BIOMECHANICAL RESPONSES OF HEAD/NECK/TORSO TO LATERAL IMPACT LOADING ON SHOULDERS OF MALE AND FEMALE VOLUNTEERS

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ABSTRACT

For a better understanding of neck injury mechanism in lateral-collisions, experiments and analyses have been conducted on human head/neck/torso impact responses and cervical vertebral motions upon loading of lateral impacts from the shoulders of human volunteers. At the same time, differences in neck muscle responses between the male and female volunteers have been also analyzed by means of experiments. Likewise, the differences between human volunteers and cadavers have been also studied and discussed.

Key Words : Injury Criteria, Muscle, Neck, Volunteers, Lateral Impact

VEHICLE OCCUPANTS INVOLVED IN AUTOMOBILE accidents but saved from fatality with the injury severity level reduced to serious - minor are increasing, owing probably to the implementation of automobile safety measures and advances made in emergency medical treatments. It can be deduced that the increase in number of those with severe - minor injuries is attributable to the above-mentioned tendency. In order to cope with this tendency, active studies are being made for further enhancement of automobile safety - particularly against vehicle frontal collisions. Despite such efforts, however, the number of those injured by rear-end collisions is increasing significantly (Kraft et al., 2002), which is considered by some researchers as a "trade-off" between the number of fatalities and the number of "severe - minor injuries", with the priority set on the reduction of the fatalities. As regards the neck injuries, the tendency of such increases is found not only in rear-end collisions but also in lateral-collisions (Hell et al., 2003). The same as in the case of rear-end collisions, the neck injury mechanism in lateral-collisions has not been clearly determined, with many questions still remaining unsolved (Kumar et al., 2005, Ito et al., 2004, Yoganandan et al., 2001). One of the reasons is the scarcity of biomechanical studies conducted on human head/neck/torso impact responses in lateral-collisions.

In this regard, a new test equipment called "head/neck inertia impactor" was used in this study in order to analyze the "human head/neck junction" while applying a lateral impact to the shoulder. To be more specific, volunteers were impacted on their shoulders to simulate automobile lateral-collisions, and study human head/neck/torso impact responses as well as cervical vertebral motions. Differences in neck muscle responses between the male and female volunteers were also investigated. A decision was also made to ask INRETS, conducting the impact tests on cadavers with the similar test method as that on humans, to provide the authors with some test results, in order to determine differences in impact responses between the human volunteers and the cadavers.

EXPERIMENTAL METHODS

LATERAL INERTIA IMPACTOR: An inertia impactor (Figure 1) specially designed for this study was used, in order to investigate head/neck/torso responses and cervical vertebral motions of subjects submitted to a lateral inertia impact. The test equipment consists of a compressed air storage/coil spring unit to eject the impactor, the impactor height adjuster, and the test subject sitting position adjuster (forward/ backward & up/down). The front plate, pushed against the impactor front, was fixed to the piston through the piston rod. The compressed air is stored in the cylinder with the piston fixed to the air chuck

located at the rear end. The impactor mass is 8.5 kg. The impactor is ejected by opening the air chuck, and the impact is applied to the back of test subject. A coil spring is provided to control the impactor stroke and the rise of impact load. The stroke setting and the rise of impact load can be varied per test.

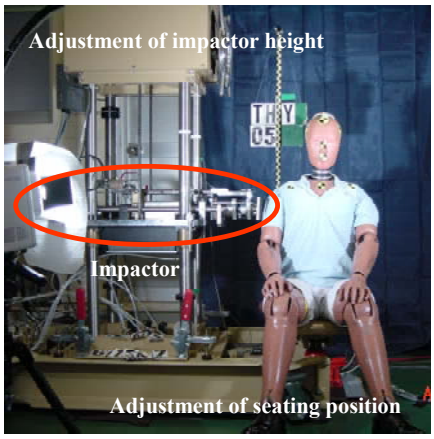

Fig.1 Lateral inertia impactor

HEAD/NECK/TORSO VISUAL MOTIONS: In order to record the kinematics of the head/neck/torso of each subject during impact, a high-speed video camera with the photographic capability of taking 500 frames/s was used. The head rotation angle and the displacement relative to the torso (the first thoracic vertebra: T1) were calculated by tracing the motion of each marker adhered to the subject according to the photographic images. A VICON motion photographic device (125 frames/s) was also used for the three-dimensional analysis of head/neck/torso motions.

CERVICAL VERTEBRAL MOTIONS USING CINERADIOGRAPHY SYSTEM: For the analysis of cervical vertebral motions of each subject during impact, a cineradiography system (Philips: BH500) was used. The system is capable of taking cervical vertebral images at the rate of 60 frames per second with 16.67 ms intervals.

EXPERIMENTTAL CONDITIONS: Using five healthy male and three healthy female adults as human volunteers, experiments on the head/neck/torso impact responses and the cervical vertebral motions upon lateral inertia impact have been conducted. Table 1 shows anthropometric data on human volunteers. The impact loading direction was set vertical (0 deg inclination) against the shoulder on one side (Figure 2). To be more specific, each test subject sat by one side of the impactor, with the back set practically straight against the stiff seat, so that the impact direction became parallel to the line connecting the acromion and the lower part of the cervical vertebrae. In order to analyze the differences in impact loading directions, the impact was also applied from 15 deg forward and 15 deg backward directions (Figure 2), in addition to the 0 deg direction. The impactor surface is rectangular with the area of 100 mm x 150 mm. The impact loading location against the subject's shoulder was so set that the position of acromion would become the same as that of impactor upper surface. In order to find the difference in effects of neck muscle response on the head/neck/torso motions, the states of muscle were set in tensed and relaxed conditions, respectively. The impact load was set at 3 different levels such as 400 N, 500 N and 600 N in order to find the differences in head/neck responses to the lateral impacts. For the direction with 0 deg inclination, the impact responses were compared between cadaver tests and those on the volunteers. Table 2 shows the different test conditions classified by differences in sex (male and female), impact loading levels, impact directions and states of muscle, with different combinations of test conditions.

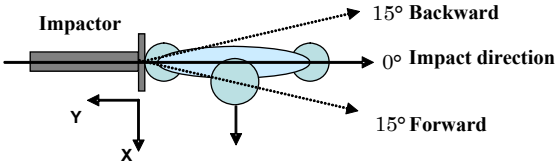

Fig.2 Impact directions

Table 1 Anthropometric data of the subjects

	Age	Sex	Height (cm)	Weight (kg)	Sitting Height (cm)	Mass of head (estimate) (kg)	Inertia of head (estimate) (10^{-2}kgm^2)
1	25	M	172	67	97	4.28	2.21
2	23	M	170	63	94	4.14	2.14
3	22	F	162	46	83	3.63	1.85
4	23	F	166	51	88	3.77	1.93
5	24	F	161	58	86	3.98	2.04
6	23	M	180	85	91	4.97	2.59
7	24	M	174	61	90	4.07	2.10
8	24	M	181	77	96	4.64	2.42

Table 2 Test conditions

No. of Subject	Sex	Impact force (N)	Impact direction	Muscle condition
8	Male Female	400	15° forward	Relaxed Tensed
		500	0 degree	
		600	15° backward	

INFORMED CONSENT FOR VOLUNTEERS: The informed consent procedure in line with the Helsinki Declaration (WHO/CIOMS, 1988) was conducted for the volunteers in order to

inform them fully of the purpose and method of experiments to ensure the full consent of each volunteer. The details/contents of the experiments were subjected to the approval of Special Committee of Ethics, Medical Department, Tsukuba University.

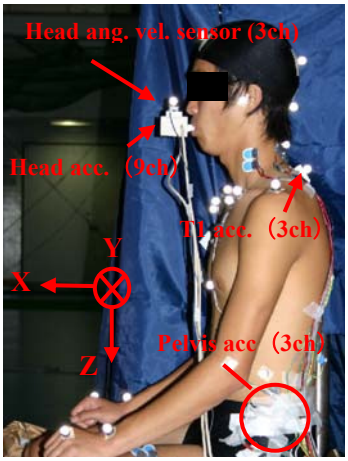

Fig.3 Mounting of accelerometers and rotational velocity sensors

ANLYTICAL METHODS

IMPACT FORCE APPLIED TO HEAD/NECK: The accelerations measured with the head 9 channel accelerometer, the acceleration measured at the first thoracic vertebra (T1), and the electromyogram during the experiments were measured, then the neck forces with the electromyogram was analyzed. The measuring instruments were the head 9ch accelerometer (X, Y & Z), head angular velocity sensor (X, Y & Z), T1 accelerometer (X, Y & Z) and the pelvis accelerometer. The locations where the sensors were attached are shown in Figure 3. A mouth-piece suitable for the tooth profile (teeth impression) was prepared for each test subject. Assuming that the head is a rigid body, the head coordinate system was set in line with the location of anatomical center of gravity. The 9 channel acceleration measurement method (Ono et al., 1980) was applied according to the coordinates of each accelerometer in this system, and the rotational and linear accelerations at the head CG were calculated.

TORSO ACCELERATION (T1): For the measurement of acceleration at T1, a three-axial accelerometer was attached onto the skin over a spinous process of T1.

THREE-DIMENSIONAL MOTIONS OF HEAD/NECK/TORSO: The three-dimensional motions of head/neck/torso were measured by means of VICON Motion Capture. Then the right-shoulder strain (displacement), left-shoulder strain (displacement), head rotation angles (X, Y and Z), T1 rotation angles (X, Y & Z) and the head rotation angles relative to T1 were analyzed.

RESULTS

CHARACTERISTIC ASPECTS OF IMPACT RESPONSES & VISUAL MOTIONS: An example of 600 N impact loading experiments (in relaxed muscle condition) is shown in Figure 4, with the sequential photographs of the head/neck/torso motions during impact. X-ray of the neck motions under the same test conditions are shown in Figure 5. Figure 6 shows the corridors of the impact forces measured in 600 N impact loading experiments (in relaxed muscle condition), the accelerations at the head CG (X, Y & Z) calculated from the values measured with the head 9 channel accelerometer, the accelerations (X, Y & Z) at the T1, the pelvis accelerations (X, Y & Z), and neck forces (Fx, Fy, Fz, Mx, My & Mz). The Figure 7,

Fig.4 Sequential motions of head/neck/torso（Impact force：600N, Relaxed condition）

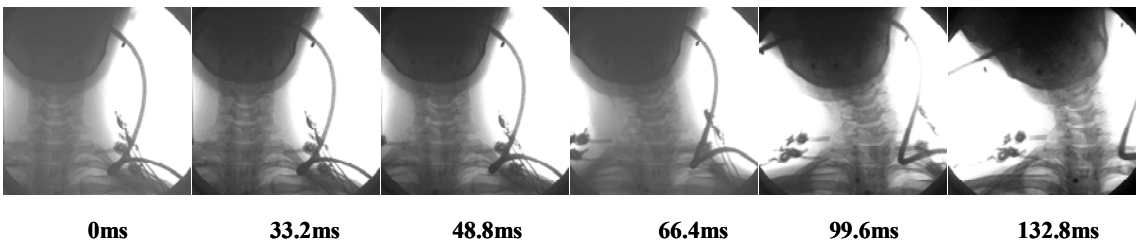

Fig.5 Sequential images of cervical vertebrae by cineradiograpy (Impact force: 600N, Relaxed condition)

on the other hand, shows the visual motions in relation to the shoulder displacements found from shoulder displacements and the strains (at the sternum upper end and the right or the left acromion) of right shoulder (right acromion) and left shoulder (left acromion), head rotational angles (X, Y & Z), T1 rotational angles and their corridors.

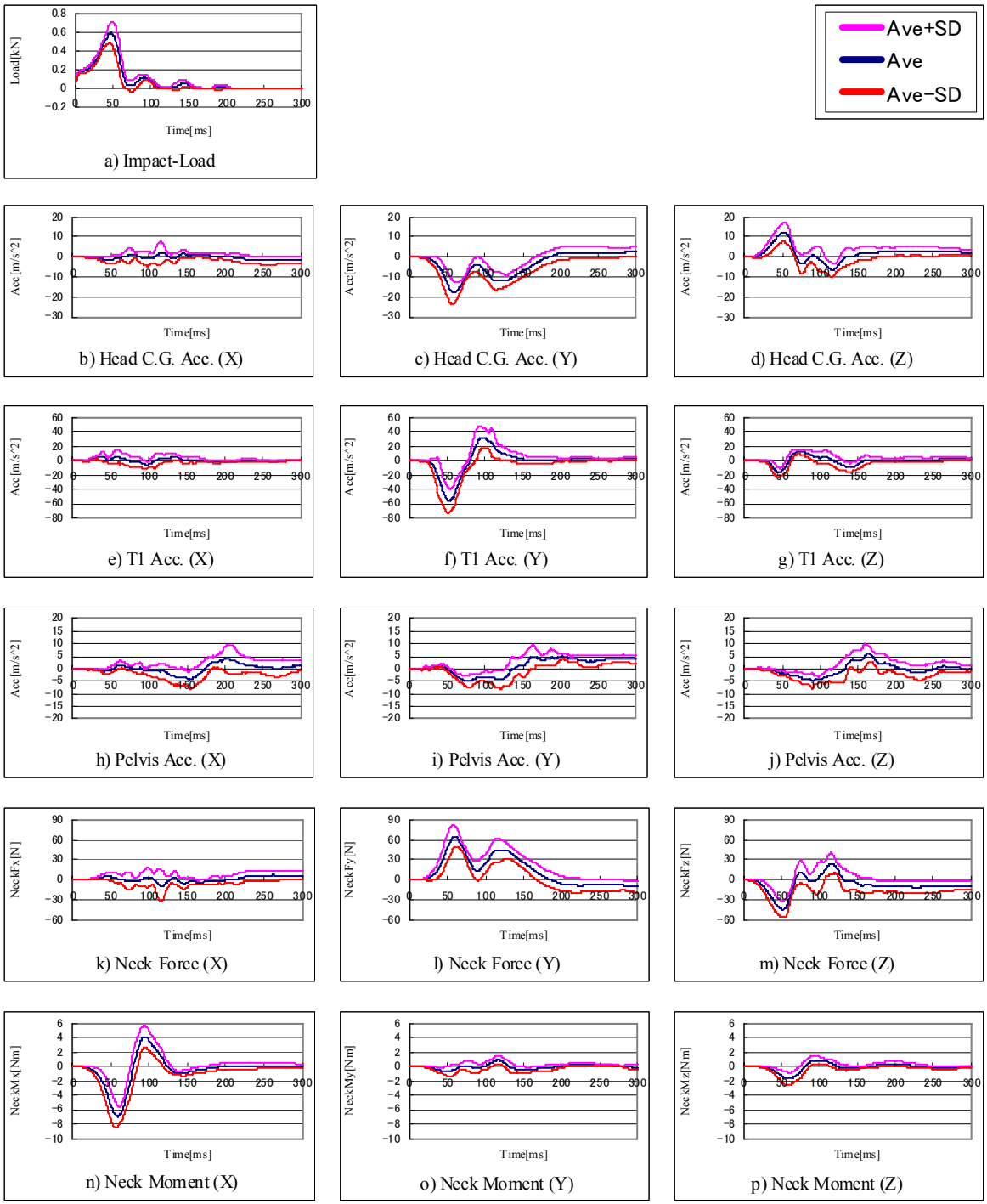

Fig. 6 Head/neck/torso responses for 600N impact under the relaxed muscle condition (8 subjects)

Phase 1 [0-50 ms]: The duration of impact for each one of 8 test subjects was 70 ms or so (Figure 6a)). The impact load peak levels are observed to be fluctuating, as the impactor and the shoulder are not contacted tightly enough in the initial stage of impact. This presumably resulted in the relatively low impact peak level in the initial stage and the relatively high peak level in the secondary stage. The T1 accelerations, on the other hand, show that the maximum value is found around 50 ms, while that of the head is found around

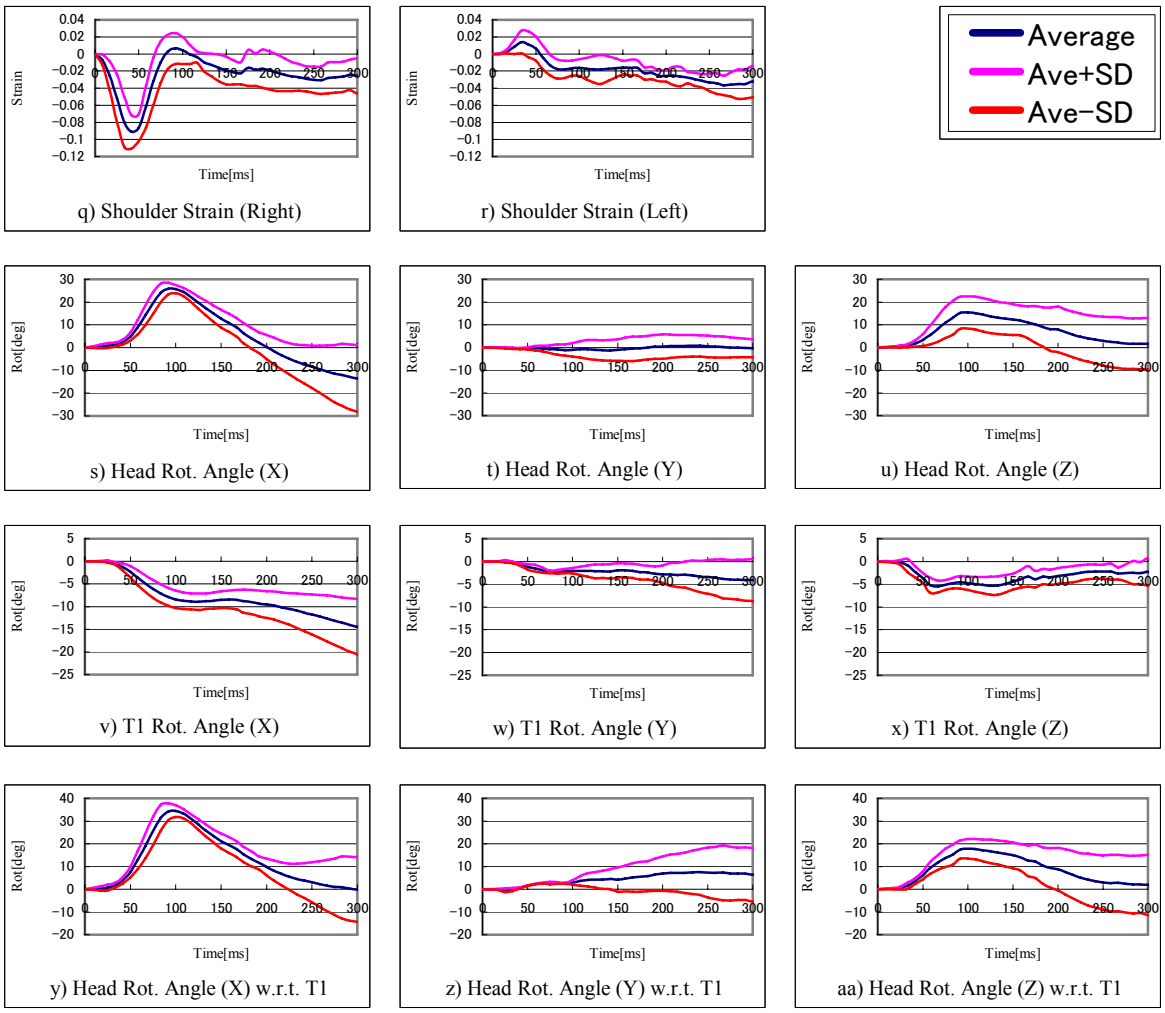

Fig. 7 Head/neck/torso motions for 600N impact under the relaxed muscle condition (8 subjects)

60 ms (Figures 6b)-6d)). The maximum values of T1 and the head in the Y-axial direction are 55 m/s^2 and 18 m/s^2, respectively. It is deduced that the axial forces between the T1 and the head acting to each other in opposite direction of compression, as the accelerations of T1 and the head in the Z-axial direction are reversed around the 50 ms. The rotations of the head and T1 around the X-axis are reversed around 30 ms. The rotations around the Z-axis are also reversed. The neck shear force (in Y-axial direction) and the neck moments around X-axis and Z-axis do not show their maximum values around 50 ms (Figures 6l), 6n), 6p)), but the axial force of neck in Z-axis shows the maximum value at 50 ms or so. The right shoulder strain (on the impacted side) shows the maximum value around 40 ms (Figure 7q)). A slight torsion of upper cervical vertebrae is found around the Z-axis (Figure 5).

Phase 2 [50-100 ms]: The impact was continually set up to 70 ms or so (Figure 6a)), and the shoulder was separated from the impactor due to the torso inertia. Hence, the acceleration at each portion of the head drops thereafter (Figures 6b)-6d)). However, the head rotates laterally against the torso, and the acceleration in the Y-axial direction starts to increase around 90 ms, as the head is subjected to a restriction by the lateral bending at the same time. The head rotation angles found from the three-dimensional motion analysis by means of VICON Motion Capture show the maximum values around 100 ms in both X and Z axial directions (Figures 7y)-7aa)). The timing is roughly the same as the timing when the head rotational angle relative to T1 becomes maximum. The maximum value around the X-axis is 35 deg or so, and 18 deg around the Z-axis. Similar to this tendency of head acceleration, the neck shear force decreases around 90 ms, but increases again as the head acceleration is restricted by the lateral bending. The displacements of right and left shoulders and the strains start resuming the initial states around 80 ms, while the upper cervical vertebral torsion and the lateral extension which occurs mainly at the lower cervical vertebra are also started (Figure 5).

Phase 3 [100-300 ms]: The impact loading is already stopped, but the entire body keeps rotating clockwise due to inertia. The T1 acceleration in Y-axial direction converges around 150 ms, whereas the head acceleration remains up to 200 ms or so (Figures 6b)-6d)). The T1 rotation angle around the X-axis shows gradual changes after 100 ms, while the head keeps on rotating. The lateral extension of cervical vertebrae starts to end, resuming the initial states while maintaining the torsion in the Z-axial direction. It is found from the three-dimensional motion data obtained with VICON that the torsion angle around the Z-axis resumes the initial state at 300 ms or so (Figure 7aa)). The lateral extension of cervical vertebrae starts to resume the initial state while maintaining the torsion in the Z-axial direction (Figure 5).

EFFECTS OF DIFFERENCES IN MUSCLE FUNCTIONS ON HEAD/NECK/TORSO IMPACT RESPONSES: It is found from the observation of Y-axial accelerations at the head CG that the head accelerations are suppressed as the stiffness of the head/neck/torso becomes greater (Figure 8). This tendency is found with all impact loads of 400N, 500N and 600N, but there are some cases where the accelerations cannot be suppressed by the muscle tension, as the impact force becomes greater. The maximum value of acceleration at the head CG in the Z-axial direction in the relaxed muscle condition is reduced by 40 % in the tensed condition (Figure 8c)), but hardly any difference is found in the Y-axial direction (impact direction). In other words, no difference in maximum acceleration in Y-axial direction is found between the tensed and relaxed muscle conditions. This is a common tendency found in every impact load. As the Y-axial accelerations at T1 indicate, the torso is subjected to a forced displacement by the impactor (Figures 8d)-8f)). Therefore, hardly any difference is found in the Y-axial accelerations despite the difference in muscle condition. The effect of the muscle tension becomes significant if neck moment or the shoulder strains are considered. The neck moment around the X-axis is suppressed by 24 % by making the tensed muscle condition (Figure 8j)). The variation of right shoulder strain shows that the maximum value is reduced by 20 %, though no difference in the response shape is found over time (Figures 8m) and 8n)).

The X-axial head rotational angle against T1 is roughly 34 deg in relaxed condition, and 25 deg or so in tensed condition, showing a 25 % reduction (Figures 8r)-8t)). Likewise, the Z-axial head rotational angle is 17 deg and 12 deg, respectively, showing about 33 % suppression effect. The X-axial head rotational angle in tensed condition is equal to 18 deg. This is presumably because the torso, head and neck become roughly one unit due to the tension, resulting in no significant change in the head-torso relative angle.

EFFECTS OF DIFFERENCES BETWEEN MALE & FEMALE ON HEAD/NECK IMPACT RESPONSES: Effects of differences between male and female volunteers on the impact responses have been investigated, though the number of female volunteers (3 persons) is relatively small. The maximum values of accelerations at the head CG in Y and Z axial directions are greater in the female volunteers by 38 % and 90 %, respectively (Figures 9a)-9c)), and the rise of acceleration is also quicker than the male volunteers. It is probably because the head mass of female is generally smaller than that of male. The rise of the T1 acceleration in Y-axial direction is also markedly quicker for female (Figures 9d)-9f)). This is attributable to the relatively short distance between the right-shoulder and T1, which can be confirmed by the quicker rise of female shoulder strain (Figures 9m)-9n)).

In the tensed condition, on the other hand, the difference in acceleration at the head CG neck moment is hardly different (Figure 10). However, the difference in right shoulder-T1 strain is greater (Figure 10m)), presumably due to the difference in muscle force. For reference, it should be noted that the difference in strain between the tensed and relaxed muscle conditions is negligible for female, while the difference is about 50 % for male, showing the great effect of muscle force (Figures 9m)-10)).

Great differences are also found in torso rotational angle around the X-axis between male and female. That is, the rotation velocity in tensed condition remains the same for male as the head and torso becomes one combined unit during the rotation due to the relatively strong muscle force. In the case of female with a weaker muscle force, the neck swings right to left and vice versa, which results in a temporary reduction of the torso rotation speed from 90 ms to 170 ms as the reaction to the above. Hence, the difference in head rotational angle around the X-axis relative to T1 becomes great between male and female (Figure 10r)). The maximum value of the head rotational angle is about 35 deg in relaxed condition for both male and female, but it becomes about 24 deg for male and 29 deg or so for female in tensed condition, showing that the neck motion is suppressed markedly for male due to the tension of muscle (Figure 10q)).

EFFECTS OF DIFFERENCE IN IMPACT DIRECTION ON HEAD/NECK IMPACT RESPONSES: In the case where the impact direction is 15 deg backward, the Y-axial acceleration of head CG, neck force, and the neck moment around the X-axis show their maximum values (Figure 11). However,

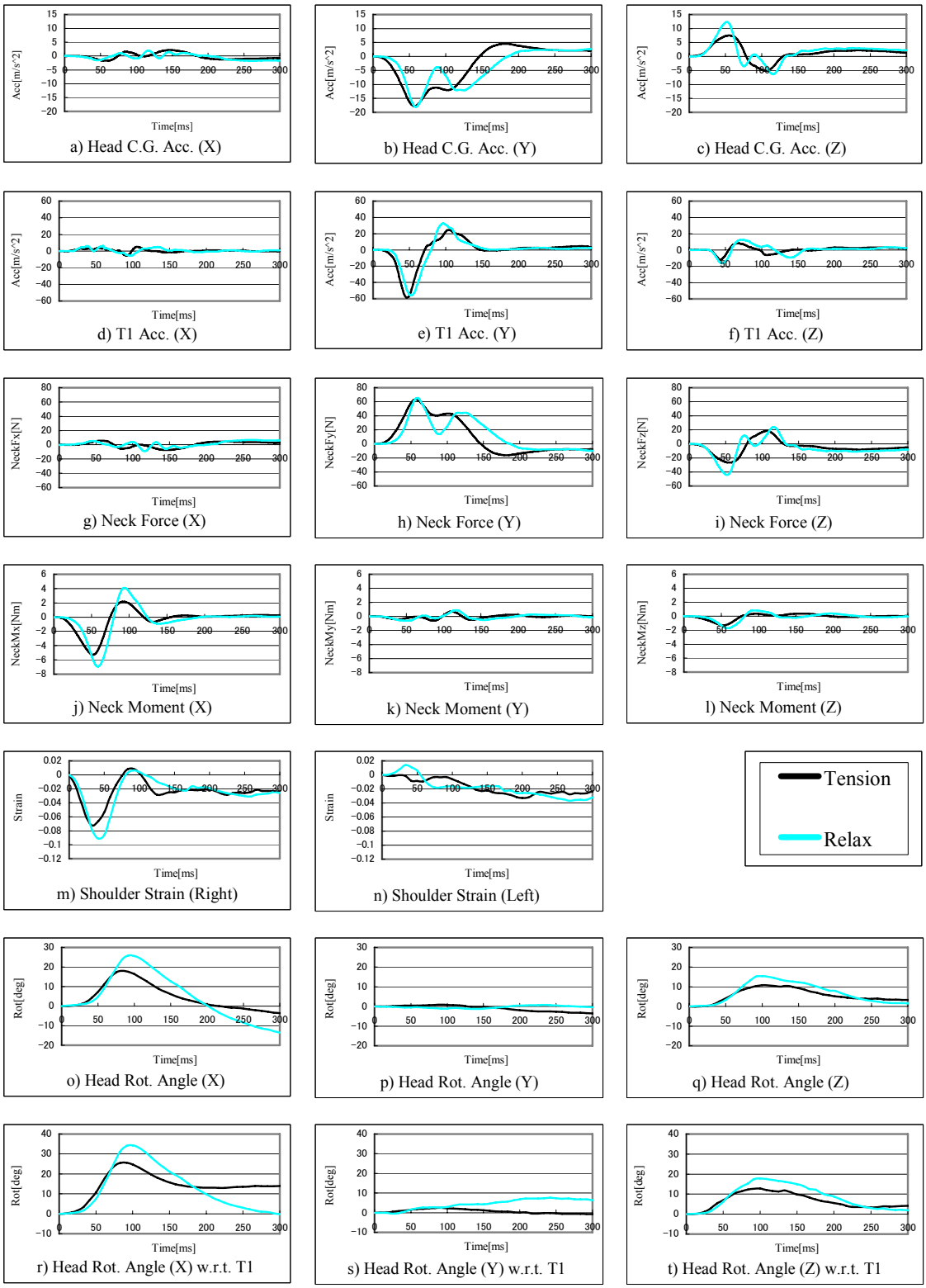

Fig.8 Comparison of head/neck/torso responses between the relaxed and tensed muscle conditions under 600N impact force (8 subjects; Averaged data under each muscle condition)

the T1 acceleration does not show much difference due to different impact directions. That is, in the case where the impact load is applied from a shoulder, the head/neck/torso impact responses differ if the impact direction is different, even though the magnitude of impact force is the same (Figure 11). This suggests that the head/neck/torso motions become different due to the difference in shoulder anatomical shape or structure. Shoulders have high three-dimensional range of motion and a wide movable range, owing to the gleno-humeral and sternoclavicular joints. Therefore, it is easy for shoulders to move vertically or

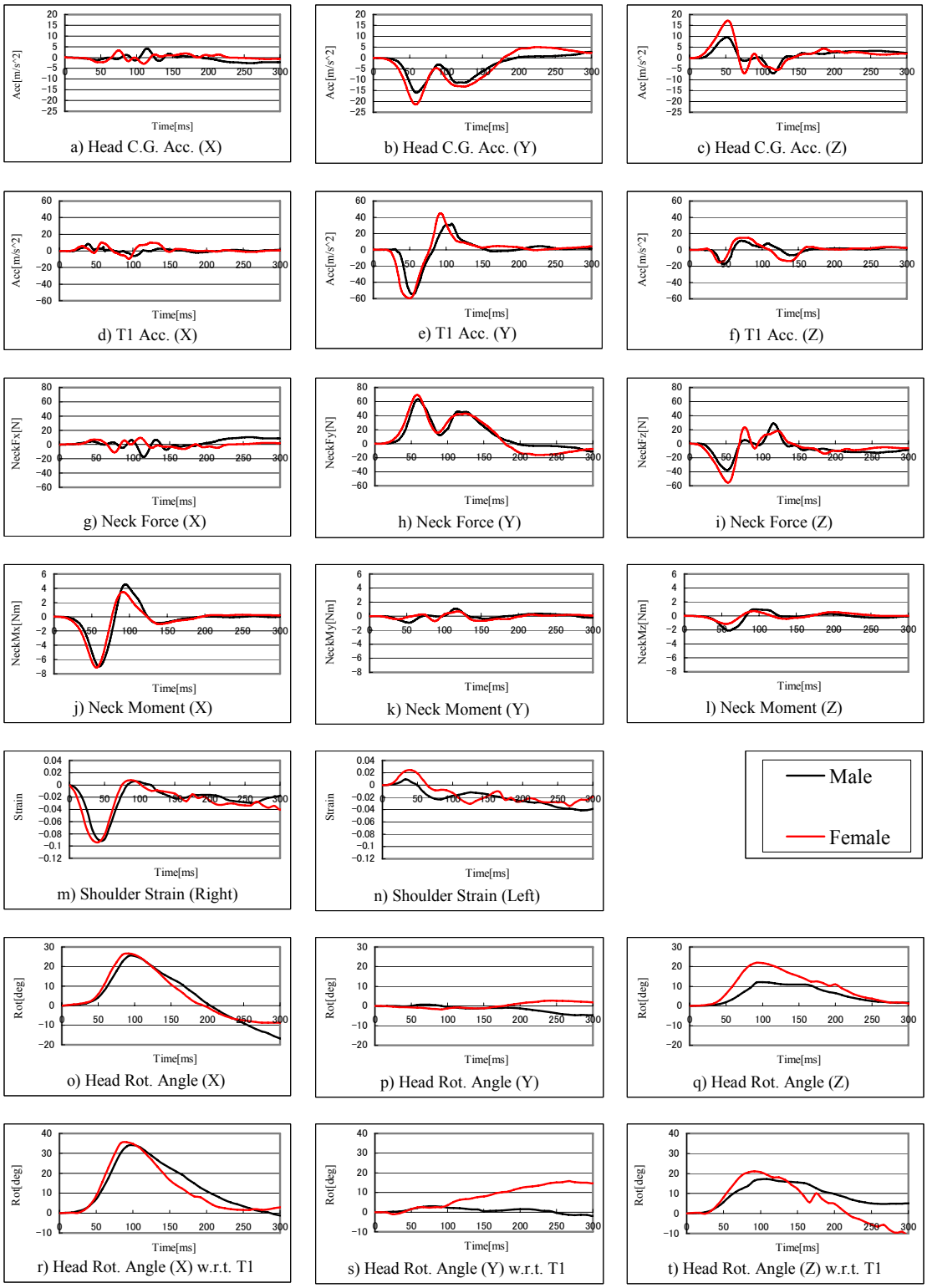

Fig.9 Comparison of head/neck/torso responses between male and female under the relaxed muscle condition (600N) (5 males and 3 females)

longitudinally against a lateral impact. However, the shoulder motion may be restricted where the lateral impact direction roughly aligns with the line connecting the gleno-humeral joint and the sternoclavicular joint - i.e., clavicular longitudinal direction. In the case where the impact is applied from the rear 15 deg direction, the impact from the back increases by 26 %, the head acceleration increases presumably due to the shoulder stiffness caused by the drag in the clavicular longitudinal direction (Figure 11a), 11m), 11o), and 11r)).

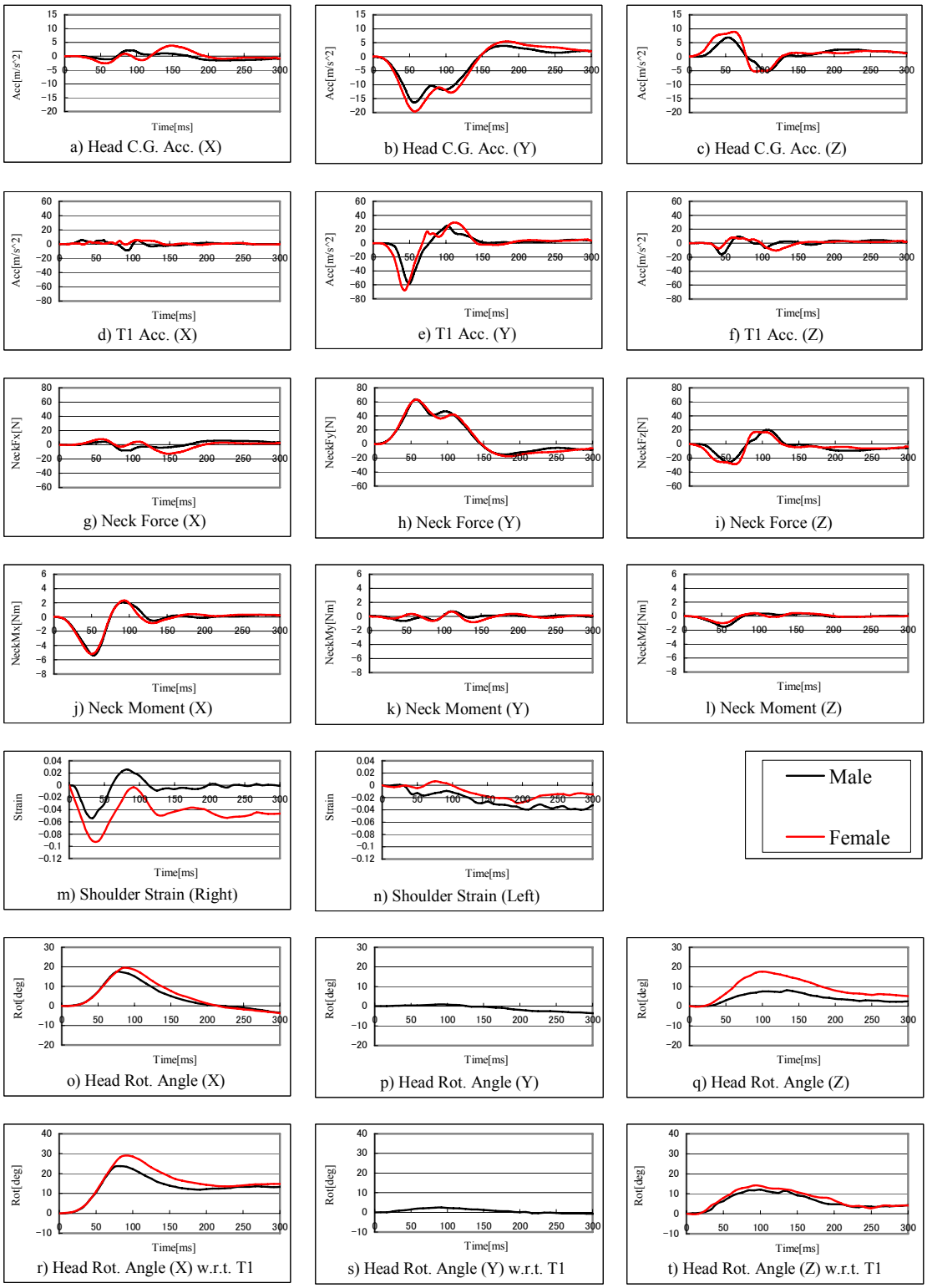

Fig.10 Comparison of head/neck/torso responses between male and female under the tensed muscle condition (600N) (5 males and 3 females)

The strain between right shoulder and T1 shows the greatest value where the impact is applied from 15 deg forward direction (Figure 11m) and 11n)). The torso rotational angle around the Z-axis shows the smallest value where the impact is applied in this direction. It is thus deduced that the rear impact is suppressed as the torso rotates, but the forward impact remains as the shoulder strain without reducing its force, due to the relatively small torso rotation. It is considered that the difference in movability is attributed to the shoulder anatomical shape and structure. The rotational angle of head itself around the

Z-axis relative to T1 hardly shows any difference, indicating a quite strong influence of torso rotation (Figures 10q) and 10f)). It is hence suggested that the head-neck torsion tends to become greatest in rear impact, as the maximum value is 25 deg (Figure 11r)).

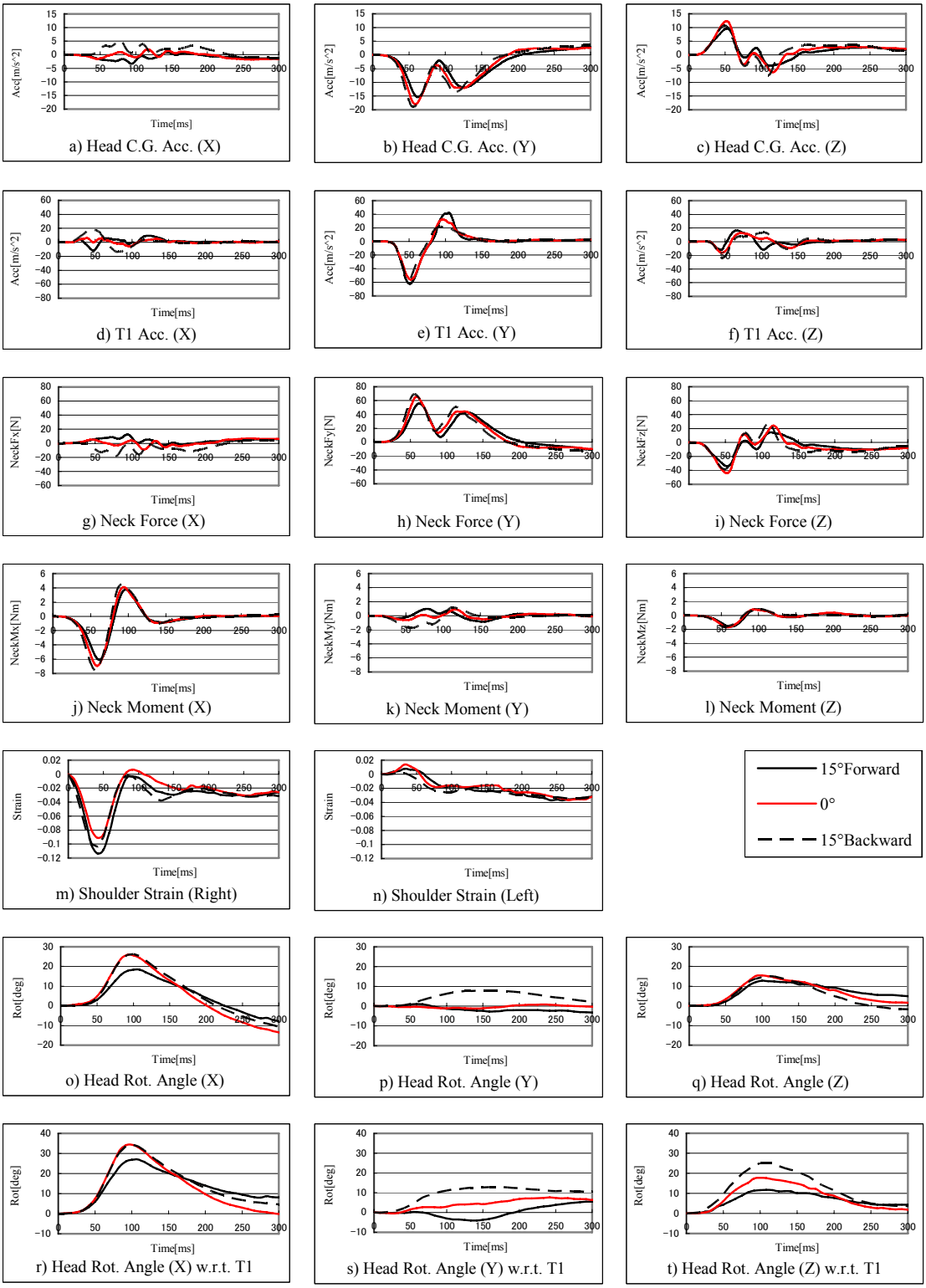

Fig.11 Comparison of head/neck/torso under the different impact directions (600N, Relaxed)

DISCUSSION

HEAD/NECK/TORSO IMPACT RESPONSES IN LATERAL INERTIA IMPACT & MECHANISM OF INJURY INCIDENCE: It is found by comparing the tensed and relaxed condition of the volunteers in lateral impact experiments that the maximum value of neck loading in the Y-axial direction - which is the impact loading direction - does not change significantly. It is suggested by the above that the state of muscle tension in lateral impact hardly affects the head/neck/torso impact responses. However, the neck force in relation to the suppression of neck lateral bending tends to become greater as the impact force becomes greater. The head rotational angle around the X-axis relative to T1 is 34 deg or so in relaxed condition, and about 25 deg in tensed condition (Figures 8r)-8t)). Considering that the head rotational angle is suppressed by 25 % in muscle tensed condition, the effect on the suppression of excessive lateral bending can be also expected. Likewise, the effect of suppressing the torsion of neck, which is the part to connect the head and torso, is also observed, as the torsion around the Z-axis is 18 deg or so in relaxed muscle condition and about 12 deg in tensed condition (Figures 8r)-8t)). It should be noted, however, that the suppressions of the excessive lateral bending and the torsion can be also attributed to the lifting of shoulder when the head and neck bend sideward, owing to the three-dimensional range of motion of the gleno-humeral joints and sternoclavicular joints. In the future, it will be necessary to take account of the shoulder anatomical shape and structure for the analysis of head/neck/torso impact responses in lateral impact. The comparison of head rotation around the X-axis between male and female shows that the maximum value of head rotation is about 35 deg for both male and female in relaxed muscle condition (Figures 9o)-9q)), but it is about 24 deg for male and about 29 deg for female with tensed muscle condition (Figures 10o)-10q)), showing the difference caused by the muscle condition. It is suggested by the above that a greater effect of muscle tension on the suppression of the head rotation can be expected for male than for female (Sferco and Lorenz, 2005).

As regards the impact direction, the greatest value of head rotational angle relative to T1 - 25 deg or so is found in the 15 deg rear impact (Figure 11r)). Due to the scarcity of past data on the relationship between the occupant seating posture and the impact direction in automobile lateral-collisions, it is difficult to predict but can be presumed that the neck torsion may become greater for rear impact in lateral-collision, due to relative motion of head and torso. Therefore, it will be necessary to take a closer look also on the relationship between the impact direction and the occupant sitting posture.

DIFFERENCES BETWEEN VOLUNTEERS & CADAVERS: It can be said that the differences in head/neck/torso impact responses in lateral impact are affected greatly by the anatomical structures and physiological elements of human, such as 1) occupant seating posture, 2) muscle function, 3) shoulder structure, etc. It was decided in this study to conduct investigation on the head/neck/torso impact response characteristics in lateral impact, including the "human" anatomical structures and physiological elements. However, the experiments on volunteers must be limited to the low impact range which could not cause injuries. In this respect, the proper method to correct the various data on human motions, injury incidence levels, etc. collected by experiments on cadavers, is by extrapolating physiological response data obtained by experiments on volunteers. Therefore, the authors et al. requested INRETS to provide the authors with some of their cadaver experiment data (Compigne et al., 2003) obtained with similar experimental protocol as that used in this study, in order to compare such data with the data obtained from the volunteers.

As for the shoulder responses of the volunteers and cadavers, the relationship between the impact forces and the shoulder displacements (acromion point-T1) are shown in Figures 12a)-12d). Figure 12a) shows the individual data of all volunteers, while Figure 12b) shows their corridors. Figure 12c) shows the individual data of all cadavers, while Figure 12d) shows their corridors. The volunteers' data were taken in relaxed muscle condition.

The peaks of impact forces on the volunteers were in the range of about 400 N to 600 N, almost similar to 450 N to 600 N found on the cadavers. The impact forces versus acromion displacements - i.e., the shoulder stiffness wo arm flesh - fluctuate in both the volunteers and the cadavers. In the case of volunteers, the displacements are in the range of 8 to 35 mm with 500 N peak force, and the displacements of cadavers are in the range of 8 to 25 mm with 500 N peak load. The fluctuation in the volunteers is presumably due to the difference in muscle function among the individual volunteers in effort to maintain the initial posture. In the case of cadavers, such physiological muscle responses are naturally absent, but differences in physical structures such as shoulder joints, bladebones, collarbones, etc. would have effects on the fluctuation. The shoulder stiffness of the volunteers tends to increase where the peak load becomes greater than 400 N or so, but this tendency is absent in the cadavers. It is considered attributable to the difference in transfer of shoulder structural deformation affected by the physiological responses due to the

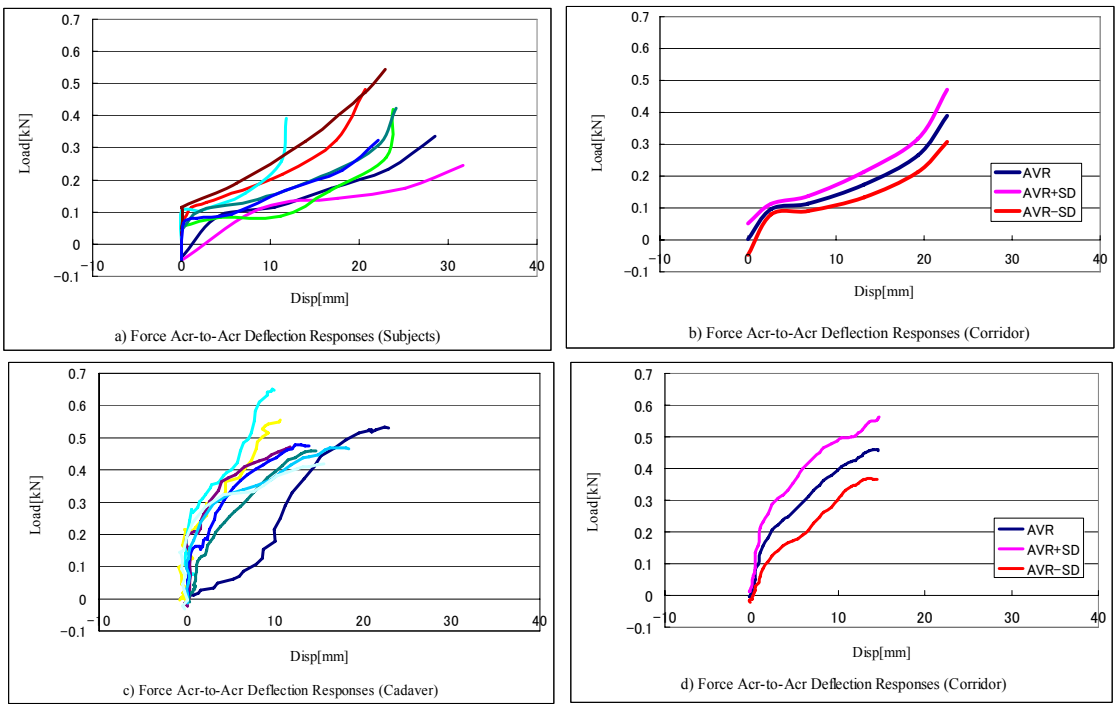

Fig.12 Comparison of force-deflection curves between volunteers and cadavers; 0 deg.

muscle function.

The relationship between the impact forces and the shoulder displacements per test subject shows that the difference in shoulder displacement among individual volunteers is greater than the difference among the cadavers, presumably due mainly to the differences in physical structures and muscle responses among the volunteers. Two female test subjects show relatively great shoulder displacements (maximum 29 mm and 35 mm, respectively). It can be deduced that the effect of difference in muscle response is most significant, considering their shorter distance between the right and left shoulders than the male volunteers. It can be pointed out that the effect of muscle function is also important, as a male volunteer with the largest mass among the male subjects is included in the group showing the smallest shoulder displacement (Viano, 2003).

By observing the corridors of the cadavers and volunteers, it is found that the stiffness of the living bodies during the impact loading process is smaller than that of the cadavers. Although the muscle force differs among the individual volunteers due to the differences in physical structures, the stiffness of the volunteers should be generally smaller than that of the cadavers, in terms of the shoulder impact responses in relaxed muscle condition. The torso inclination angle in the force removal process is greater for the volunteers than the cadavers, presumably due to the volunteer torso rotation.

The ability to hold the seating posture in the initial stage of impact differs between the volunteers and the cadavers due to differences in physiological reactions such as the muscle function, which also cause differences in shoulder impact responses. It is deduced that the differences in impact loads applied from the upper portion of T1 to the lower portion of neck are caused by the difference in the input direction from the shoulder in lateral impact, which result in the relative distortion of head and neck, causing differences in local cervical vertebral motions. Therefore, it will be necessary to analyze the head/neck/torso responses - cervical vertebral motions in particular, based on a clear understanding of the relationship between the impact load applied to the shoulder.

CONCLUSIONS

Using five healthy male and three healthy female adults as human volunteers, experiments on the head/neck/torso impact responses and the cervical vertebral motions upon lateral inertia impact have been conducted, with the impact forces set at 400 N, 500 N and 600 N, respectively. The findings obtained from the above are as follows:

1) Effects of Neck Muscle Functions on Head/Neck/Torso Impact Responses. The suppression of head/neck/torso motions is greater in tensed muscle condition than in relaxed condition. The moment around the neck X-axis is suppressed by the tension of muscle but the shear force in the neck Y-axial direction is hardly suppressed.

2) Effects of Sex Difference on Head/Neck/Torso Impact Responses. Regardless of state of muscle tension, the displacement of acromion with respect to the first thoracic vertebra (T1) is tend to be greater for male than for female. As female shoulders tend to have less flexibility against impact than male, the female cervical vertebral motions are likely to show longer lateral extensions than male. Hence, the effect of shoulder anatomical structure on the head/neck motions is considered significant. According to the foregoing, it is suggested that the differences in muscle responses should be taken into account, in addition to the differences in shoulder anatomical structures, as marked differences between male and female.

3) Effects of Difference in Impact Direction on Head/Neck/Torso Impact Responses. When an impact is applied from a shoulder, the head/neck/neck impact responses become different if the direction of impact differs, even if the magnitude of impact on the torso is the same. It is suggested by the above that the differences in head/neck/torso motions are caused by the differences in shoulder anatomical shape and/or front-rear structural differences. A shoulder has high three-dimensional flexibility and a wide range of movability, owing to the gleno-humeral and sternoclavicular joints, which facilitate vertical and lateral motions against lateral impacts. However, the shoulder movability would be restricted, if the direction of the lateral impact roughly aligns with the line connecting the acromio-clavicular joint and the sternoclavicular joint - i.e., the longitudinal direction of the clavicle.

4) Differences Between Volunteers and Cadavers. The ability to hold the seating posture in the initial stage of impact differs between the volunteers and the cadavers due to differences in physiological reactions, which also cause differences in shoulder impact responses. It is deduced that the differences in impact loads applied from the upper portion of torso (T1) to the lower portion of neck are caused by the difference in the input from the shoulder in lateral impact, which result in the relative distortion of head and neck, causing differences in local cervical vertebral motions. Therefore, it will be necessary to analyze the head/neck/torso responses - cervical vertebral motions in particular, based on a clear understanding of the relationship between the impact load and the shoulders.

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REFERENCES

Compigne S., Caire Y., Quesnel T., and Verriest J., Lateral and Oblique Impact Loading of the Human Shoulder 3D Acceleration and Force-Deflection Data, IRCOBI Conference, Lisbon (Portugal), September, (2003), pp. 265-279

David Viano, Seat Influences on Female Neck Responses in Rear Crashes: A Research Why Women Have Higher Whiplash Rates, Traffic Injury Prevention, Volume 4, Issue 3, 2003, pp.228-239

Hell W., Hopfl, F., Langweider K., and Lang D., Cervical Spine Distortion Injuries in Various Car Collisions and Injury Incidence of Different Car Types in Rear-end Collisions, Proceedings of IRCOBI Conference, 2003, pp.193-206

Kraft M., Kullgren A., Ydenius A., Tingval C., Influence of Crash Pulse Characteristics on Whiplash Associated Disorders in Rear Impact-Crash Recording in Real-Life Crashes. Journal of Crash Prevention and Injury Control, 2002

Narayan Yoganandan, Srirangam Kumaresan, Frank Pinter, Review Paper, Biomechanics of the cervical spine Part2. Cervical spine soft tissue responses and biomechanical modeling, Clinical Biomechanics 16, 2001, pp.1-27

Ono K., Kikuchi A., Nakamura M., Kobayashi H., and Nakamura N., Human Head Tolerance to Sagittal Impact - Reliable Estimation Deduced from Experimental Head Injury Using Subhuman Primates and Human Cadaver Skulls, 1980 Transactions of the Society of Automotive Engineers, SAE Paper No. 801303, 1980, pp. 3837-3866

Raimondo Sferco and Bernd Lorenz, Neck Injuries Real World Data Male/Female Comparison, Unpublished Ad-hoc 2nd GTR Meeting on April 2005

Shigeki Ito, Paul C. Ivancic, Manohar M. Panjabi, and Bryan W. Cunningham,, Soft Tissue Injury Threshold During Simulated Whiplash A Biomechanical Investigation, SPINE Volume29, Number9 , 2004, pp979–987

Shrawan Kumar, Robert Ferrari, and Yogesh Narayan, Looking Away From Whiplash: Effect of Head Rotation in Rear Impacts, SPINE Volume30, Number7, 2005, pp760–768

WHO/CIOMS proposed guidelines for medical research involving human subjects, and the guidelines on the practice of ethics committees published by the Royal College of Physicians, The Lancet, November 12, 1988, pp.1128-1131