A Study on Occupant Kinematics Behaviour and Muscle Activities during Pre-Impact Braking Based on Volunteer Tests

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ABSTRACT

The objective of the current study is to analyze the influence of changes of the physical motion on human body under low-level impact accelerations. Five human volunteers in good health participated in the experiment under the supervision of an ethics committee. Each subject sat on a seat mounted on the sled that glided forward on the rails and simulated actual car motion when the passenger applied the emergency brake. Applied acceleration was around 1.0 g. During the experiments, the physical motion (measured by three-dimensional motion capturing system), acceleration, and EMG signals were recorded. In the relaxed case, muscle responses were observed to start activating at around 100ms after the impact, when the pelvis acceleration was at its peak value due to the interaction between the lumbar spine and the seat. Depending on the location within the body, the reflex time of each muscle is different. Furthermore, the head-neck-torso accelerations were strongly influenced by the muscle activities after impact. These results can contribute in clarifying the relationship between the physical motion and muscle forces with respect to the difference in activation level of muscles. This study was also design to establish an injury prediction approach to verify the influence of human body posture changes just before impact on the occupant injuries in a traffic accident.

Key Words: Impact Biomechanics, Effect of Muscles, Low-speed front impact, Pre-braking.

DRIVER'S SEAT OCCUPANTS TAKE VARIOUS POSTURES due to their age, gender and physique. Moreover, further posture changes occur just before the collision due to occupant evasive maneuvers. Therefore, it is difficult to keep the standard posture such as that of the Anthropomorphic Test Dummy (ATD). **Figure 1** shows the accident type and evasive maneuvers obtained from the Institute for Traffic Research and Data Analysis (ITARDA) in Japan (1993-2004). This data sources consist of 860 cases of front impact collision (CDC:11F-1F). According to the accident analysis [1], 60% of the drivers made evasive maneuvers (braking or swerving) in front accident cases. Namely, most drivers made evasive maneuvers just before the collision. 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 by an individual occupant or posture differences.

The effect due to driving posture and passenger individuality have been studied in PRISM (Proposed Reduction of Car Crash Injuries through Improved Project Smart Restrain Development Technologies) [2][3]. In this study, the accident scenario and the posture change caused by pre-braking were investigated using accident data and the volunteer tests. Furthermore, the posture change effect was predicted by computer simulation [4]. However, 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 examined. According to volunteer tests at the low-speed impact [5][6][7], muscle activation strongly affects to the motion of the human body. Therefore, detailed examination of this muscle effect is needed.

In this study, the posture of the driver at the moment of pre-braking just before the impact was examined in low-speed frontal impact tests with volunteers. At the same time, the basic data of posture changes and muscle activation were measured for the simulation with the computer human model. To begin with, the posture changes were captured by a 3D motion capturing system, and muscle activation was also measured by electromyography. Based on the results of the 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.

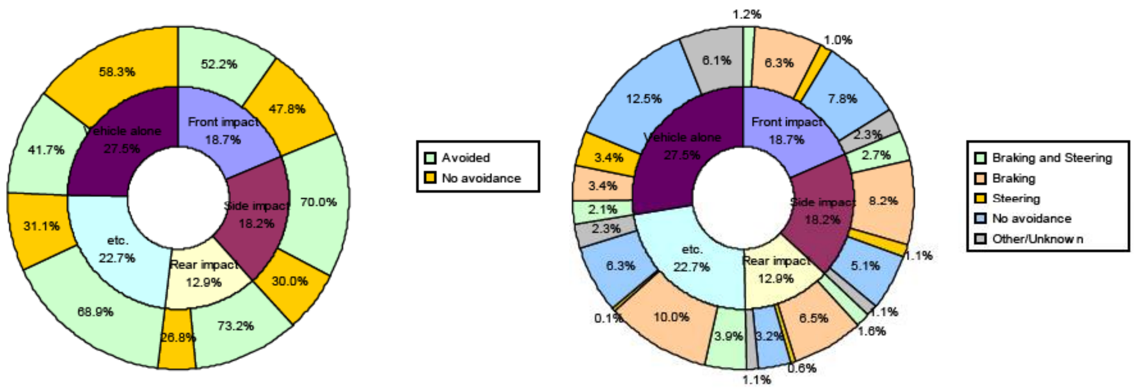


Figure 1. Accident type and evasive maneuver

METHODS OF EXPERIMENT

VOLUNTEERS AND INFORMED CONSENT – Five healthy 23-year-old volunteers (three 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. The physical data of these subjects are shown in **Table 1**. The position of the head C.G. was located 5mm in front of the external auditory meatus and 20mm above the Frankfurt line that connects the lower orbital margin and the center of auditory meatus [8]. Then, X axis was defined as parallel to the Frankfurt plane and Z axis was perpendicular to the X axis (**Figure 2**). The location of acceleration and occipital condole was measured based on the X-Z coordinate system as shown in **Table 1**. Also, the head mass and the inertia moment were calculated based on the regression curves [9][10].

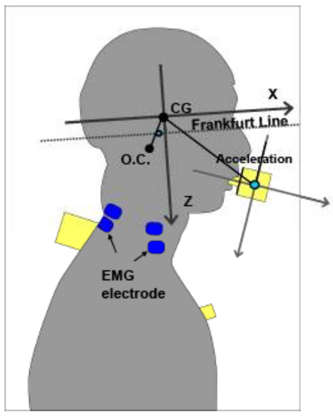


Figure 2. X-Z Coordinate system

Table 1. Subjects

Volunteer	Height(cm)	Weight(kg)	Tragion to vertex height(mm)	Location of Acceleration(mm)		Location of Occipital		Head mass(kg)	Head inertia moment(kg・cm^2		
				X	Z	X	Z		X	Y	Z
YSM	177.7	68	133	130.2	55.4	-8.3	41.6	4.0	194.8	171.9	119.8
AOM	173	63	146.8	121.9	66.5	-19.4	38.8	4.3	220.5	196.5	131.6
SWM	176	67	138.8	128.5	66.8	-15.4	37.3	4.2	208.6	185.0	126.2
TIF	172	62	146.5	138.8	57.8	-15.4	38.8	4.0	200.6	177.3	120.8
AMF	159	53	126.4	119.5	58	-18.6	40.6	3.3	140.0	121.3	90.3

SLED APPARATUS FOR SIMULATION OF LOW-LEVEL IMPACT: In the first place, **Figure 3** illustrates the initial outlook of the front-impact simulation sled system (hereafter referred to as "Mini Sled"). The mini sled was designed based on the actual car pre-impact experiments, and an oil shock damper was installed in order to simulate the deceleration when the passenger applied emergency brakes. The length of the rail was 4m, and the sled, which has 10-degree slant against the ground, slides on the rails at a maximum speed of 6.5 km/h. The collision of the Mini Sled against the damper set at the end of the rail simulates a real front collision. The sled has a rigid seat (hereafter referred to as "R-seat"), which does not include a head restraint.

Low-level frontal impact was applied to the volunteer by collision with the damper. To change the sliding speed and damping coefficients of the damper, various impact level (0.2-1.0g with the duration of around 200 ms) were applied to the volunteer. The R-seat made of steel was mounted on the sled. In this experiment, the effect of muscle forces coming from the lower limbs was neglected by tying the legs to the sled. For safety purposes, a seatbelt was used to immobilize the waist. In this case, the waist seatbelt was adjusted to the length of abdominal region, thus these belts were not pre-tensioned at the initial stage. The muscles were conditioned to be relaxed or tensed, and three pre-tests were conducted to check the repeatability in each condition.

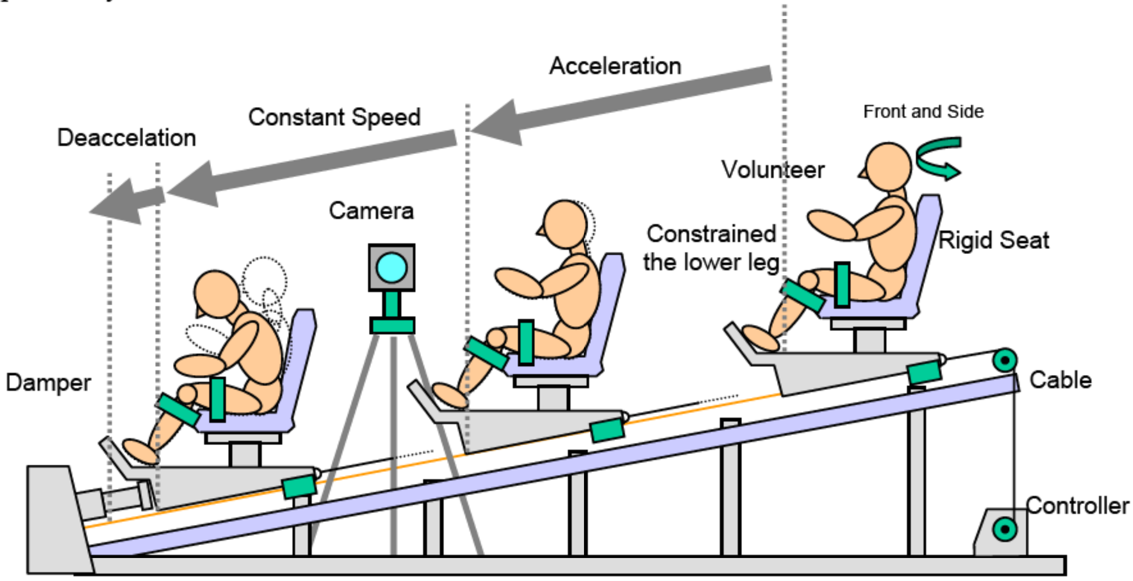


Figure 3. Outlook of the front-impact simulation sled system

MOTION CARPTURING 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 the video image caught with two or more cameras (Eagle Digital Camera)[11] and is translated into three-dimensional coordinates. The resolution of the camera is 1280 x 1024 pixels. The images were incorporated into Calcium (NAC Inc.) and analyzed. This technology on capturing motion is also widely used in computer graphics or medical researches, and is an effective as a technology of collecting human motions. In the experiment, eight sets of cameras were used and the landmarks were attached to a typical body parts. The arrangement of the landmark in this experiment was attached to the head (Parietal, Tragic point), shoulder (Acromion), chest (Rib12th, Sternum), back (T4, T10), lumber (Iliac Crest), arm (Elbow), hand (Wrist, Back) and leg (Knee) as shown in **Figure 3**. These markers were used as the reference points for the determination of the head, neck, torso, abdomen, hip and lower extremities.

ACCELERATION MEASURMENT: In order to monitor the motion of the volunteer at the time of an impact, Tri-axial accelerometers were placed on the a body surface and the sled.

SLED ACCELERATION: Tri-axial 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.

HEAD ACCELERATION: Since the head motion was three-dimensional, tri-axial accelerometers and the tri-axial angular velocity meter attached to the mouth via a mouthpiece were used for the measurement. The fixture shown in **Figure 2** was fabricated for the installation of accelerometers on the head of each subject.

T1, SHOILDER, CHEST AND PELVIS ACCELERATION: The acceleration of the first thoracic vertebra, shoulder, chest and pelvis were measured by the tri-accelerometer attached to the surface of the first thoracic vertebra, the acromion, the front chest around the sternum region, and the iliac crest with a surgical tape over which double-coated tape was adhered as shown in **Figure 5**.

ELECTOROMYOGRAPY: The muscle condition of each subject was measured in the relaxed state on the preliminary test of this experiment. Muscle activities were measured by means of surface electromyogram synchronized with the three-dimensional movement. EMG electrodes were attached onto the skin over major muscles of the subject. The electrodes with diameter of 5mm were arranged as bi-polar electrodes with a distance of roughly 2cm between the electrode centers. The locations of the surface electrode are shown in **Figure 4**.
[Neck] Streoclaidomastoid, Parvertrabral muscle
[Torso] M. Latissimus Dorsi
[Abdomen] M. Rectus Abdominis, M Obliquus Externus Abdomains
[Leg] M. Biceps Femoris, M.Rrectus femoris, M. tibialis anterior

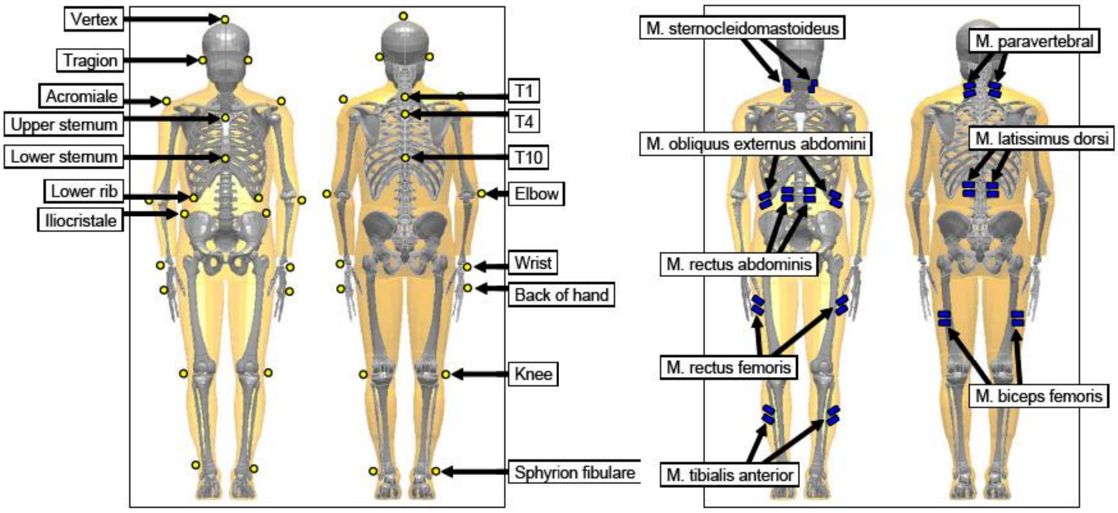


Figure 4. Location of the landmark and electromyography

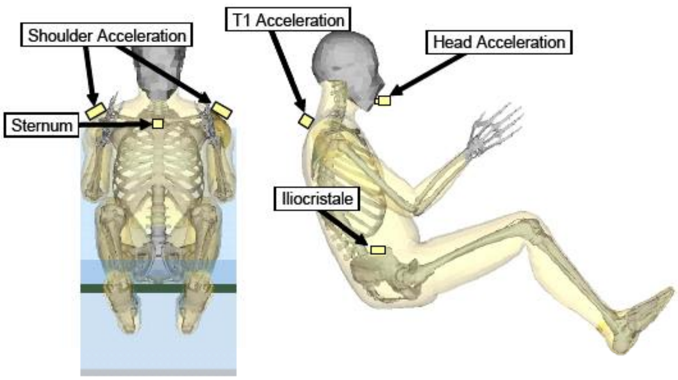


Figure 5. Frontal and lateral view of the head/neck/torso/pelvis with mounted accelerometer

EXPERIMENTAL CONDITIONS: Three healthy males and two healthy females were selected as test subjects. In order to examine the effect of muscle action 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 tense state, in which volunteers intentionally tensed up their muscles. Test subjects were instructed regarding their muscle condition. During the test, the muscle activation was monitored whether 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 hands, 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 maintain their initial posture even though all the muscles are tensed. Applying the impact to the same initial posture, the difference of the muscle activation was clearly defined by the motion of upper torso.

Table 2. Test Matrix

	Impact acceleration	Direction	Muscle condition
5 adult (3 male 2 female)	0.2 G	Front	Relaxed
	0.6 G		
	1.0 G		Tensed

ANALYSIS

DEFINITION OF SEGMENT AND JOINT ALONG THE FULL BODY REGION: The physical motion on human body and head-neck-torso kinematics at the low-level impact acceleration were measured by three-dimensional motion capturing system. In this motion capturing system, a stick figure is generated based on segments that are consists by the landmark on the body surface. Each segment of the head, neck, torso, abdomen, thigh, and the lower legs are defined as follows and shown in **Figure 6**. With this segment motion, the rotational angle at the joint was recorded, and the differences between subsequent rotational angles were calculated.

Head segment (S1) : the segment between a vertex and head C.G.
Neck segment (S2) : the segment between head C.G. and the first thoracic vertebra
Torso segment (S3) : the segment between the first thoracic vertebra and the 10^{th} thoracic vertebra.
Abdomen segment (S4) : the segment between the 10^{th} thoracic vertebra and the iliac crest.
Thigh segment (S5) : the segment between the iliac crest and the knee joint
Lower segment (S6) : the segment between the knee joint and the ankle

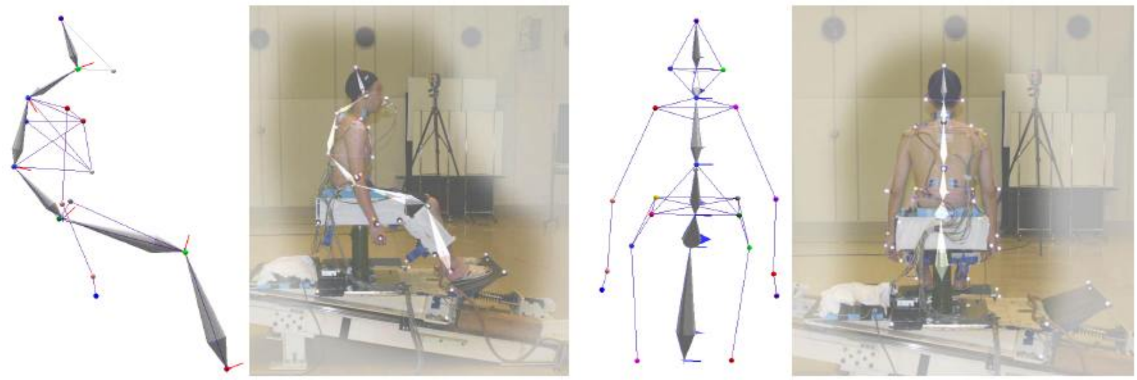


Figure 6. Stick picture from the motion capturing system

HEAD ACCELERATION AND NECK LOAD - The location of anatomic center of gravity of the head was determined using the methods reported by Ono et al [8]. 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 with the combination of the tri-axial accelerometer and the 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 with this method.

EMG SIGNAL PROCESSING – The major muscle activation during the impact was monitored with the surface electrodes and analyzed with the systematic processing. At first, the raw EMG signals were filtered with band pass filter (low pass filter: 500Hz, high pass filter 25Hz). Then, the full wave rectification was applied to each signal. Finally, the smoothing was applied and the average rectified value (ARV) was obtained. Each muscle was normalized with their own maximum muscle activation value (ARV) in the relaxed case. In this study, the muscle reflex time was defined based on the assumption that the criteria of activation is more than 0.4 (normalized ARV) in the EMG signal after the collision (Time=0.0). In addition, the reflex time was always zero in the tensed muscle case, because of the subjects intentionally activated their muscle before the collision.

RESULTS

SUBJECT'S MOTION, ACCELERATION REPONSE AND EMG: In order to investigate the effect of the impact level and muscle condition, a series of experiments were conducted on the five 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. Subject's motion, acceleration response, and EMG are divided into four phases. In this paper, results of the experiments conducted with the impact acceleration of 1.0 g using a rigid seat are described. The following results were summarized according to each phase of impact as sequential changes with time: 1) the motions observed by three-dimensional movement analysis system, 2) acceleration at each region of the subject and loads against the neck, and 3) the electromyographic responses.

Figure 7 shows the sequential images of a subject's motion taken by the 3D motion capturing system. In addition, subject's motion, response to the acceleration and EMG are divided into four phases as presented in **Figure 8**. This figure shows the time histories of acceleration of the sled, head, shoulder, chest, and hip for the X directions. Moreover, the time histories of EMG response of each muscle of the subject are indicated. Finally, the time of collision is set at zero (0ms) in the time history diagram.

Phase 1 (0 -100ms, Initial Response Phase)

1) No significant motion was identified in this phase although the 1.0 g deceleration was applied to the sled during this phase.
2) After the impact, the dumper decelerates the sled, and maximum acceleration was around 1.0(m/s^2) (Seld_Acc : **Figure 8(a)**). This deceleration appeared with the acceleration on the X-direction of hip (Hip_Acc : **Figure 8(b)**) which was close to the seatbelt. Then, no major deceleration was shown in this phase except for the hip.
3) The major muscle activation was not detected in the monitored muscles.

Phase 2（100 - 200ms, Muscle Active Phase）

1) Because of the inertial force from the deceleration of the sled, the subject's upper torso started to move forward, and the head was moving backward relative to the first thoracic spine (T1). As a result of this phenomenon, the neck that links the head and torso started to extend. Moreover, T10 and T1 showed a ramping up motion when the sled was stopped by the damper (200ms).
2) The hip acceleration (Hip_Acc : **Figure 8(b)**) reached maximum in this phase, and this acceleration was transferred to the chest-shoulder and T1-head one by one. According to the angular velocity of T1 and the iliac crest (T1_Rot_Vel, Hip_Rot_Vel : **Figure 8(c)**), T1 and hip joint started to rotate around 100ms. The head also started to rotate at 150ms (Head_Rot_Vel : **Figure 8(c)**). On the other hand, the angular acceleration of the head indicates a positive value (extension) because of the initial extension of the head (Head_Rot_Acc : **Figure 8(c)**). Then the negative (flexional) value increased around 200ms.
3) In relation to the neck link motions, discharge of sternoclaidomastoid muscles started around 100 ms (SCM : **Figure 8(e)**). Moreover, the position of the upper torso moved forward, and M. Latissimus Dorsi muscles started to discharge (M_LD : **Figure 8(g)**).

Phase 3（200 -400ms, Forward Motion Phase）

1) With the subject's lower part restrained to the sled with a belt, the arched rotation of the upper torso started at around 200 ms. Simultaneously, the neck also started to rotate (flexion) in this

phase. The T1 movement was synchronized with the neck shear force (Shear: **Figure 8(d)**) and electromyogram (PVM: **Figure 8(f)**). Then, the maximum value was indicated during 200ms-300ms. In addition, the rotational flexion angle of the head and neck (Head_Rot_Vel, T1_Rot_Vel : **Figure 7(c)**) reached maximum at around 300ms, even though the spine deformation decreased.

2) The magnitude of chest and shoulder acceleration (Chest_Acc, Shoulder_Acc : **Figure 8(b)**) indicated the maximum value at around 250 ms, and the head and T1 acceleration (Head_Acc, T1_Acc : **Figure 8(b)**) also showed maximum value at around 350 ms. The magnitude of these acceleration decreased due to the activation of the muscles. On the other hand, the hip angular velocity (Hip_Rot_Vel : **Figure 8(c)**) converged to zero, and the magnitude of T1 and head C.G. angular velocity(Head_Rot_Vel, T1_Rot_Vel : **Figure 8(c)**) reached a maximum value during this phase.

3) The muscle discharge of the abdomen (M_OEA: **Figure 8(h)**) disappeared at around 350 ms and the discharge of the paravertebral muscles (PVM: **Figure 8(f)**), M. latissimus dorsi (M_LD: **Figure 8(g)**) maximizes at around 250-400ms.

Phase 4（400ms → End, Final Phase）

1) The upper torso started to resume the original position.
2) The accelerations of head, neck, and torso converged to zero.
3) The muscular discharge of the neck, torso, and leg disappeared.

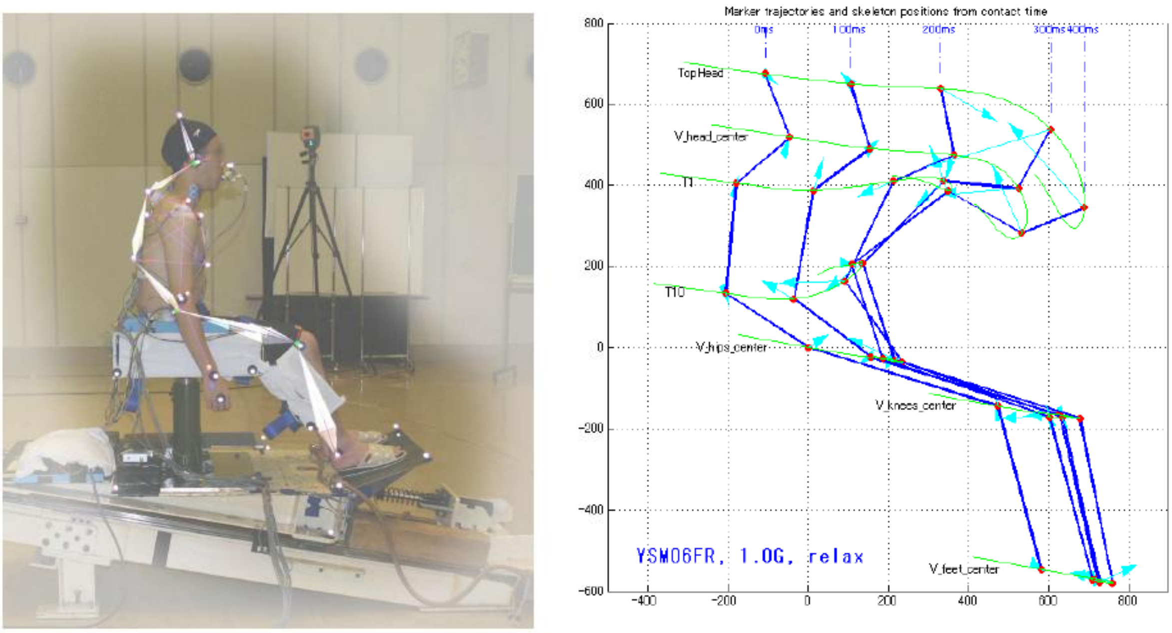


Figure 7. Captured volunteer motion with motion capturing system

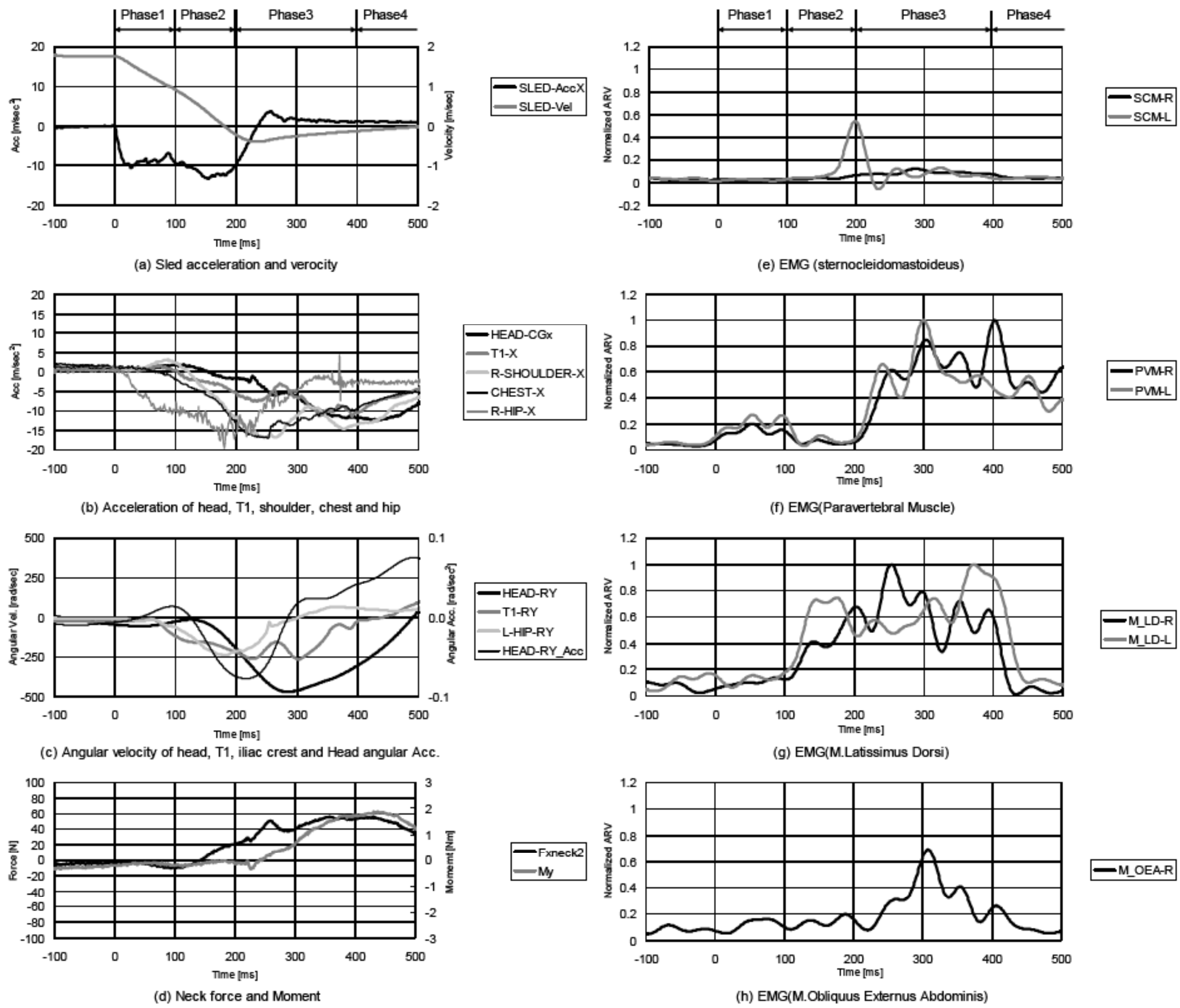
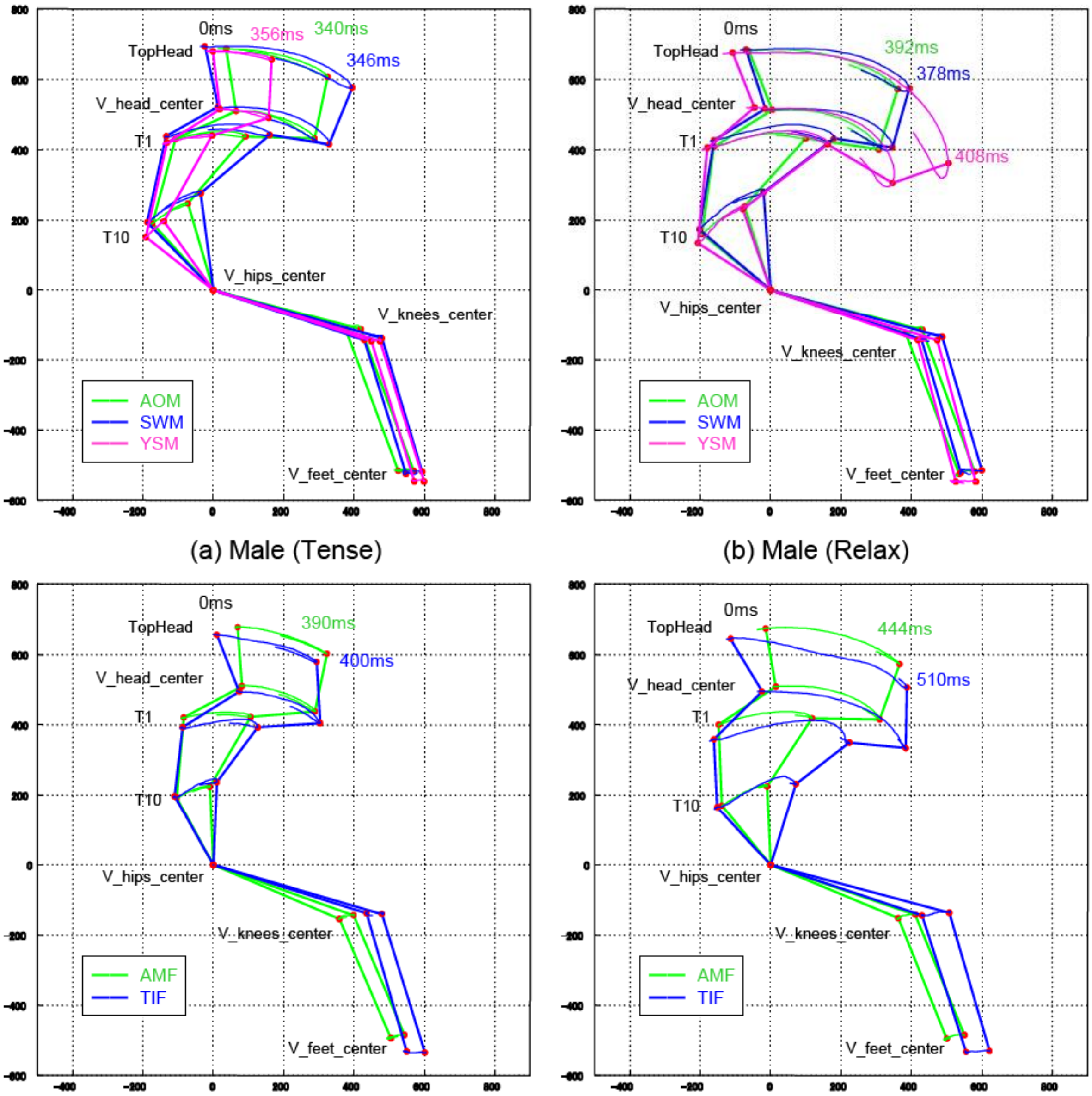


Figure 8. Time history of the acceleration, velocity, angular velocity, angular acceleration, neck force, and EMG response (SCM, PVM, M. LD and M. OEA) on the relaxed subject

KINEMATICS OF THE UPPER TORSO WITH RESPECT TO RELAXED AND TENSED MUSCLE CONDITION: **Figure 9** shows the trajectory of the subjects (three males and two females) under different muscle conditions in 1.0g case. In these figures, the hip-center is fixed to the origin, and the motion of each landmark follows to the hip-center. The pair of initial posture and most forwarded posture is shown with the solid line in each male and female subject. The trajectory is also 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. 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 (**Figure 9(a)**-**(b)**). Compared to the relaxed muscle case, the motion of upper torso was controlled by the muscles in the tensed case (**Figure 9(a)**). On the other hand, the female subjects showed the different trajectory from the males, especially for TIF (**Figure 9(c)**-**(d)**). In addition, the difference in the trajectory was very wide in the females subjects.



(C) Female (Tense) (d) Female (Relax)

Figure 9. Trajectory of the five subjects (three males and two females) in 1.0g case

DIFFERENCES IN HEAD, NECK, AND TORSO MOTIONS RELATED TO THE MUSCLE RESPONSES: 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. **Figure 10** indicates the location of landmarks and segments. In order to represent the hip motion separately, a virtual marker was created based on the seat marker. To be concrete, the upper torso was separated by five segments (Head, Neck, Chest, Abdomen, Hip), and the joint angle at the connection point (Head-C.G., T1, T10, Iliac-Crest, Trochanter (virtual marker) was calculated with the motion capturing software. In the volunteer test, each volunteer's lower extremity is tied to the rigid seat during the experiments. Therefore, we could assume that the trochanter is moving with sled as one rigid body.

Figure 11 shows the average value of the maximum flexional and extensional angle at the joint in each impact level. The average value was calculated from the three male volunteers. For the purpose of comparison, the tensed and relaxed muscle cases are shown in the same figure. 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 lower extremity was tied to the sled, the major angle difference of lower limb could not be seen in the experiment.

In the muscle tension case, the major flexional motion was detected in the abdomen and hip area, and the rest of the body part slightly showed extensional motion. On the other hand, in the case of relaxed condition, the flexional motion of the hip region was also dominated similar to the tensed case. Because of the initial extensional motion in Phase 2, the extensional (+) rotation was detected in the head, neck, and chest. Then, the head-neck flexional motion which was not seen in the tensed case was clearly revealed in the relaxed case.

- Head-C.G. : Angle between vertex-targion and tragion-T1
- T1 : Angle between tragion - T1 and T1 - T10
- T10 : Angle between T1 - T10 and T10 - Iliac Crest
- Iliac-Crest : Angle between T10 - Iliac Crest and Iliac Crest -Trochanter
- Trochanter : Angle between Iliac Crest –Trochanter and sheet horizontal line

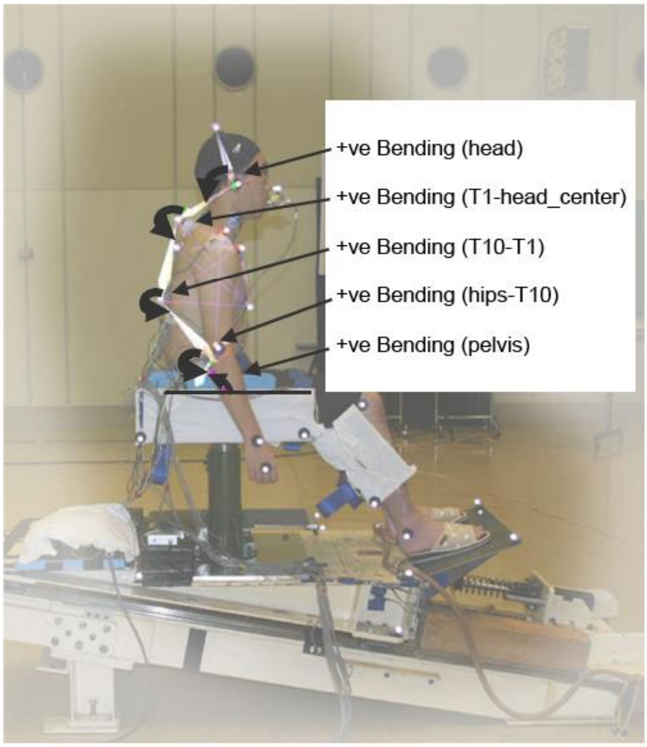
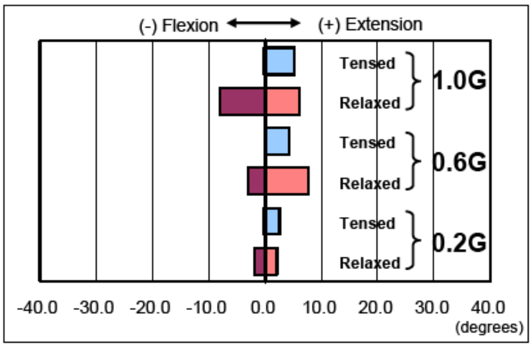
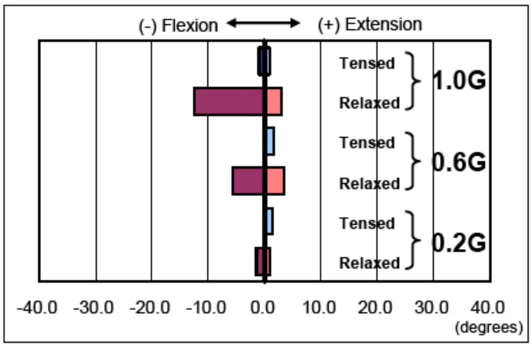


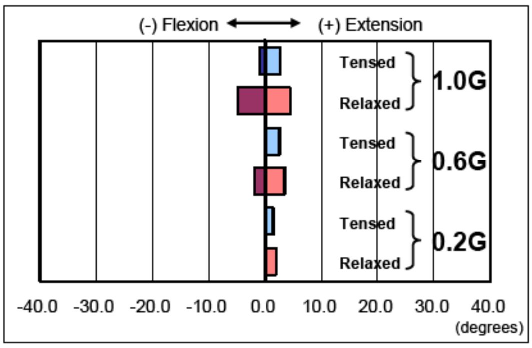
Figure 10. Definition of the rotational angle between each segment.



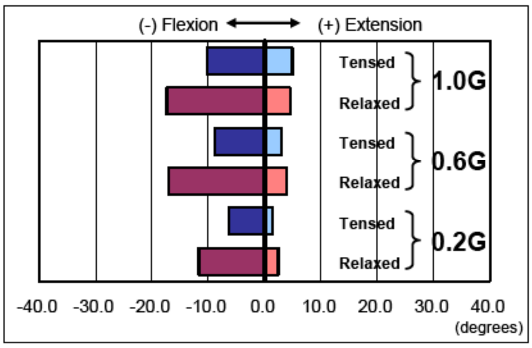
(a) Head C.G. (Head)



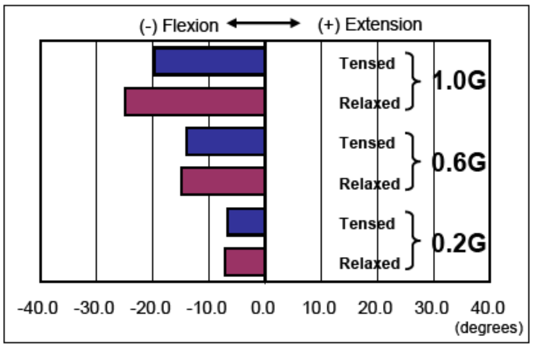
(b) T1 (Neck)



(c) T10 (Chest)



(d) Iliac-crest(Abdomen)



(e) Trochanter (Hip)

Figure 11. Maximum flexion and extension angle of each joint with the muscle tensed and relaxed case

PHYSICAL MOTIONS AT EACH IMPACT LEVEL BETWEEN MALE AND FEMALE: In order to compare the displacement of posture changes during each impact level, the trajectory of head, T1 and T10 was measured with the motion capturing system. The definition of the displacement is the length of each landmark traveling from time zero to the most flexional posture in each volunteer test. **Figure 12** shows the trajectory of the vertex, head center, T1, and T10 respectively to the hip joint. The black line indicates the relaxed trajectory (0 ms - 408ms), and the gray line shows the tensed case (0ms - 356ms). The male and female average displacement of each landmark (head, T1, T10) is indicated in **Figure 13**. In this figure, each of the muscle conditions (relaxed and tensed) are shown separately. The displacement of each landmark tends to increase as the impact level increases. Also, compared to the tensed muscle case, the displacement of each landmark travels a long distance at each impact level.

Figure 14 shows the posture-control effect on the posture changes with the muscle activation. The definition of posture-control effect of muscle pretension is the difference of the length between muscle tensed and relaxed divided by the length of relaxed case. The length discussed in the figure is the arched length which is described in the previous section. The high posture-control effect means that the movement is constrained by the muscle activation. This figure indicates the difference between the male and female volunteers at each impact level respectively to the head, T1 and T10. Compared to females, the males have a constrained the posture due to the muscle effect. This difference was clearly seen in the low-level impact. The reason for this difference between males and females is the thickness of muscles. Moreover, this is the common phenomenon in males and females that the larger posture-control effect of the posture changes is seen at the higher impact levels. In other words, the volunteers automatically control the muscle force depending on the impact level.

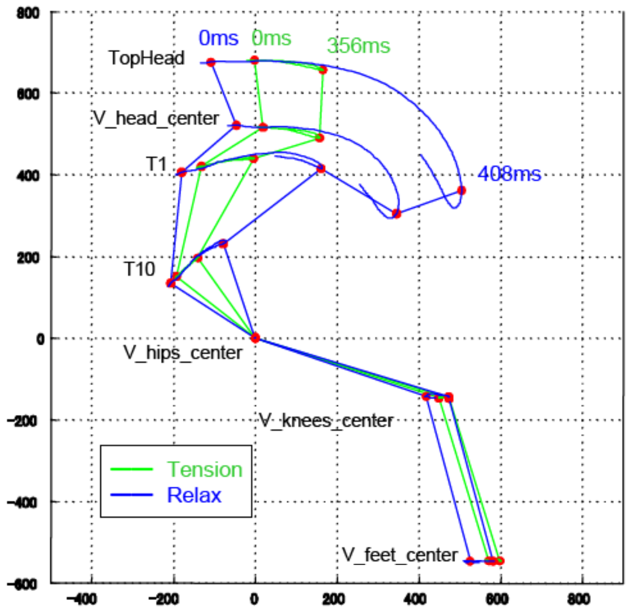
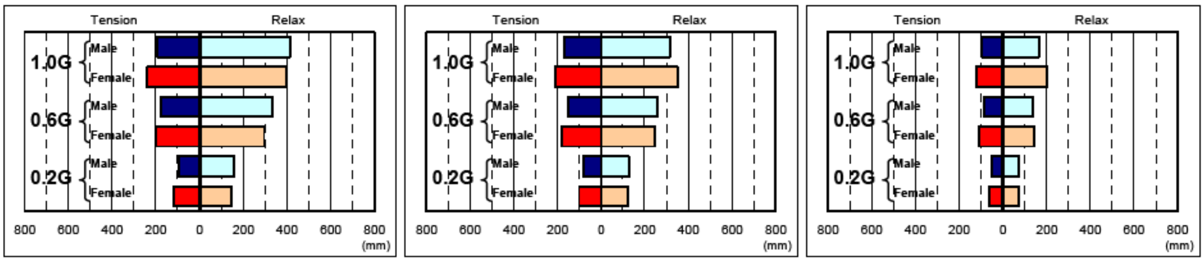


Figure 12. Definition of the length of trajectory in each landmark



(a) Head C.G. (Head) (b) T1 (Neck) (c) T10

Figure 13. Trajectory of head C.G., T1 and T10 between male and female.

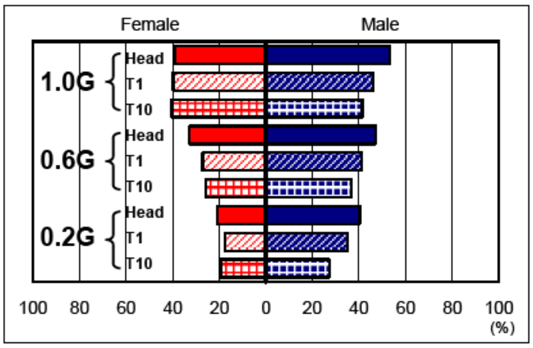


Figure 14. Posture-control effect of muscular pretension with the upper torso between male and female

DISCUSSION

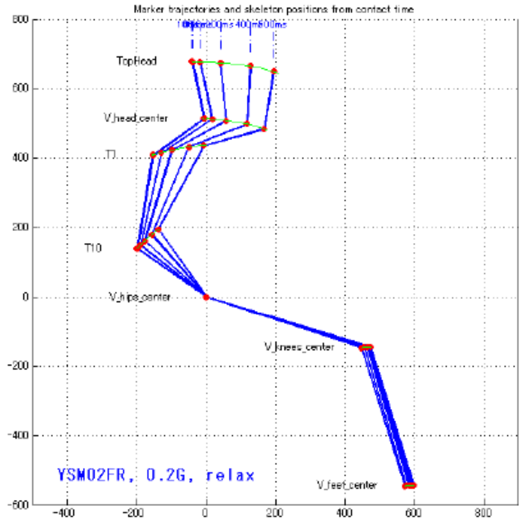
MUSCLE ACTIVITIES DURING PRE-BRAKING: **Figure 15** shows the sequential images of forward motion (100ms interval during 0-500ms), while **Figure 16** shows the time history of the muscle activation at each impact level.

During the muscle active phase (Phase 2: 100 - 200ms after the impact), the subject's upper torso starts to move forward and the head moving backward relative to the first thoracic spine (T1). As a result of this phenomenon, the neck that is the link between the head and the torso starts to extend. In relation to these neck link motions, the discharge of sternoclaidomastoid muscles starts at Phase 2. The magnitude of the EMG is proportional to the head-neck link motions. Moreover, the position of the body center of gravity moves forward and M. latissimus dorsi muscles start to discharge. On the other hand, the forward motion phase (Phase 3 : 200 - 400ms after the impact) indicates different motions. Restraining the subject's lower part to the sled with the belt, the arched rotation of the upper torso shows around 200 ms. Simultaneously, the rotational flexion angle of the head and neck reaches maximum at around 300 ms. The X-direction of head acceleration indicates the maximum value at around 300 ms due to the activation of neck muscles. The muscle discharge of the abdomen disappears at around 300-350 ms, and the discharge of the paravertebral muscle, M. latissimus dorsi reaches maximum at around 250-400ms. Based on the volunteer test, the muscle activities during pre-braking was predicted, and the back muscle such as M. latissimus dorsi and paravertebral muscles were mainly working against the forward motion. These muscle activities were strongly related to the motion of the upper torso.

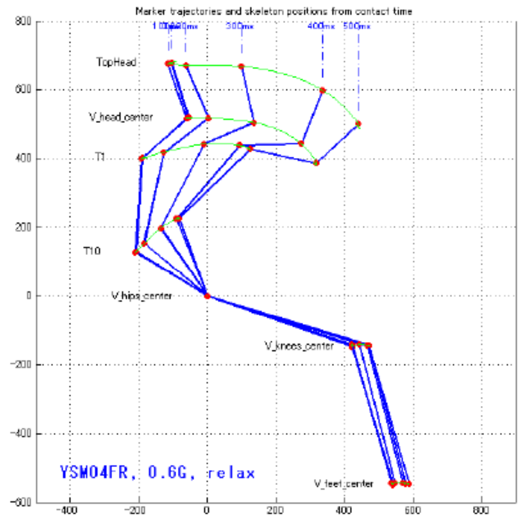
The limitation of the posture-control with muscle activation was not clearly identified due to the limited delta-V, and because the deceleration lasted only 200ms in this study. However, the forward motion is proportionally increased depending on the magnitude of the acceleration. Therefore, a larger delta-V would provide a larger deformation with the same trend of this study. These muscle activations should be taken into account when predicting this pre-braking phenomenon in the larger delta-V with the computer model.

EFFECT OF MUSCULAR TENSION: It was identified that the pre-impact tension of muscles affected the physical motions in the low speed impact. In addition, the effect of the electoromyographic responses hardly exists even though the volunteer relaxed their muscles before the impact. The EMG monitored in this study shows that the subjects were relaxed, and the muscular response affected the physical motions. As a result of rotational angle of the body region, the posture changes in muscle tensed anteflexional motion at the abdomen and hip region. The posture-control effect of the rotational angle due to the pre-impact tension of muscles was 40% in low speed impact (0.2g～1.0g). On the other hand, hip region did not show major muscle effect. In the relaxed muscle case, the hip region showed the largest anteflexional motion. Also, the head-neck region showed 10 degrees of flexion-extension motion. As a result, the rotational angle of hip region strongly affects the upper body motion for the front impact case. Therefore, the relation between the angle of hip and the related muscles is important when discussing the maintenance of the posture in the low speed impact.

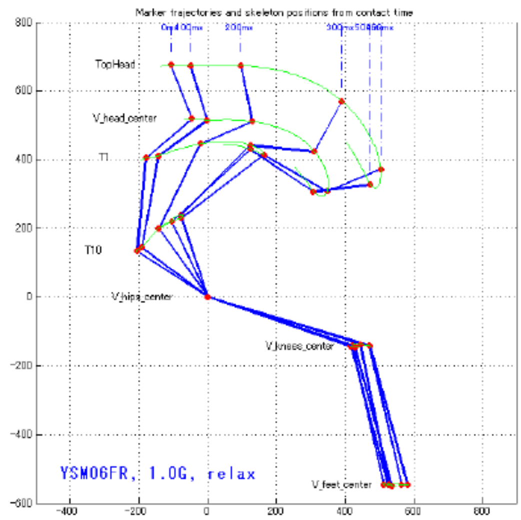
MECHANISMS OF POSTURE CHANGE: During the impact acceleration, the back muscles are extended as a result of forward motion of the upper torso and M. latissimus dorsi muscles when mainly discharged. There exists an adequate correlation between the discharge of muscle and the acceleration of each body part estimated from the results of measurements system. For example, the torso acceleration shows the positive value in Phase 2, but it becomes negative in Phase 3. This is because the back muscle (M. latissimus dorsi) is discharged from the 100 ms, and the upper torso was subjected to posture-control by these muscle forces. Following the upper torso motion that starts from the previous phase, the neck also starts showing the flexional motion. Similar to Phase 3, neck muscles activate to control this motion and the head acceleration decreased. Thus, not only the back muscles but also the neck muscles are activated to control the body motion. From the anatomical point of view, there are muscle spindles and a tendon organ acting as exterocepter that control the muscle activation. Especially, the muscle spindle that is located parallel to the muscle fiber changes its length due to the stretching of surrounded muscle fibers. Therefore, when the muscle fiber is extended, the muscle spindles transmit the signal to the spinal cords to decrease the muscle length. This activation is called the 'stretch receptor' and control the excessive motion in each body part. This protective mechanism works when the muscle spindle is applied in an unexpected impact. This phenomenon was observed in the volunteer test. Stretch receptor is strongly related to the mechanism of posture change, and the level of muscle activation is correlated with the posture-control effect at each impact level.



(a) 0.2 g

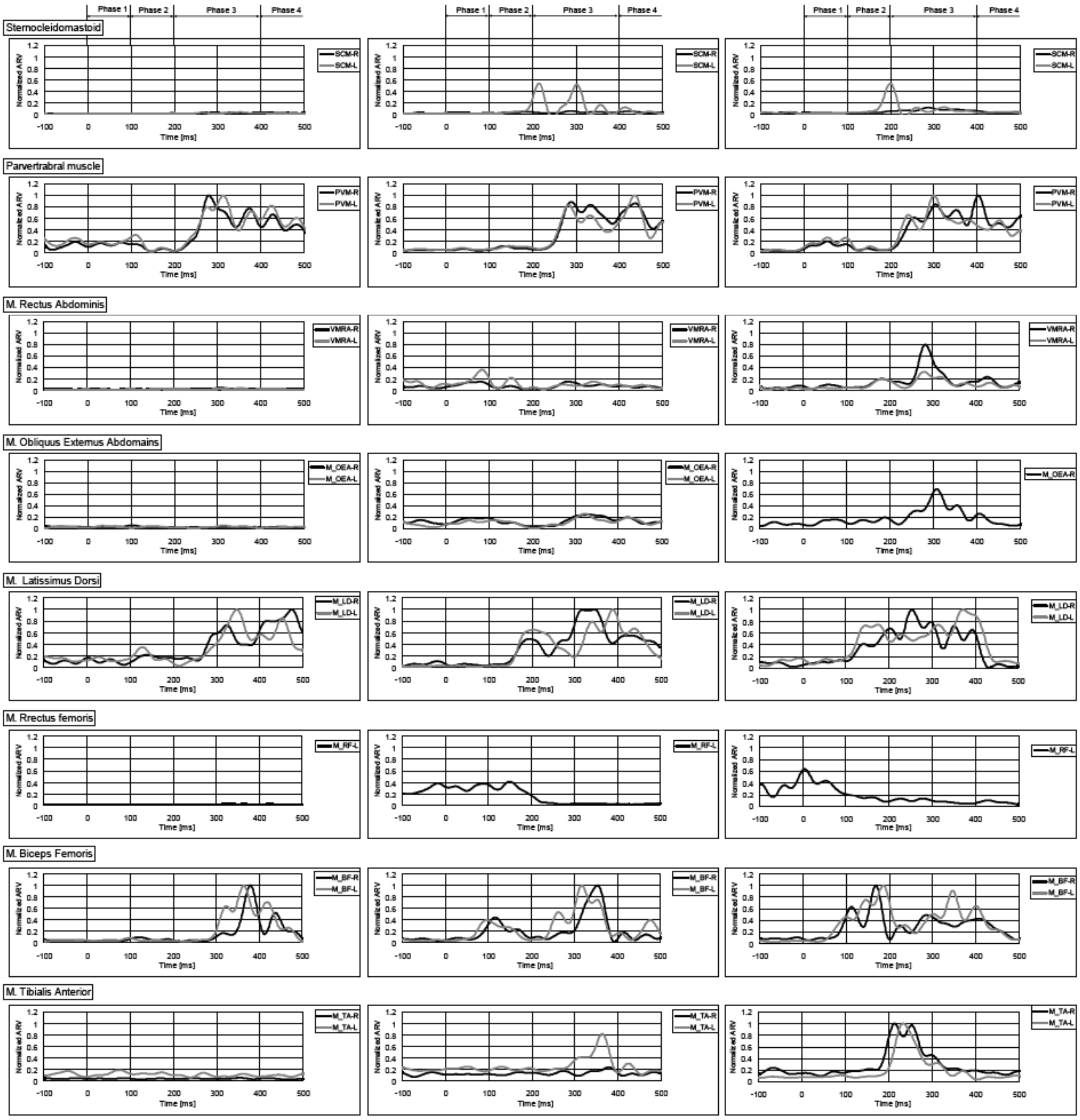


(b) 0.6 g



(c) 1.0 g

Figure 15. Trajectory of the landmark respect to the hip joint (Relax)



(a) 0.2 g (b) 0.6 g (c) 1.0 g

Figure 16. Time history of the muscle activation of each muscle (Relax)

LIMITATION OF THIS STUDY AND SUGGESTIONS FOR FURTHER RESEARCH: The number of the subjects in this study was limited to 3 male and 2 female human volunteers. Therefore, the test data were insufficient to conclude that the results can be reproduced in other studies. In addition, the subjects were not pretending the braking or the lean back motion when the sled bumped against the dumper. Therefore, the posture presented in this experiment can be rather thought as the posture of a passenger. Furthermore, the aim of this study is to establish an injury prediction approach to verify the influence of posture change on the occupant injuries in a traffic accident. In order to verify the relationship between the occupant posture and injury, the picture data recorded by EDR (Event Data Recorder) are needed. By using EDR data, the posture changes just before collision are evaluated from the video data in the real accident. Quantification of the input (crash conditions, posture and physical parameters) and the output (injury level and location) can clarify their causal relationship for occupants in real-life crash situations.

CONCLUSION

As a result of this study, the effects of muscular tension on the head, neck and torso motions have been clarified. The physical motion under the impact acceleration at low-level was classified into 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, back muscle such as M. latissimus dorsi and paravertebral muscles were mainly activated when the passenger made a pre-braking action. Depending on the location of each muscle, the range of reflex time of head, neck, and torso muscles was around 70 ms – 200 ms. These parameters are important factors in discussing the subject's motion just before the collision.

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