Occupant Kinematic Behavior and Effects of a Motorized Seatbelt on Occupant Restraint of Human Volunteers during Low Speed Frontal Impact: Mini-sled Tests with Mass Production Car Seat

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Abstract The objective of this study is to evaluate occupant posture change during pre-impact braking and explain the effects of a motorized seatbelt (MSB) on occupant restraint. In order to simulate the pre-impact condition, low-speed sled tests on young adult male volunteers were conducted with a vehicle seat, a seatbelt and a foot rest. In this study, two seatbelt systems and two muscle tone states were provided as test conditions. In order to evaluate the potential benefits of the MSB for posture maintenance, the volunteers were restrained by either a current three point seatbelt (SB) or the MSB. In order to examine the effect of muscle exertion on the physical motion, the experiments were conducted in two states: a relaxed state and a tensed state. Compared with the relaxed case, the upper torso motion was constrained by the muscle exertion in the tensed case. In addition, in both relaxed and tensed cases, the MSB worked effectively in restraining the upper body of the volunteers at the accelerating phase. The experiments and the results will provide useful information on occupant kinematics in an emergency situation, which is considered difficult to be measured in the vehicle cabin with high accuracy.

Keywords low-speed sled test, motorized seatbelt, muscle exertion, occupant kinematics, pre-crash.

I. INTRODUCTION

Changes in driver's posture and velocity during emergency maneuvers exert influence on the injury risks in frontal impact collisions [1]. In addition, the effects of an occupant's posture, muscle bracing and evasive maneuvers at the pre-crash phase on injury outcomes have been studied through computer simulation and crash tests [2-5]. These results show that posture change and muscle exertion affect injury outcomes in the chest and neck, in particular.

The behavior and posture change at the pre-crash phase have been investigated in order to improve the efficiency of occupant protection systems. Studies with a driving simulator showed that drivers brace rearward into the seat and straighten their arms against the steering wheel, or turn the steering wheel to swerve around the obstacles during emergency situations [6-7]. Volunteer tests with a sled system have also been conducted by several researchers. Low-impact frontal and side sled tests simulating acceleration at the pre-crash phase were conducted by Ejima et al. [8-10], and whole body kinematics and muscle activation were evaluated in terms of various accelerations, vehicle equipment and volunteer's gender. Similar investigations on low-speed frontal and side impacts were conducted to obtain input data for numerical simulation [11]. Beeman et al. [12] conducted low-speed frontal sled tests with human volunteers, the Hybrid III dummy and post mortem human surrogates (PMHS) and reported that forward displacements on body regions of a relaxed human volunteer are fairly similar to those of PMHS and significantly larger than those of the Hybrid III dummy. In combined volunteer tests and numerical study, Choi et al. [13] developed a human computer model with activating muscle, which was validated by volunteer tests, and demonstrated the relation between reaction force of the steering wheel, brake pedal and seat back, and muscle forces of upper and lower extremities when occupants brace in anticipation of a crash.

Furthermore, in recent years, volunteer tests with a real moving car on a road have been conducted by

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several researchers to examine occupant kinematics and/or evasive maneuvers in real-world situations. Behr et al. [14] reported that volunteers activate lower extremity muscles in order to operate a brake pedal and maintain driving posture when the volunteers drive a real car on a test track and are suddenly faced with an obstacle forcing them to brake suddenly. Carlsson and Davidsson [15] conducted volunteer tests to evaluate the driver and passenger kinematics during medium harsh braking with a real vehicle equipped with the system to avoid or mitigate low speed rear-end collisions by automatically applying the brakes. Mages, Seyffert and Class [16] evaluated the effects of reactive reversible pretensioning on a passenger's kinematics when an AEB worked and reported that the pretension can reduce head displacement from an average 232 mm without the system to an average 143 mm. Their study showed that occupant stature and gender affect forward motion during pre-impact braking. These tests with a real vehicle enable us to obtain beneficial information for development of active and passive safety systems because the data obtained by these tests represent the most realistic occupant behavior. However, it is difficult to measure three-dimensional whole body motion with high accuracy because of visual obstruction and less space for equipping enough cameras. Therefore, low-impact tests with a stationary sled system under more realistic conditions are required in order to study occupant kinematics at the pre-impact braking phase and add to existing knowledge for the improvement of restraint systems.

The objective of this study is to evaluate occupant posture change during pre-impact braking and explain effects of a motorized seatbelt (MSB) on occupant restraint. In order to simulate the actual pre-crash condition that occurs when the Automatic Emergency Braking system is activated, low-speed sled tests with human volunteers on a real vehicle seat were conducted.

II. METHODS

Two methods were used in this study: (1) to test the effect of muscle activity on the physical motion during low-level impact and (2) to compare the potential benefits of a MSB with a SB in terms of keeping in-position posture. Two low-speed sled tests were conducted with a current 3-point seatbelt (SB) and another two tests with a MSB. In both test groups, the experiments to evaluate the effects of muscle activity were conducted under two muscle states: a relaxed state, in which the volunteers were subjected to the impact in the state of relaxed muscles, and a tensed state, in which volunteers intentionally tensed their muscles in order to brace in anticipation of acceleration.

Volunteers and informed consent

Three adult male volunteers in good health participated in the series of experiments. The average and standard deviation of the volunteers' anthropometric data are shown in Table 1. The protocol of the experiments was reviewed and approved by the Tsukuba University Ethics Committee, and all the volunteers submitted their informed consent in the document in compliance with the Helsinki Declaration.

TABLE I
VOLUNTEER DATA

	Height (cm)	Weight (kg)
Mean	170.3	68.6
SD	2.5	5.9

Sled Apparatus for the Low-Speed Impact

This study employed a low-speed sled system. The sled system can generate acceleration that simulates vehicle braking when the AEB system activates in an emergency situation and is equipped with a vehicle seat, a seatbelt system and a foot plate. In order to capture spine motion, a part of the seat back was removed and non-stretch tapes were fixed over the hollow as the substitute for the seat back. Fig. 1 shows the scheme of the sled system.

The acceleration pulse has a half-sine shape of 8.0 m/s^2 maximum acceleration and a rise time of 0.5 s shown in Fig. 2. In this study, the kinematics of the volunteers was measured at the rising phase of the acceleration pulse.

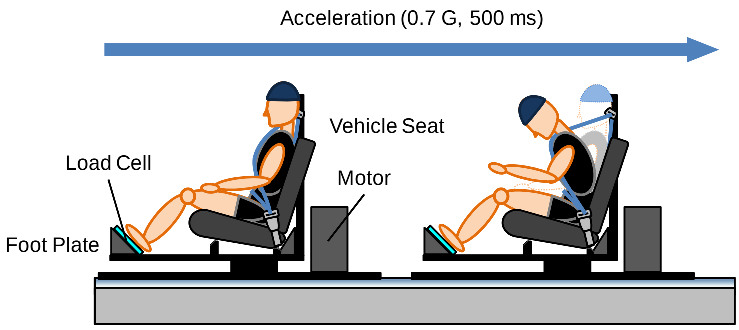


Fig. 1. Scheme of the low-speed sled system.

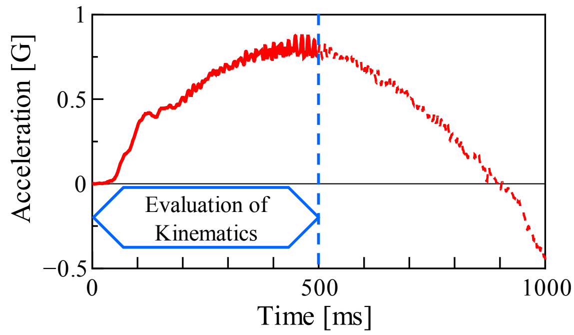


Fig. 2. Acceleration pulse generated by the sled system.

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

The physical motion of the human body and head-neck-torso kinematics at low-level impact accelerations were measured using a three-dimensional motion capturing system (The Raptor-E Series, Nac Image Technology Inc. / Motion Analysis Corporation). The feature of this capturing system is that the position of each marker is extracted automatically from a video image caught by several cameras and is translated into three-dimensional coordinates. The resolution of the camera is 1280 x 1024 pixels and the sampling rate is 500 fps.

The images were incorporated into CORTEX software (Nac Image Technology / Motion Analysis Corporation) and analyzed. In this study, the motion of the markers on the head, T1, T12, L3 and left and right shoulder shown in Fig. 3 were evaluated. The maximum measuring error estimated from the 3D motion capturing system by using body surface landmarks was around 1.5 mm in this setting.

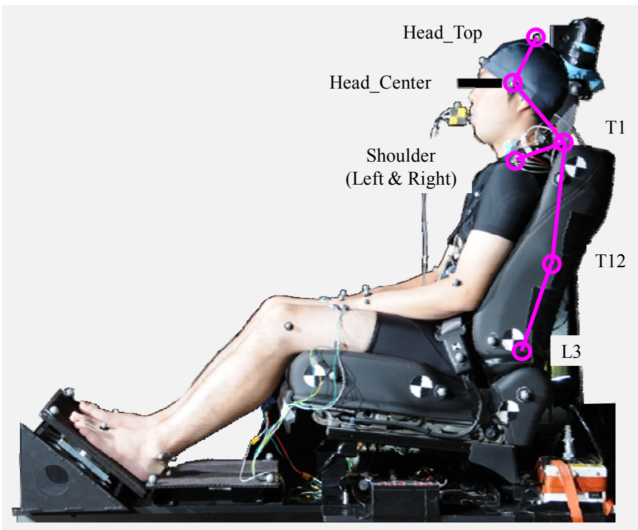
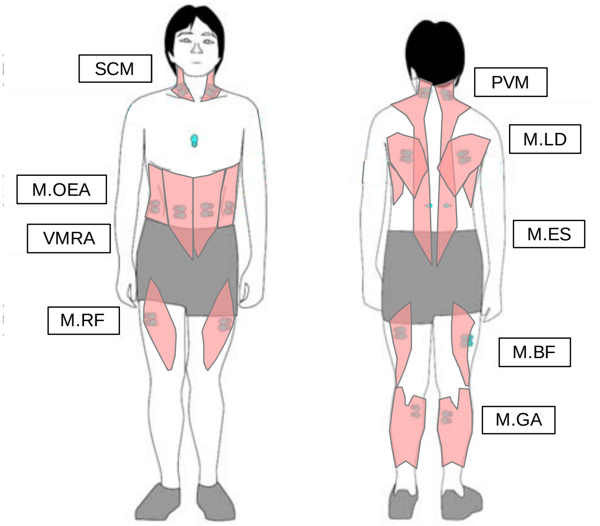


Fig. 3. Sitting posture of the volunteer and kinematics evaluation points of body surface.

Electromyography

Muscle activity was measured by means of surface electromyogram (EMG). The timing of EMG data was synchronized with the three-dimensional kinematics data. EMG electrodes were attached onto the skin of the major muscles of the volunteer. The electrodes with diameter of 5 mm were arranged as bipolar electrodes with a distance of roughly 20 mm between the electrode centers. Fig. 4 shows the locations of the surface electrodes.

The muscle activation measured during the impact was analyzed by systematic processing [8][10]. First, the raw EMG signal was filtered through 10-350 Hz band pass filter. Next, full wave rectification was performed on the filtered signal. Finally, the average rectified value (ARV) was obtained by smoothing with a 10 Hz low pass filter. Each muscle response was normalized by its own ARV magnitude of maximum voluntary isometric contraction which was measured before a series of sled tests.



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

Fig. 4. Location of the electromyography.

Experimental Condition

This study consists of a series of four low-speed sled tests composed of a combination of two parameters: the effects of muscle exertion on keeping posture and the in-position function of an MSB in comparison with a SB. Both belt systems are furnished with an emergency locking retractor. In addition, the MSB has a motorized retractor, which automatically tightens the belts when the vehicle's pre-collision sensing system determines an imminent collision. The electric motorized retractor generated approximately 120N pretensioning shoulder belt load. The time span to reach the peak load was 120ms in the static condition.

In order to examine the effect of muscle activity on the physical motion in the pre-impact braking situation, the experiments were conducted under two conditions: a relaxed state, in which the volunteers were subjected to the impact in the state of relaxed muscles, and a tensed state, in which volunteers intentionally tensed their muscles in order to brace in anticipation of acceleration. The volunteers were instructed so that they could assume each of these muscle configurations. In the relaxed case, the volunteers were required to be fully relaxed until the body motion was stopped by the seatbelt. On the other hand, in the muscle-tensed cases, the volunteers were instructed to tense all their muscles intentionally.

When the acceleration to the sled was applied while the volunteers were assuming the same initial posture, the differences due to the restraint system and the muscle activation could be clearly observed in the motion of their upper torso.

Table 2 below shows the test matrix from this study.

TABLE 2
TEST MATRIX

Test No.	Seatbelt	Muscle activity
1	SB	Tense
2	SB	Relax
3	MSB	Tense
4	MSB	Relax

III. RESULTS

Kinematics of whole body

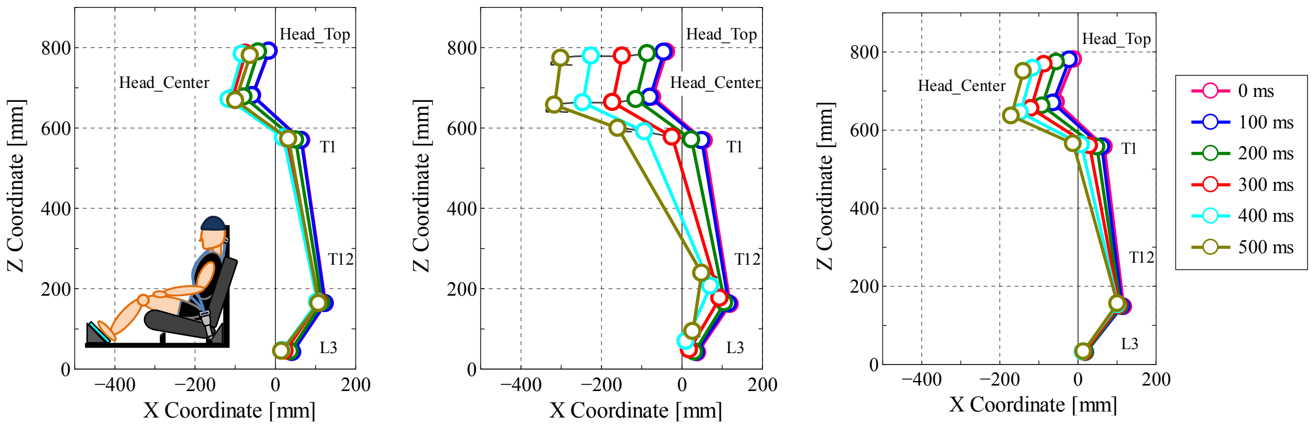
The motions of the markers on the head, T1, T12 and L3 in the side view are shown in Fig. 5, and those of the left and right shoulders in addition to overhead markers in the top view are shown in Fig. 6 at 100ms intervals up to 500ms. The differences caused by initial muscle configuration and restraint system was the focus of this study. Thus, the kinematics are discussed of the three conditions (#1, #2, #4) where the difference can be observed as being significant; the case of the smallest motion (#3) was omitted in Fig. 5 and Fig. 6.

Note that the "Head_Center" marker was virtually located on the mid-point of markers at both lateral sides of the head. This definition is different from that of head center of gravity (COG) employed by Ono et al. [17] and Ejima et al. [8], which was based on the Frankfurt line. Although the rotational motion of the head regarding three orthogonal axes passing through its COG could not be evaluated, this difference between these two points was an acceptable approximation because the point of "Head_Center" in this study was very close to that of the head COG. In addition, the direction of the left-right axis of the two-dimensional projection was almost coincident with that passing through its COG, which is the principal rotation axis in a frontal impact.

As shown in Fig. 5, larger forward movement distances of the upper body at the acceleration phase in the relaxed case were measured in comparison with the tensed case and the MSB case. Average of maximum Head_Center and T1 displacement of the three volunteers in the longitudinal direction was 255 mm, 203 mm in the relaxed state with the SB, 56 mm, 33 mm in the tensed state with the SB and 125 mm, 64 mm in the relaxed state with the MSB, respectively. While forward motion was dominant for the head and T1, the markers of T12 and L3 moved upward in the relaxed state with the SB. Despite the same seatbelt system, the forward movement in the tensed state was small for all marker points.

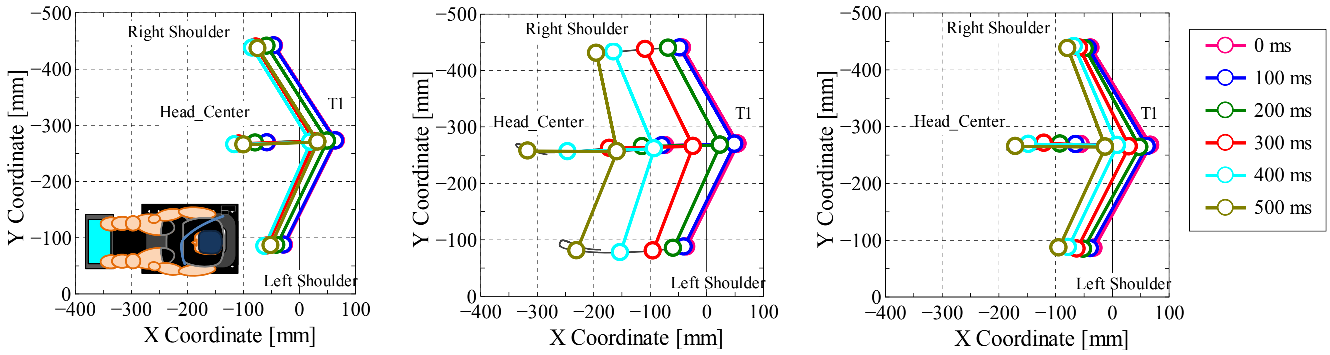
In addition, it seems that the MSB worked effectively in reducing forward displacement by tightening the belt in advance and restraining the upper body. It should be noted that the T12 marker in the MSB case scarcely moved compared with the upward movement in the SB case, and the head in the MSB case moved obliquely downward whereas the head with the SB moved forward.

As shown in Fig. 6, the kinematics in the top view show that all markers moved almost straight under all conditions. It can be observed that the displacement of the left shoulder was larger than the displacement of the right shoulder acted on by the shoulder belt force in the relaxed state with both the SB and the MSB.



#1 SB_Tense #2 SB_Relax #4 MSB_Relax

Fig. 5 Physical motions with 2D projection of the head, neck, spine in the side view.



#1 SB_Tense #2 SB_Relax #4 MSB_Relax

Fig. 6 Physical motions with 2D projection of the head, neck, spine in the top view.

In order to evaluate changes in posture with time, time histories of forward displacement of the head and T1 were compared. Fig. 7 shows the average time history of the forward displacement during the impact. These curves represent the average time history with three volunteers under all conditions, and the shadings represent the corridor (±1 S.D.).

In general, rise timing and peak timing in the time history of the head were almost identical to those of the T1 under each test condition. In the relaxed-state cases with both restraint systems, similar time histories of displacement of the head and T1 were observed up to 100 ms, and the head and T1 reached the peaks of the forward displacement at around 350 ms and began to move backward after that. In the tensed case with the SB, the peaks of forward displacement of the head and T1 were shown at about 150 ms in Fig. 7. On the other hand, the 150 ms peaks could not be observed in the MSB case.

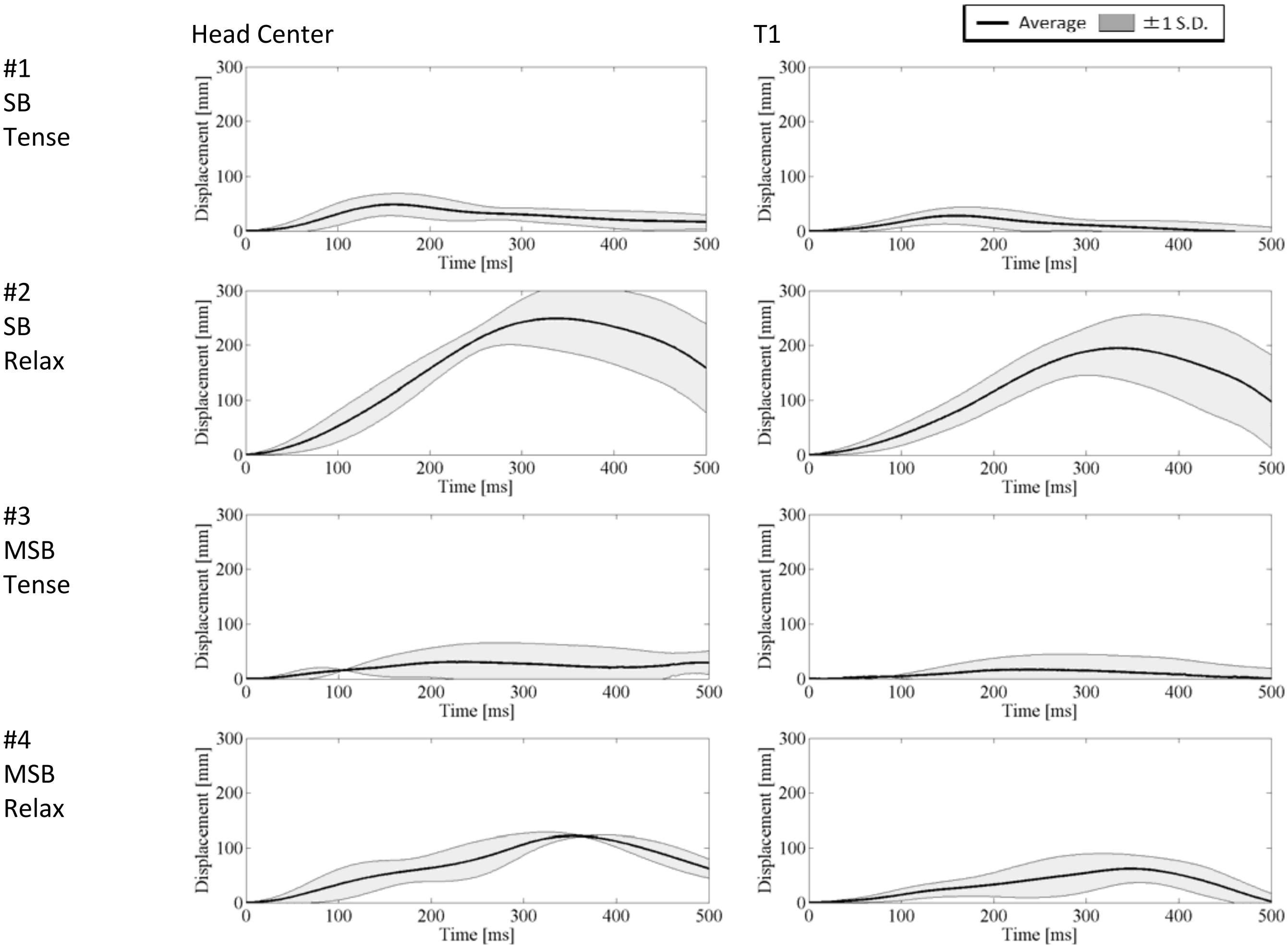
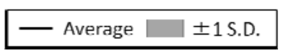


Fig. 7 Forward displacement of head and T1 markers in each test condition.

Loading condition on the volunteer

Fig. 8 shows the comparison of the shoulder belt force measured at the upper right hand side of the volunteer. The belt force increased rapidly after 300 ms in the relaxed state with the SB (#2), which is the case that the largest forward displacement was observed in all tests. This is because the sled pulse caused the retractor with the ELR function to lock instantly, and the belt secured the volunteer's upper body. In the tensed state with the SB (#1), little belt force was generated during acceleration. While the time history curves shows a plateau at approximately 150 N in the tensed state with the MSB (#3), the curves in the relaxed state (#4) show gradual rise. These tendencies might be derived from the different magnitude of contribution of the belt force for posture keeping between two muscle states. Accordingly, these results suggest that muscle exertion in anticipation of sled acceleration also affected the belt forces.

Fig. 9 shows the time history of average force measured by the load cell under the foot plate. These curves represent the average time history with the three volunteers and the shadings represent the corridor (±1 S.D.).



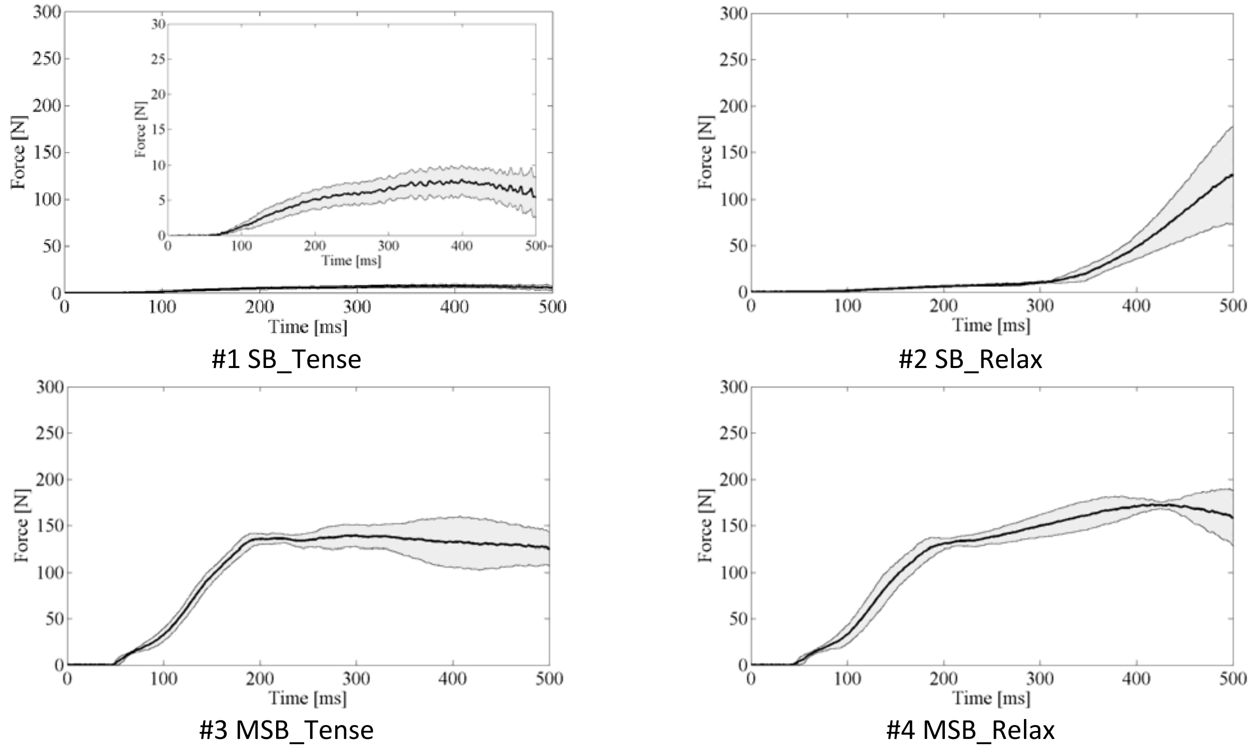


Fig. 8 Shoulder belt force in each test condition.

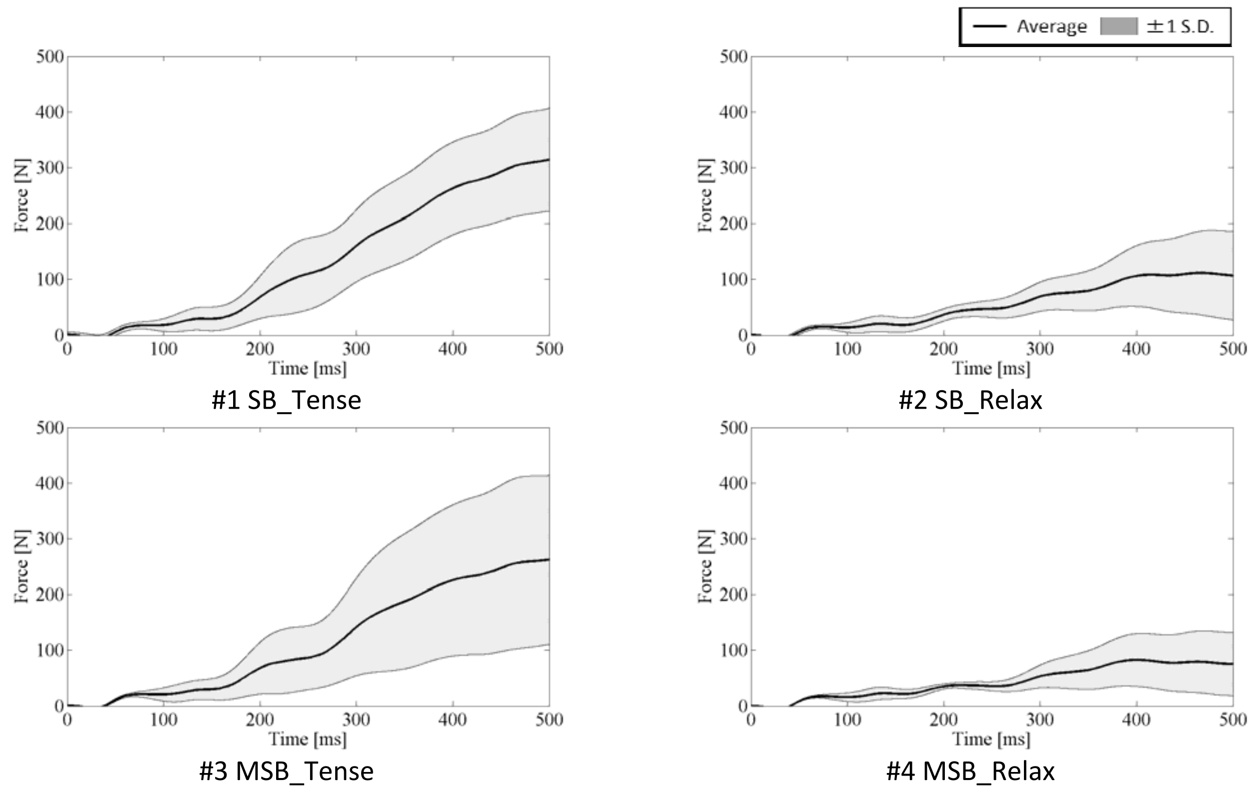


Fig. 9 Foot plate force in each test condition.

The foot forces in the tensed-state conditions were larger than those in the relaxed state with both seatbelt systems. It can be observed that the foot force with the MSB was similar to that with the SB, which was common between tensed and relaxed states.

EMG condition on the volunteer

In this section, focus is on the relationship between the forward movement and muscle activation. Thus, the results in relaxed and tensed states with the SB, which are very different in the forward movement are compared.

Fig. 10 shows the average time history of the normalized ARV in order to evaluate each muscle activation during the impact. These curves represent the average time history with three volunteers under two conditions (#1, #2), and the shadings represent the corridor (±1 S.D.).

There were distinct differences in PVM, M_RF, M_BF, and M_GA. In regard to M_ES, the rising timing of curve of the tensed state was earlier than that of the relaxed state. The PVM and M_ES are the muscles located on the back side and support the neck and upper body, respectively. It seems that these muscle activations in the tensed state resulted in prevention of the head and torso from forward bending as shown in Fig. 5 and Fig. 7. The activations of M_RF and M_GA, which contribute to knee extension and foot plantar flexion, respectively, generated pressing behavior on the foot plate to maintain posture.

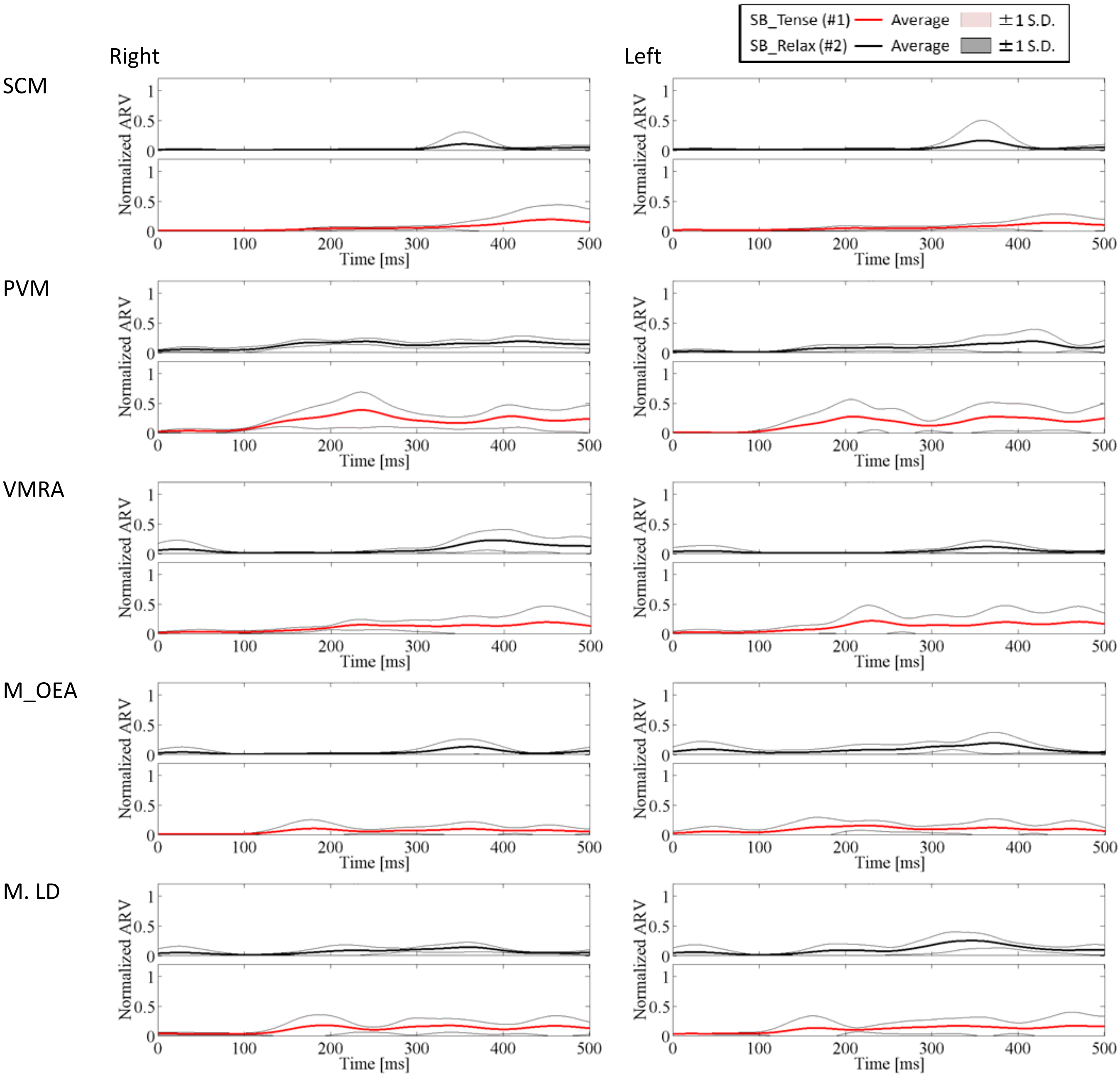


Fig. 10 Time history of EMG in each test condition.

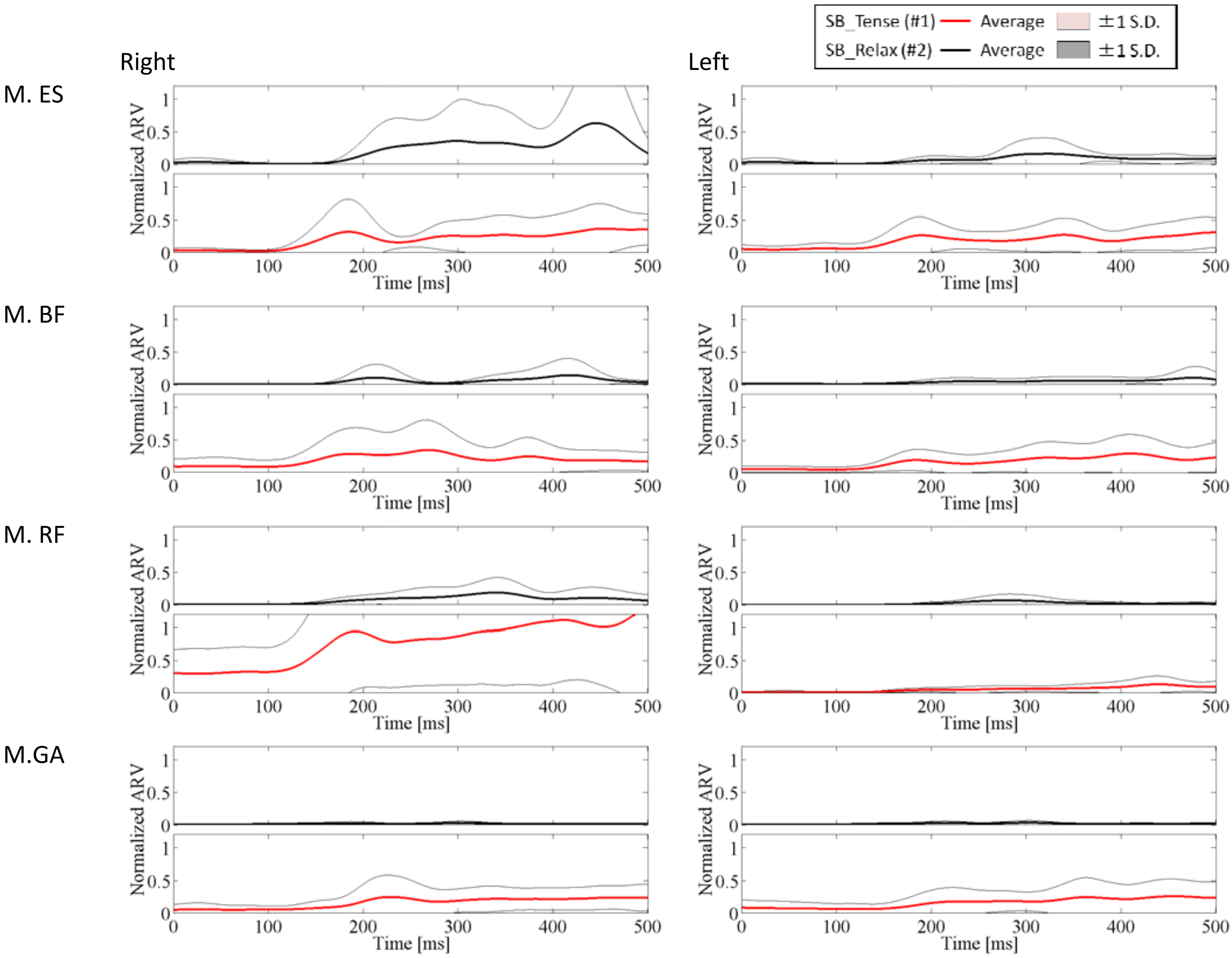


Fig. 10 Time history of EMG in each test condition (continued).

IV. DISCUSSION

In this study, it was confirmed that muscle exertion and the MSB contributed to keeping in-position posture during a low-speed frontal impact. In order to clarify the mechanism, the relationship between kinematics of upper body, external forces and EMG are discussed as follows.

Muscle exertion produced effects on the posture maintenance despite the SB condition. The forward movement in the tensed-state case with the SB (#1) was smaller than that in the relaxed-state case (#2) shown in Fig. 5 and Fig. 7. This result indicates that smaller displacement must be due to its own voluntary muscle exertion. As can be seen in Fig.8, little belt force was measured in the tensed state with the SB. This low level of force indicates that the volunteers did not rely on the seatbelt and their own muscle tension made a large contribution to keeping in-position posture. In addition, in the tensed state, the increase of foot plate force shown in Fig. 9, the exertion of the muscles for posture maintenance such as M. Paravertebralis (PVM) and M. Erector Spinae (M_ES) and pressing the foot plate such as M. Rectus femoris (M_RF) and M. Gastrocnemius (M_GA) shown in Fig. 10 were observed. As a result, it can be said that adult male occupants can maintain their upper body posture by their muscle exertion under low acceleration simulating pre-impact braking when they brace in anticipation of emergency situations.

On the other hand, larger forward displacement and larger belt force were observed under the relaxed state with the SB (#2). From this result, we concluded that the upper body in the relaxed state was mainly restrained by the belt webbing. The maximum belt force in the relaxed state with the SB (#2) was approximately 150 N at 500 ms, which is close to the inertial force of the thorax and abdominal parts that was estimated from the product of the mass of these parts (36% of weight of whole body, referred to as ratio of body segmental mass [18]) and the maximum acceleration of the sled. Furthermore, at 300 ms, the seatbelt force began to increase rapidly due to retractor lock shown in Fig.8, and the head and T1 reached the maximum forward displacement and moved backward after that (Fig. 7). Therefore, the belt force exerted strong influence on the kinematics of the upper body under this condition (#2).

The MSB dramatically reduced the forward displacement after 100 ms, particularly in the relaxed state. Comparing the results in the relaxed state, the MSB reduced the head displacement by 51% and the T1 displacement by 68%. In addition, the marker on T12 did not move in the MSB case while it moved obliquely upward in the SB case after 200 ms, which was close to the time of belt force generation by the MSB. Unlike the tensed state (#1) and the relaxed state with the SB (#2), the head moved obliquely downward in the relaxed state with the MSB (#4). This was because the belt force acted on only the torso and lumbar parts, and the head which was not subjected to the belt force revolved round the neck. Consequently, the MSB can reduce the forward displacement of the upper body when AEB works in an unexpected situation. Therefore, it can be expected that the MSB leads to the prevention of out-of-position during pre-impact braking and the optimized interaction with the airbag after vehicle crash.

In this study, the forward displacements of human volunteers on a vehicle seat were evaluated in low-speed frontal sled tests simulating an actual pre-crash condition. When comparing the average relaxed volunteer response on the sled with the average response on a real moving car on road [16], no significant differences were observed for head displacement: head displacement in the relaxed state with both the SB (a conventional SB in the literature) and the MSB (reversible pretensioning in the literature) was 255 mm/125 mm in our tests in comparison with 232 mm/143 mm in the literature, respectively. This consistency suggests that our sled tests can represent an actual situation when AEB works. Furthermore, the sled tests enable us to measure three-dimensional kinematics of the whole body, which is difficult to measure in a real moving car due to visual obstruction. In our tests, it can be observed that displacement of the left shoulder was larger than that of the right shoulder acted on by the belt force in the relaxed state with both the SB and the MSB. Therefore, our study provides beneficial data to complement the recognition of occupant kinematics at the pre-impact braking phase and to improve an integrated safety device.

It can also be expected that the results regarding kinematics during low-speed frontal impact contribute to the development of a human computer model taking into account muscle activation. For this development, in addition to referring to the kinematics data directly, joint force and muscle force during pre-impact braking should be estimated by using an inverse dynamics method and reflected in model validation.

In the tests, volunteers braced after the sled start was announced to them. However, it is thought that the cognition and decision of the volunteers in a preannounced situation on the sled may be different from those in an emergency situation faced on a real road. Therefore, it is possible that the differences in a test condition also affect muscle bracing. Future research is necessary to investigate the difference of occupant behavior and bio-signal such as electrocardiogram, heart rate and so on, in addition to EMG between a sled and a real vehicle scenario.

V. CONCLUSIONS

In this study, we have conducted a series of sled tests and evaluated the relationships among body kinematics, restraint force, muscle activation and the effect of a motorized seatbelt on keeping in-position posture during low-speed frontal impact simulating pre-impact braking. The test results have shown that muscle exertion influenced the posture maintenance in both restraint system conditions. The differences between muscle configurations were observed in seatbelt force and EMG of the muscles for posture maintenance. In addition, the MSB dramatically reduced not only the forward displacements of the head and T1 but also those of T12 and L3, particularly in the relaxed state. It is expected that these results will contribute to the development of integrated safety devices and computer human models with active muscle in order to improve safety performance in vehicle crashes.

VI. ACKNOWLEDGEMENT

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