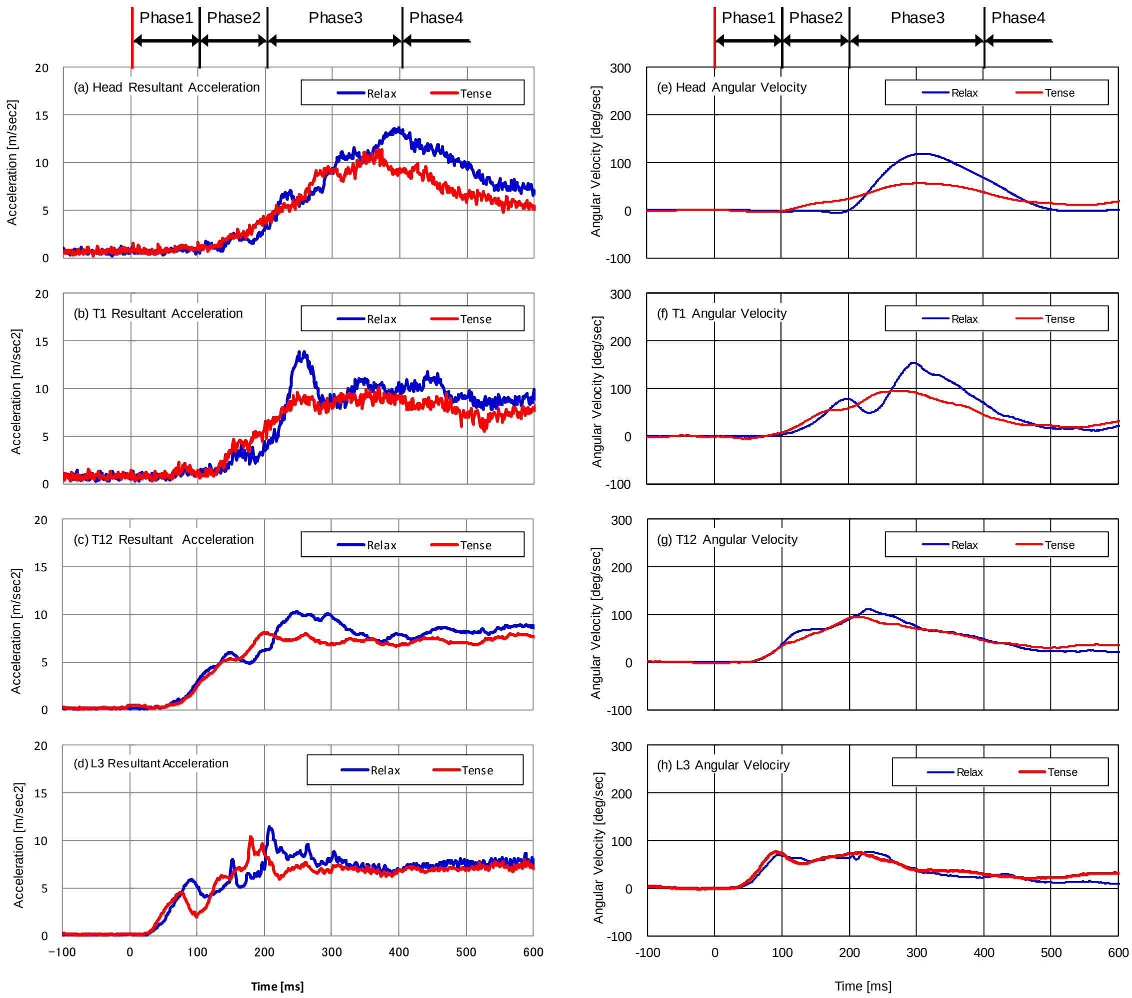


Fig. 15- Time histories of resultant acceleration, angular velocity (0.6G: Ave. Relaxed & Ave. Tensed)

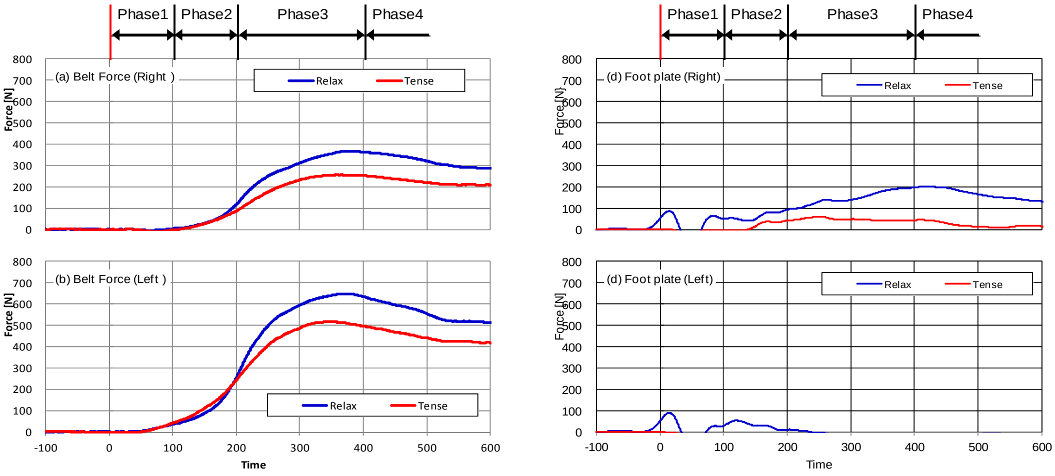


Fig. 16- Time histories of belt and footplate force (0.6G: Ave. Relaxed & Ave. Tensed)

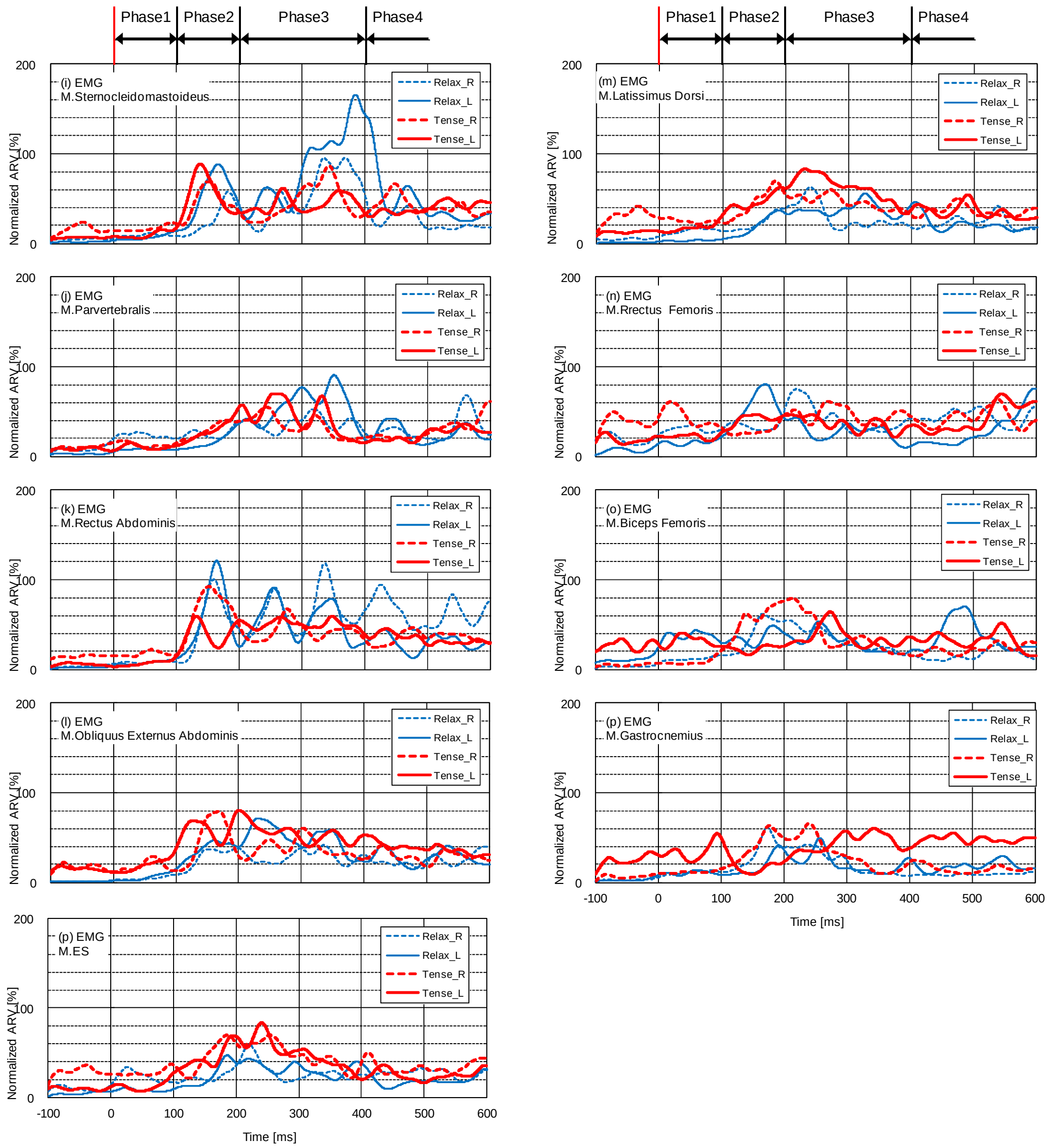


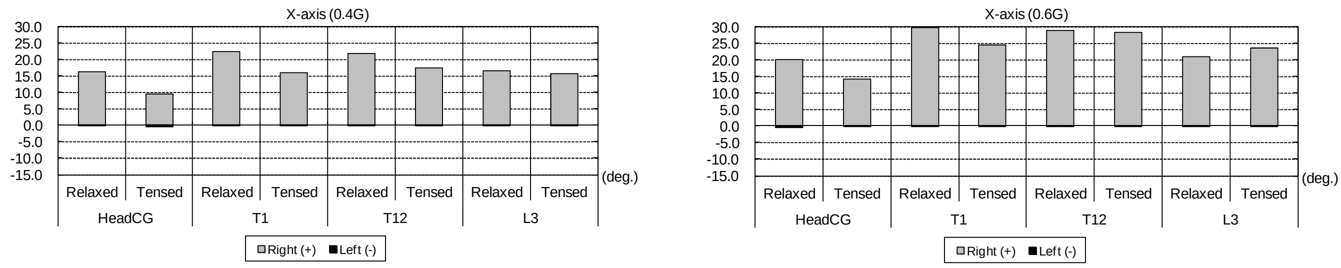
Fig. 17- Time histories of ARV (0.6G: Relaxed & Tensed)

Rotational Angle of Head, Neck and Torso

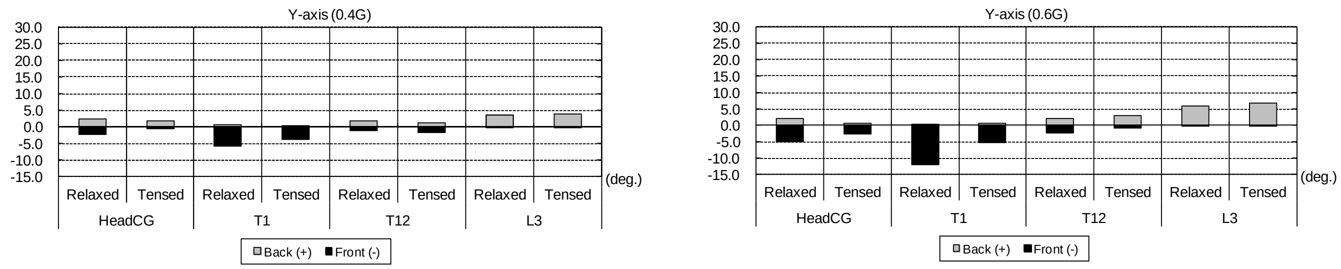
It has been detected that the pre-impact tension of muscle affects the physical motion at low level impact, and this muscle effect is mostly related to the rotational angle of the head, neck and torso. Therefore, the rotational motion of the upper torso was analyzed based on the angular velocity measured from the tri-axial angular velocity meter. Fig. 18 shows the average value of the maximum angle at HeadCG, T1, T12 and L3. The average value was calculated from three male volunteers. For the purpose of comparison, the tensed and relaxed muscle cases and the responses with two different impact levels are shown in Fig. 18. As for the rotational angle of each location, the primary value was set as zero (0). The upper row of Fig. 18 shows the lateral motion (X-axis rotation), the middle row shows flexion and extension (Y-axis rotation) and the lower row shows turning motion (Z-axis rotation). The plus (+) and the minus (-) direction of rotation is defined in Fig. 5. It was confirmed that the rotational angle of each location in the case of 0.6G is larger than that of 0.4G. X-axis rotation represents the lateral movement away from the midline of the body and this rotational motion is dominated in the side-impact sled test. The rotational motion of the spine (T1, T12 and L3) is mainly rotated in the lateral flexion motion as shown in Fig. 18(a). On the other hand, Y-axis rotation shows the flexion motion at the neck region (T1), even though the magnitude of angle is relatively smaller than the X-axis rotation. This is because of the turning of the spine to the side due to the belt restraint at the hip area during the lateral flexion motion. The restraint effect also can be seen in the Z-axis rotation and the rotational angle of L3 which is close to the belt restraint and shows a larger effect than other locations.

Because of the muscle activities, the major angle difference of Head (HeadCG), Neck (T1), thoracic spine (T12) and lumbar spine (L3) could be seen in the experiment. In the muscle relaxed case, the lateral flexion motion of the neck (T1) and thoracic spine (T12) are dominated in the body trunk (Fig. 18-(a)). In addition, the flexion motion (Y-axis rotation: Fig. 18-(b)) of the Neck (T1) is shown in the muscle relaxed condition. On the other hand, in the case of tensed condition, the major lateral flexion motion was decreased in the Head (HeadCG) and Neck (T1) compared to the muscle relaxed case. Moreover, it was confirmed that the effect of muscle activity is shown in the flexion motion (Y-axis rotation) of T1 and the turning motion (Z-axis rotation) of T12 and L3.

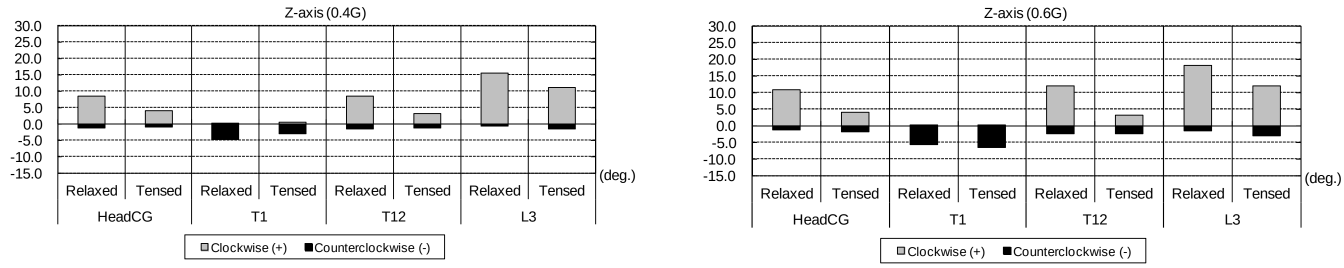
The result of the volunteer experiment indicates that the posture change caused by the side impact was lateral flexion motion and Neck (T1), thoracic spine (T12) and lumber spine (L3) were mainly rotated. Moreover, the posture-control effect of the lateral flexion due to muscle tension was around 20-40% at the head and neck region.



(a) X-axis rotation (lateral flexion)



(b) Y-axis rotation (flexion and extension)



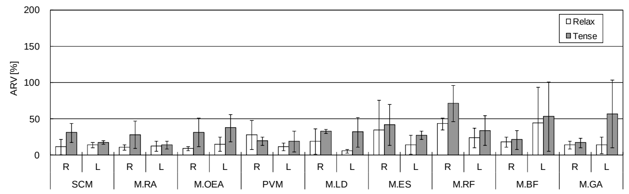
(c) Z-axis rotation (turning)

Fig. 18- Maximum rotational angle of each point (HeadCG, T1, T12, L3)

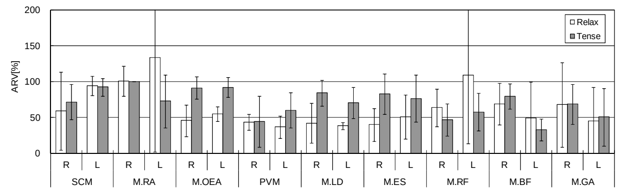
Muscle Activities During Low-speed Side Impact

Instantaneous mean (SD) muscle activation [5] is calculated based on the normalized ARV value (%) for defining the muscle activation level of the volunteer. Activation levels are categorized in the four phases defined in the previous section and expressed as a percentage of the maximum possible activation level in each phase as shown in Fig. 19. The listed muscles are mainly working against the sideward motion and the level of muscle activation in the tensed case is larger than that of the relaxed case except M. Sternocleidomastoideus (SCM), M. Rectus Abdominis (M.RA), M. Rectus Femoras (M.RF) and M. Biceps Femoris (M.BF) in phase 2 (Muscle active phase). In Phase 3, the activation level of M. Sternocleidomastoideus (SCM) and M. Rectus Abdominis (M.RA) is more than 100% with large variation. Compared to the muscle tensed case, the relaxed volunteer tends to delay the muscle activation and does not reach the higher activation level at Phase 2 and Phase 3. These muscle activities were strongly related to the kinematic of the head-neck and upper torso.

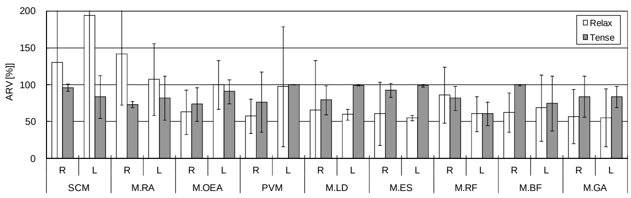
Based on the volunteer test, the muscle activities during pre-braking were predicted in each phase, and the muscles close to the body trunk were mainly working against the sideward motion. The relationship of the posture-control with muscle activation was identified based on the activation level. These muscle activations should be taken into account when predicting this pre-impact phenomenon with the computer model.



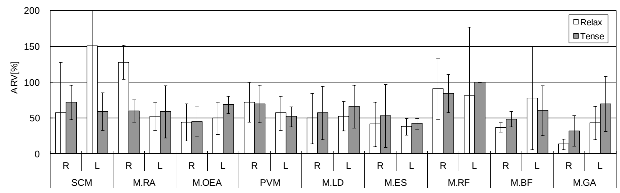
(a) Phase 1



(b) Phase 2



(c) Phase 3



(d) Phase 4

Fig. 19- Instantaneous mean muscle activation levels in the low-speed side impact

V. DISCUSSION

Mechanisms of Posture Change during Low-speed Side Impact

In the current study, the lateral motion of human volunteer was evaluated in low-speed side impact test. Each subject was applied two low-speed sled impulse (4.0m/s^2, 6.0m/s^2, duration is set 600ms). Each volunteer participated in two muscle condition (relaxed and tensed) at each impact level. According to the posture change captured by the high speed camera as shown in the result, the possible lateral motion due to low-speed side impact was described in this test. Over all it was confirmed that the 6.0m/s^2 of the impact significantly affect the kinematic response of subject compare to the case of 4.0m/s^2. Additionally, all the muscles selected in this side impact test were activated in the lateral motion. Therefore, the muscles were contributed to the posture control effect in the low-speed impact especially at the muscle tensed case.

In the tensed case, due to the muscle force, the volunteer's seating posture at the time of 500ms is bending less sideward by 5 degree at 4.0m/s^2 impact acceleration, 10 degrees at 6.0m/s^2 impact acceleration (the relative angle of the torso against the seat) than the relaxed case. From this phenomenon, it can be considered that the posture change employed in this experiment reproduces the occupant's sideward-bending behavior (relaxed and tensed) at the time of swerving/steering when the occupant is restrained by the standard three point seat belt because the shoulder belt is not a factor in pure lateral motion.

There exists an adequate correlation between the discharge of muscle force and the acceleration of each body part, as estimated from the results of the measurement system. For example, the head-neck-torso acceleration (HeadCG, T1, T12, L3: Fig. 10, Fig. 15) increases in Phase 3, but decreases or remains constant in Phase 4. The muscle tensed effect is clearly seen in the magnitude of acceleration and angular velocity of HeadCG , T1 and T12 compared to the relaxed muscle condition. This is due not only to the muscles close to the body trunk (neck and torso), but also to the lower extremity muscles being discharged from the impact (0 ms). In other words, the upper torso was subjected to posture-control provided by this pre-tensed muscle condition in which the activation level is around 20-60 % of maximum muscle. Following the timing of the muscle activation with lower extremity such as M. Rectus Femoris (M. RF) and M. Biceps Frmoris (M. BF), the belt and the right 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 belt and footplate in phase 3 and phase 4.

Difference in Head, Neck and Torso Motion Related to the Muscle Response

In this study, a two-point belt was installed to constrain the hip in order to model simple conditions. It was identified that the pre-acceleration tension of muscles affected the physical motions when compared to the relaxed case. This is because of the initial muscle activation detected in the EMG. When comparing rotational angles between tensed and relaxed cases by body region from pelvis to head, ante-lateral-flexion motion due to muscle tension was indicated at the head-neck region in the subjects (HeadCG, T1: Fig. 18). In this region, the posture-control effect of the rotational angle due to muscle tension was 42 % reduction for the head (HeadCG) and 18% reduction for the neck (T1) in the lateral flexion motion. On the other hand, the torso (T12) and hip region (L3) did not show any anti-lateral-flexion motion but indicated the reduction of turning motion in the muscle tensed case. As a result of this phenomenon, M. Obliquus Externus Abdominis (M.OEA) which is related to spine (lumber) movement effectively works for the turning motion relative to the lateral flexion.

According to a previous study [19] with cases involving a lap-belt, the hip region showed the largest flexion motion in the frontal pre-impact test with volunteers. Consequently, the rotational angle of the hip region is seen to be strongly affected by the upper torso motion. The same tendency is seen in the lateral pre-impact test at the hip region. Therefore, the restraint effect of the hip is important when discussing the stability of the posture under low-speed acceleration.

Muscle Activities

The pattern diagram of the spine related muscle activation with seat pressure distribution in the relaxed case is shown in Fig. 20. This figure visualizes muscle activities by using the result of low-speed side impact test (Relaxed, 6.0m/s^2) as shown in Fig. 17 for understanding the relationship between muscle activities and physical motion. The color bar in this figure emphasizes the level of muscle activation with respect to the physical motion of the volunteers in the low-speed side-impact sled test. According to the kinematics, the subject's upper torso and head-neck starts to move sideward 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. Sternocleidomastoideus (SCM: Fig. 17) starts around 120 ms. Simultaneously the position of the body's center of gravity moves away from the midline of the body and the muscles close to the body trunk such as M. Rectus Abdominis (M.RA: Fig. 17) are activated to generate the resistance force to the spine motion. In addition, the left side of M. Obliquus Externus Abdominis and M. Paravertebralis (M.OEA, PVM: Fig. 17) respond in Phase 3 due to the thorax moving to the side toward the hip. Then, the right side of seat sensor indicates the higher pressure distribution in the period of phase 3 through phase 4. This diagram shows a good correlation with the muscle activity which is estimated from instantaneous mean muscle activation levels (Fig. 17). The muscle activity of M. Sternocleidomastoideus (SCM), M. Rectus Abdominis (M.RA), M. Obliquus Externus Abdominis (M.OEA) and M. Paravertebralis (PVM) were close to 100% or more than that value during Phase 2 and Phase 3. This is because of the relaxed condition. 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. This muscle activation controls excessive physical motion and activations reach much higher levels with a wave shape.

Limitation of this Study and Suggestion for Further Research

It is necessary to increase the number of subjects and analyze the muscle activity to maintain consistency. The initial posture reconstructed in this experiment was the occupants seated as a passenger in a car. Therefore, the posture of the driver at the moment of pre-braking just before the impact should be examined with a steering wheel. In addition, each muscle response was normalized with its own maximum muscle activation value (ARV). However, the muscle activation of each muscle has a variation even though the subjects were instructed to tense all their muscles intentionally in the muscle tensed cases. Therefore, maximum muscle activation levels need to be verified from various perspectives based on the experimental data.

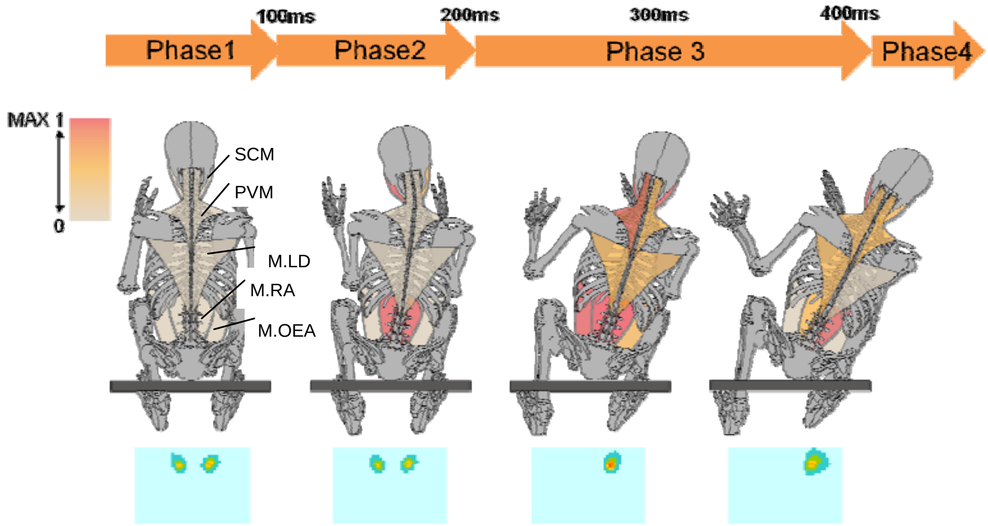


Fig. 20- Physical motion of the volunteer with muscle activity at the pre-impact steering/swerving

VI. CONCLUSIONS

In this study, the effects of muscle tension on the head, neck and torso motions were explained in four phases. Furthermore, it was identified in this study that the effect of the difference in muscle activity governs the motion of each phase based on the acceleration and EMG electrodes. Finally, neck, abdomen and back muscles such as M. Sternocleidomastoideus (SCM), M. Rectus Abdominis (M.RA), M. Obliquus Externus Abdominis (M.OEA) and M. Paravertebralis (PVM) muscles were mainly activated when the occupant made a pre-impact swerving/steering action. These parameters such as the muscle activation are important factors in discussing the subject's motion just before collision. This in turn indicates that the activation level measured from EMG data of major muscles were significantly useful for the prediction of occupant motion, since all these variables were directly measured in the volunteer test.

VII. ACKNOWLEDGEMENT

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