

## Consideration on Gender Difference of Whiplash Associated Disorder in Low Speed Rear Impact

Yuichi Kitagawa, Katsunori Yamada, Harutoshi Motojima, Tsuyoshi Yasuki

**Abstract** Low speed rear impacts were simulated using finite element (FE) models. THUMS Version 4 AM50 and AF50 Occupant Models were used for representing average-sized male and female occupants, respectively. Capsule models were introduced into the cervical joints to calculate tissue strain for estimating the neck load. Rear impact simulations were conducted to investigate the gender difference of head and cervical spine kinematics. The first series of simulations compared the head and cervical spine kinematics of the male and female occupants without muscle tone, assuming the same backset. The next series was performed with muscle tone in the neck. The other series was conducted to investigate the effect of backset for the female occupant. Three acceleration pulses were used for the first series. The simulation results indicated that the greater neck extension of the female occupant was due primarily to the less stiff neck. The muscular force worked both raising and lowering the joint capsule strain and resulted in bridging the gender difference. The shortest backset did not necessarily give the lowest joint capsule strain. The relative displacement between the head and torso was given by the head and torso displacement pushing against the head restraint and seatback.

**Keywords** rear impact, whiplash associated disorder, human body FE model, female occupant.

### I. INTRODUCTION

Researchers indicated that female occupants were more likely to sustain whiplash associated disorder (WAD) in low speed rear impacts than male occupants. Carlsson et al. reviewed the previous studies and summarised that the risk of WAD in female occupants was between 1.5 and 3 times higher than that in male occupants [1]. Kullgren et al. proved the effectiveness of anti-whiplash seat system based on their insurance data, but mostly for male occupants rather than for females [2-3]. Linder et al. investigated rear collision data collected from the particular Swedish vehicle make in a certain model year range, and analysed the influence of gender, height, weight, age, seated position and collision site [4]. One of their conclusions was that females aged 35–44 had higher risk of WAD than males in the same age group, and females in braking scenarios also had higher risk than males. Despite such evidence showing the gender difference in the risk of WAD, its mechanism and reason have not yet been discovered. Carlsson et al. analysed rear impact test data at delta-V's of 4 km/h and 8 km/h, with volunteer subjects in groups of 21 males and 21 females [5]. They noted relatively earlier and higher head acceleration peaks in the female subjects. They also suggested that the exclusive use of 50th percentile male dummies could limit the assessment and development of whiplash prevention systems for protecting both male and female occupants. Carlsson et al. developed a 50th percentile female rear impact crash dummy FE model, named EvaRID, by scaling the BioRID, the 50th percentile male rear impact crash dummy [6]. They validated the EvaRID model to the volunteer rear impact test data [7] and found good correlations for most acceleration and displacement responses. Research efforts were made to understand the gender difference of cervical spine kinematics in low speed rear impacts. Sato et al. analysed two series of rear impact sled tests with volunteer subjects including both males and females [8]. They noted that the females had a peak flexion of the head relative to the neck link, defined as a line between T1 and the occipital condyle, while the neck link was in extension at the time of peak head flexion. On the other hand, the males had flexion in both the head relative to the neck link and in the neck link relative to T1. Factors causing the gender difference have not yet been identified, however. Human subject test data may include variety among individuals and show variability in impact responses. The variability could hide effects of factors, especially in low severity impacts.

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Numerical simulations reveal the effect of each factor and magnify it by applying high severity load. This study conducted rear impact simulations using male and female occupant FE models in order to understand the gender difference of head and cervical spine kinematics for identifying the factors. Different acceleration pulses were used to ensure the gender difference could be observed in the simulation results.

## II. METHODS

The average-sized male and female occupants were represented by the THUMS Version 4 AM50 and AF50 Occupant Models, respectively. The THUMS series has been jointly developed by TOYOTA MOTOR CORPORATION and TOYOTA Central R&D Labs., Inc. The Version 4 is characterised by its fine mesh for precisely representing the human body structure. The AM50 geometry was based on a CT scan data of a 29-year-old male with a height of 173 cm and a weight of 77 kg. The AF50 geometry was generated by scaling CT scan data of a 38-year-old female with a height of 154 cm and a weight of 52 kg. The use of CT scan data for modelling was permitted by the Michigan Institutional Review Board. Tissue properties were defined based on the literature data [9-10]. Mechanical responses of the model against various impacts have been validated to the literature data [11]. The model is capable of simulating common injuries induced by car collisions, such as bony fracture, ligament rupture, brain damage, internal organ injury, etc. The cervical spine part has been updated for estimating potential soft tissue damage during low speed rear impacts. Activation of cervical muscles was approximated for simulating muscle tone of living human subjects. A prototype vehicle seat model and the Euro NCAP acceleration pulses were used for simulating rear impacts.

### ***Modelling of Cervical Spine***

The skeletal spine system consists of vertebral bodies, intervertebral discs and ligaments connecting them. The cervical vertebrae have relatively great transverse processes with foramens. The atlas (C1) and the axis (C2) have unique geometry, while the other vertebrae (C3-C7) have similar geometry. The facet joints are located on bilateral sides of the spinal column; the joint surfaces are diagonal to the horizontal plane. When flexion or extension occurs, the vertebra slides along the joint surface of the adjacent one. The joint is lubricated with the synovial fluid and is covered by the capsule tissue. The range of motion is restricted by the ligaments. The intervertebral discs help mitigate impact to the spine column. The cervical spine model included all such tissues with material properties documented in the literature. The geometry of the cervical spine was obtained from the CT scan data described above. The cortical bones were modelled with shell elements, while the trabecular bones were modelled with solid elements. Elasto-plastic materials were assumed for the bony tissues. The intervertebral discs were modelled with solid elements with visco-elastic material. The ligaments, including the joint capsules, were modelled with membrane elements assuming an elastic material. Joint motions were simulated by sliding of joint surfaces with resistance from the surrounding ligaments. The synovial fluid was modelled with solid elements assuming the bulk modulus of saline. The material properties were defined based on the literature data [9][12]. In this study, the joint capsule was focused on as a potential site of neck pain. Figure 1 shows the cervical spine model for the AM50 model. The model was validated at component level first, then implemented in the whole AM50 model with muscle model described later. It was also scaled approximately by 0.86, to match the mid-size female dimensions [13]. Then the cervical spine part of the whole AF50 model was replaced with the scaled cervical spine model. No change was made for the material properties.

### ***Modelling of Muscles***

The cervical spine model included four major muscles contributing to flexion and extension of the neck. The sternocleidomastoid muscle is one of the largest and most superficial cervical muscles, passing obliquely across the side of the neck, originating from the manubrium sterni and the medial portion of the clavicle, inserting into the mastoid part of the temporal bone and the superior nuchal line. The longus capitis muscle originates from the transverse processes of C3-C6, and inserts into the inferior surface of the basilar part of the occipital bone. The splenius capitis muscle arises from the nuchal ligament and the spinous processes of C7-T3, and inserts into the mastoid process of the temporal and occipital bone. The semispinalis capitis muscle originates from the transverse processes of the inferior cervical and superior thoracic column. One-dimensional discrete elements

were used for representing these muscles. For each left and right, the sternocleidomastoid muscle was substituted by a single element; the longus capitis muscle was represented by four elements; the splenius capitis muscle was simplified as a single element; the semispinalis capitis was modelled with four elements. Figure 2 shows the muscle model for the AM50 model. Ono et al. conducted rear impact sled tests with twelve volunteer subjects and optically measured the head and cervical spine kinematics [14]. Their data set included cases with and without muscle tone at a delta-V of 6 km/h, and indicated that the muscle tone reduced the maximum angle of the head rotation by approximately 30–40%. Iwamoto et al. estimated the muscle activity of the cervical muscles for simulating the head and cervical spine kinematics of the tensed volunteer subject [14] using their human body FE model [15]. In this study, muscle tone was simulated with one-dimensional muscle elements. The force magnitudes were adjusted so that the maximum angle of the head rotation became 30% smaller than that without muscle tone. The contribution ratios of the cervical muscles were assumed to be the same as those estimated by Iwamoto et al. The force magnitudes for the AF50 model were scaled from those for the AM50 by 0.725, which was the ratio of muscle volumes between genders.

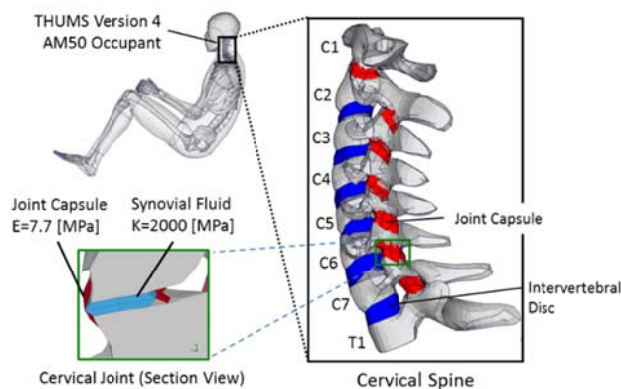


Fig. 1. Cervical Spine Model.

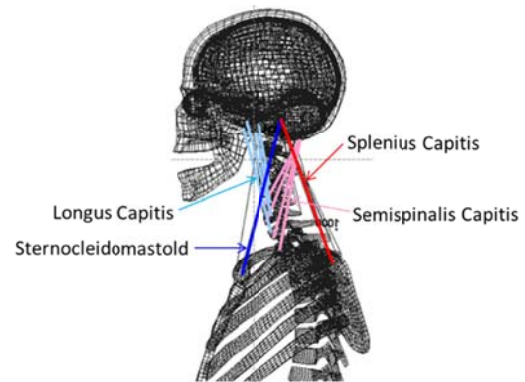


Fig. 2. Muscle Model for AM50.

### Seat Model

A common configuration of vehicle seats includes a cushion, a seatback and a head restraint. Each part has components such as frames, springs, urethane foams and covers. Adjusting gears are also installed. The cushion frame is mounted on longitudinal rails for sliding back and forth. An FE model of a prototype seat was generated for this study. The prototype seat had the common configuration and components described above. Solid elements were used for representing the urethane foams; shell elements were used for the frames and covers; one-dimensional elements were used for the springs. Elasto-plastic material was assumed for the metal parts; low density foam material was assumed for the urethane foam; fabric material was assumed for the seat cover; and kinematic joints were used for representing the adjusting gears. Such modelling techniques are commonly used for generating product seat models. Figure 3 shows the seat model used for the study.

### Component Validation

The cervical spine model was validated at component level first, in order to verify the vertebral motions under a prescribed loading condition. Siegmund et al. examined mechanical response of cervical spine segments under compressive and posterior shear forces [16]. They tested seven C3-C4 segments and six C5-C6 segments. The test data of the C3-C4 segments were used for the model validation. The C3 displacement and rotation with respect to C4 were monitored. An array of markers was applied to the external surface of the capsular ligament for calculating the joint capsule strain. The C3-C4 segment was extracted from the cervical spine model (AM50) for representing the test. The combination of compressive and posterior shear forces was applied to the segment model. Figure 4 shows the validation model and the force time history curves applied to the C3 part. The X-displacement and Y-rotation were calculated from nodal displacement on the C3 part. The joint capsule strain was represented by the highest value among maximum principal strains calculated at the elements belonged to the capsule part.



Fig. 3. Prototype Seat FE Model.

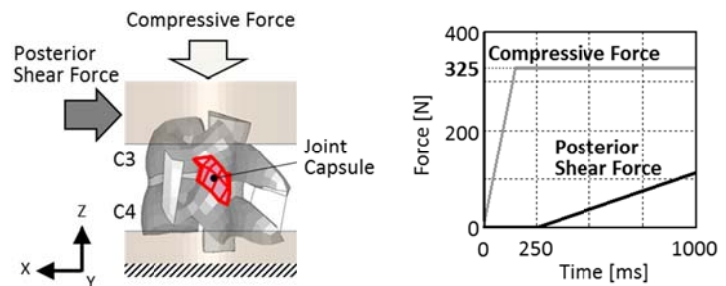


Fig. 4. C3-C4 Segment Validation Model.

### Whole Body Validation

Whole body kinematics of the model was verified by comparing to those of human subjects under simulated rear impact conditions. Ono et al. conducted a series of sled tests with six volunteer subjects, four males and two females, sitting on a rigid seat without head restraint [17]. The sled was accelerated along horizontal rails at a delta-V of 6 km/h. The head horizontal displacement and rotational angle with respect to C7 were measured. The cervical vertebral rotation angle was analysed using X-ray images. The AM50 and AF50 models were used for representing the male and female subjects, respectively. The subject models were placed on a rigid seat model with the same seat cushion and seatback angles. Then the seat model was accelerated, simulating the sled test. The head and cervical spine kinematics were calculated from the nodal displacement of the corresponding part in the model. Figure 5 shows the entire view of the validation model, with the geometry profiles of the volunteer subjects superimposed.

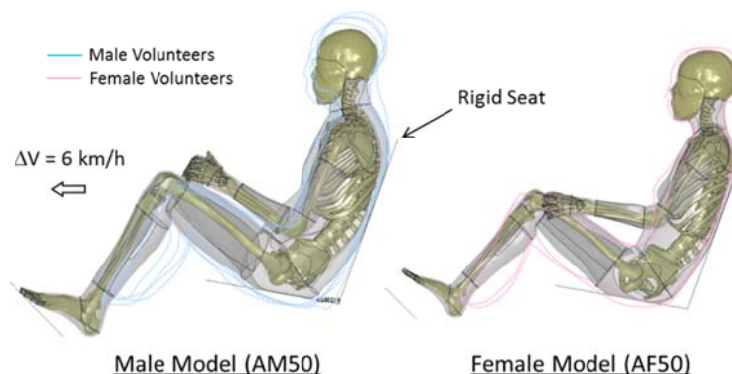


Fig. 5. Whole Body Validation Model.

### Rear Impact Simulation

The gender difference was investigated under simulated rear impact conditions assuming with and without muscle tone, three different backsets, and three different acceleration pulses. Figure 6 shows the AM50 and AF50 models used for the simulations. The AM50 model was placed on the seat model basically following the Euro NCAP whiplash protocol 3.1. The bottom parts of the longitudinal rails were fixed to a rigid plate representing a sled used in rear impact tests. The seatback angle was adjusted to 25 deg. The head restraint position was adjusted to the middle height. The backset was adjusted to 40 mm. The AF50 model was placed on the same seat model. The head restraint was positioned at the lowest height. The sitting height of the AF50 model was lower than that of the AM50 by 40 mm. The backset was adjusted to the same distance in a baseline case while it was changed in some other cases. After seating the occupant model to the seat model and stabilising under the gravity, the sled was accelerated forward using the Euro NCAP acceleration pulses shown in Fig. 7. The termination time of simulation was set to 150 ms. Table I shows the simulation matrix, including 10 cases in total. Cases 1 and 2 compared the head and cervical spine kinematics between the AM50 and AF50 models without muscle tone at a delta-V of 16 km/h with a triangular pulse. Cases 3 and 4 compared two models with muscle tone under the same impact condition. In Cases 5 and 6, the backset for the AF50 model

was shortened and elongated, respectively, by changing the head restraint position forward and rearward. Cases 7 and 8 compared the AM50 and AF50 models at a  $\Delta V$  of 16 km/h with a rectangular pulse. Cases 9 and 10 compared two models at a  $\Delta V$  of 24 km/h with a rectangular pulse. The head and cervical spine kinematics were analysed, and the joint capsule strain was calculated.

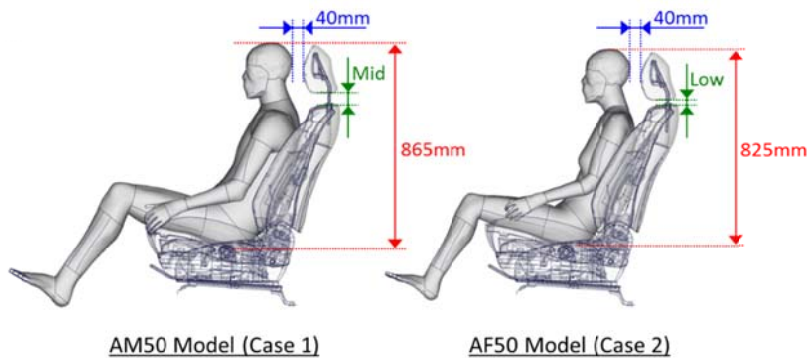


Fig. 6. Rear Impact Simulation Model.

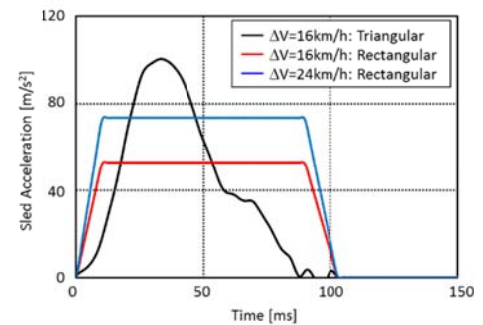


Fig. 7. Euro NCAP Acceleration Pulses.

TABLE I  
SIMULATION MATRIX

| Case | Gender | Muscle Tone | Backset [mm] | Acceleration Pulse                |
|------|--------|-------------|--------------|-----------------------------------|
| 1    | AM50   | No          | 40           | $\Delta V = 16$ km/h, Triangular  |
| 2    | AF50   | No          | 40           | $\Delta V = 16$ km/h, Triangular  |
| 3    | AM50   | Yes         | 40           | $\Delta V = 16$ km/h, Triangular  |
| 4    | AF50   | Yes         | 40           | $\Delta V = 16$ km/h, Triangular  |
| 5    | AF50   | No          | 20           | $\Delta V = 16$ km/h, Triangular  |
| 6    | AF50   | No          | 80           | $\Delta V = 16$ km/h, Triangular  |
| 7    | AM50   | No          | 40           | $\Delta V = 16$ km/h, Rectangular |
| 8    | AF50   | No          | 40           | $\Delta V = 16$ km/h, Rectangular |
| 9    | AM50   | No          | 40           | $\Delta V = 24$ km/h, Rectangular |
| 10   | AF50   | No          | 40           | $\Delta V = 24$ km/h, Rectangular |

### III. RESULTS

The component validation was achieved to ensure the prediction accuracy of joint capsule strain calculated in THUMS as well as vertebral motions under the well-controlled loading condition. The whole body validation was conducted to examine the validity of whole spine kinematics of THUMS on the rigid seat model. The study assumed that the validated THUMS could simulate head motion and cervical spine kinematics, vertebral motions and joint capsule strain in rear impacts and show comparable responses when changing factors such as muscle tone and other impact conditions.

#### Component Validation

Figure 8 compares the X-displacement, Y-rotation and joint capsule strain of the C3-C4 segment between the test and the validation models. The test data were shown as corridors, while the calculation results were plotted as curves. The posterior shear force was plotted in the horizontal axis in each graph. All the calculated curves were within the test corridors. The joint capsule strain exhibited an initial rise, while the other two responses increased linearly as the posterior shear force increased. This is due to the incompressive characteristics of the synovial fluid in the joint. As shown in Fig. 4, the compressive force was applied first followed by the shear force. When the joint was compressed, the synovial fluid expanded in the lateral direction (joint surface plane) pushing the capsule tissue outward. It was postulated that the cervical spine model was capable of estimating joint capsule strain due to vertebral motion.



### Whole Body Validation

Figure 9 shows the whole body motions of the AM50 and AF50 models from a lateral view at every 100 ms. The AF50 model showed greater head rotation compared to the AM50. Figure 10 compares the time history curves of X-displacement and Y-rotation of the head between the test and the validation model. The test data were plotted in thin colours, while the calculated data were plotted in thick colours. The blue colour indicated the male responses and the red colour indicated the female responses. Both the male subjects and the AM50 model showed greater X-displacement of the head than the female subjects and the AF50 model approximately from 70 ms to 180 ms, but finally converged at the same magnitude. Both the female subjects and the AF50 model showed greater Y-rotation of the head than the male subjects and the AM50 during the second half. Figure 11 compares the maximum vertebral rotations between the male and female obtained from the test and the validation model. The test data were the averages of the volunteer subjects. The positive angle indicates the extension while the negative angle indicates the flexion. In both test and validation model, the females showed greater rotations compared to the males. The agreement in such tendencies between the test and the validation model suggested that the THUMS AM50 and AF50 models were valid for investigating the gender difference of cervical spine kinematics in low speed rear impacts.

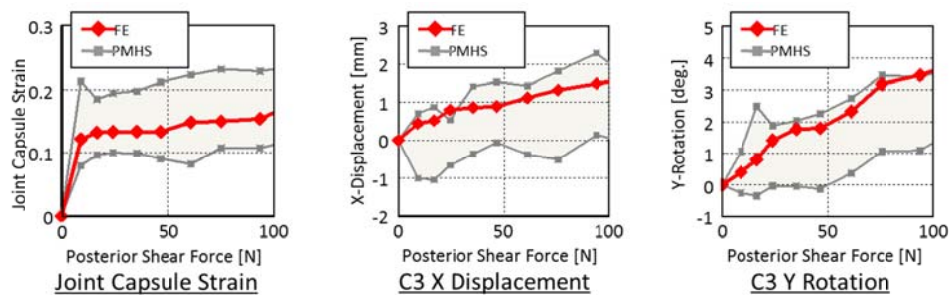


Fig. 8. C3-C4 Segment Validation Result.

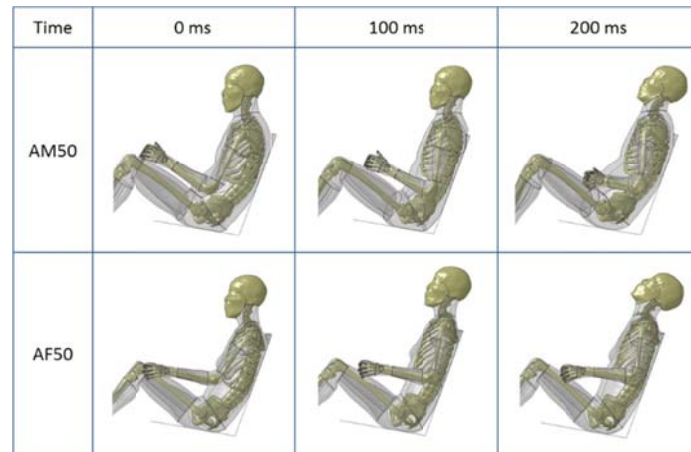


Fig. 9. Whole Body Kinematics.

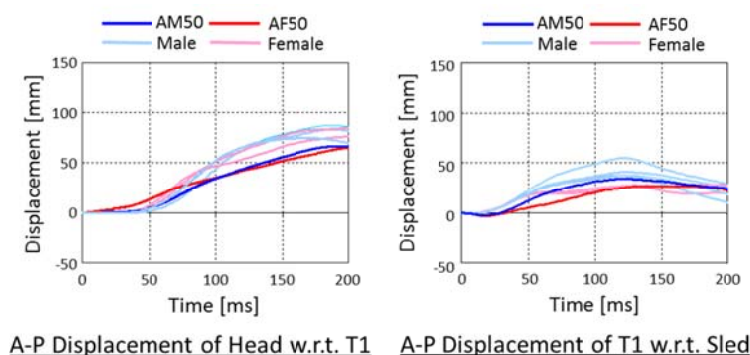


Fig. 10. Anterior-Posterior Displacement of Head and T1.

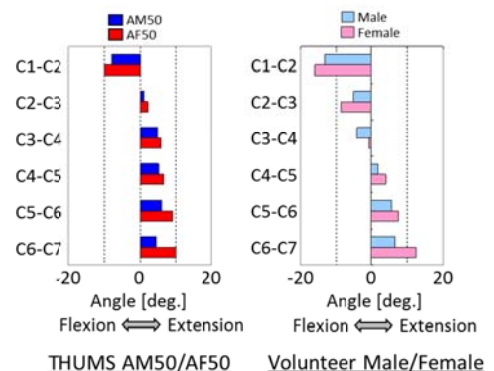


Fig. 11. Vertebrae Rotations at 90 ms.

### Rear Impact Simulation

Figure 12 shows the whole body kinematics of the AM50 and AF50 models at every 25 ms in Cases 1 and 2. The figure shows the section geometry in the median plane in the sled coordinate system. The gross motions were similar to each other. The torso contacted the seatback while the head approached the head restraint during the first half. (Note that the torso was pushed forward by the seatback while the head stayed at the initial position in the global coordinate system.) The contact timings between the head and the head restraint were 54 ms in both cases. The head pushed against the head restraint and finally rebounded from 102 ms in Case 1 and from 96 ms in Case 2. While the torso pushed against the seatback, the spine was straightened. Figure 13 compares the head and cervical spine kinematics between two models. The head and cervical spine kinematics were displaced in the T1 coordinate system. The initial geometries were superimposed in grey colour. The AM50 cervical spine showed an extension, while the AF50 exhibited an S-shape mode. When rebounding, the cervical spine was flexed with the head moved forward in both models.

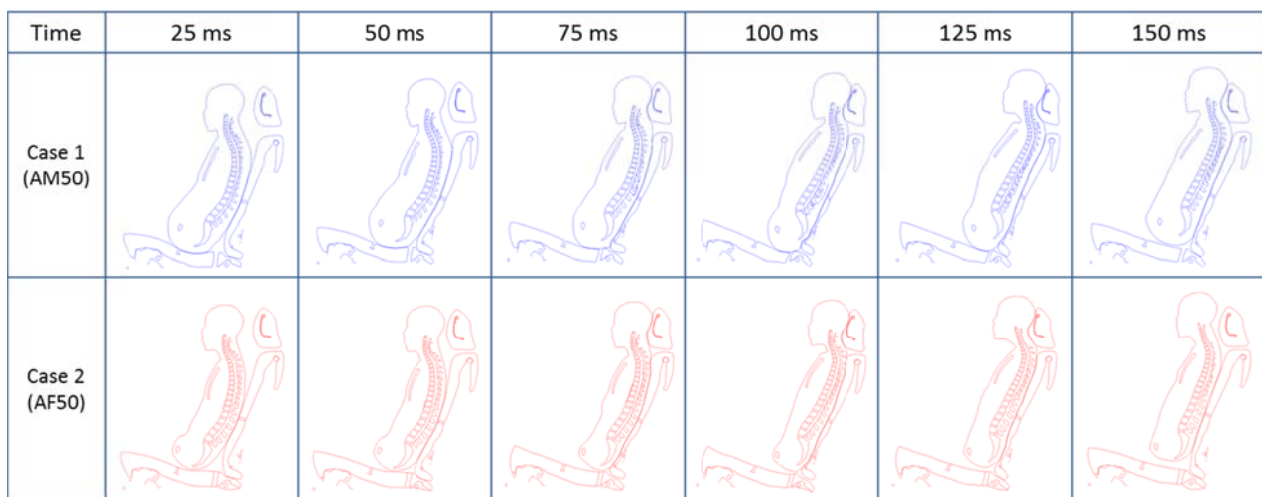


Fig. 12. Whole Body Kinematics (Cases 1 and 2).

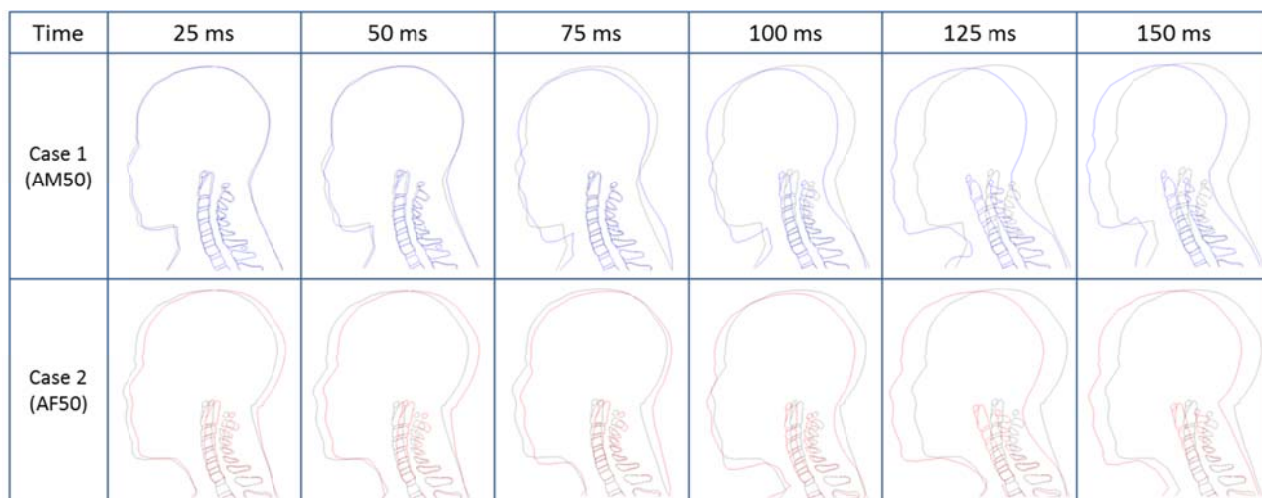


Fig. 13. Head and Cervical Spine Kinematics (Cases 1 and 2).

Figure 14 shows the time history curves of relative vertebral rotations in two models. The positive value indicates extension of the vertebra with respect to the adjacent inferior one. The AM50 model showed greatest extension in C7-T1 up to 100 ms, while the AF50 model showed greatest extension in C4-C5 from 50 ms to 100 ms. Figure 15 shows the time history curves of joint capsule strains from C2-C3 to C7-T1 in two models. The AM50 model showed the first strain peak in C7-T1 when the relative vertebral rotation of that joint was extended. The AF50 model showed the first and highest peak in C4-C5 when the relative rotation of that joint was extended. This study focused on the first peak of joint capsule strain, during neck extension, as a possible indicator for estimating the risk of WAD. The AM50 model showed the first strain peak (0.083) in C7-T1, while

the AF50 model showed the first peak (0.118) in C4-C5. Note that each rear impact simulation was started after seating the occupant model to the seat model and stabilising under the gravity. The strain value was compensated to be zero at the timing of impact. The negative strain value indicated loosening of the capsule ligament compared to the state at the timing of impact. Figure 16 shows the time history curves of C2-T1 distance changes in two models. The negative value indicates shortening of the distance. The AM50 model showed the negative peak approximately at 80 ms, when the joint capsule strain in C7-T1 reached the first peak. The AF50 model showed the negative peak approximately at 95 ms, when the joint capsule strain in C4-C5 reached the first and maximum peak.

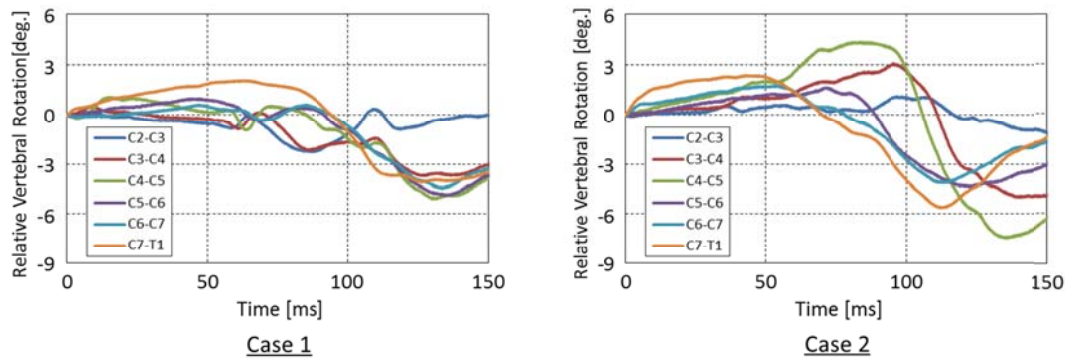


Fig. 14. Time History Curves of Relative Vertebral Rotations (Cases 1 and 2).

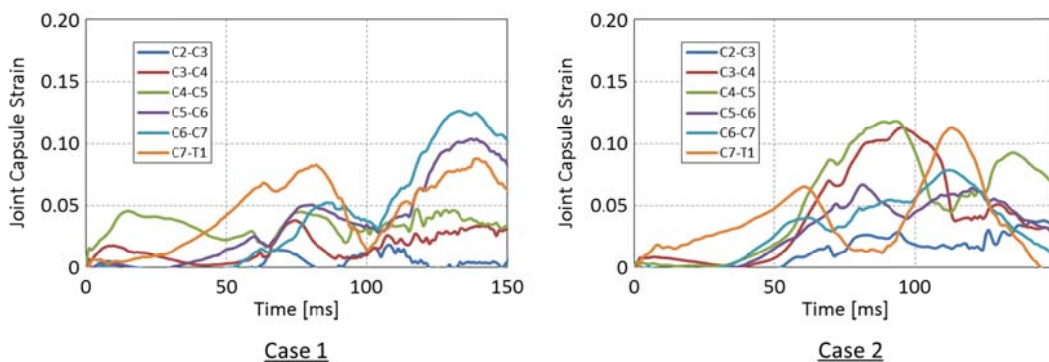


Fig. 15. Time History Curves of Joint Capsule Strains (Cases 1 and 2).

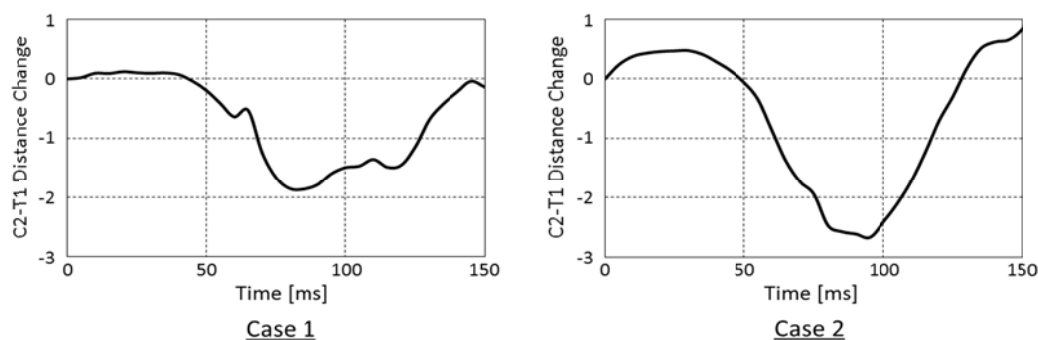


Fig. 16. Time History Curves of C2-T1 Distance Changes (Cases 1 and 2).

Figure 17 shows the head and cervical spine kinematics in Cases 3 and 4. The head motions were shown in the T1 coordinate system. The head motions in Cases 1 and 2 were superimposed in grey colour. Basically, the head motions were similar between with and without muscle tone. However, the head was slightly forward and low position in the cases with muscle tone. The head forward position indicated a smaller neck extension, while the low position indicated compression. Figure 18 shows the time history curves of joint capsule strain in Cases 3 and 4. The AM50 model showed the highest strain peak (0.130) in C3-C4, while the AF50 model also showed the highest peak (0.145) in C3-C4. The strain values were higher than those without muscle tone in Figure 15. The



gender difference in strain values was smaller than that without muscle tone.

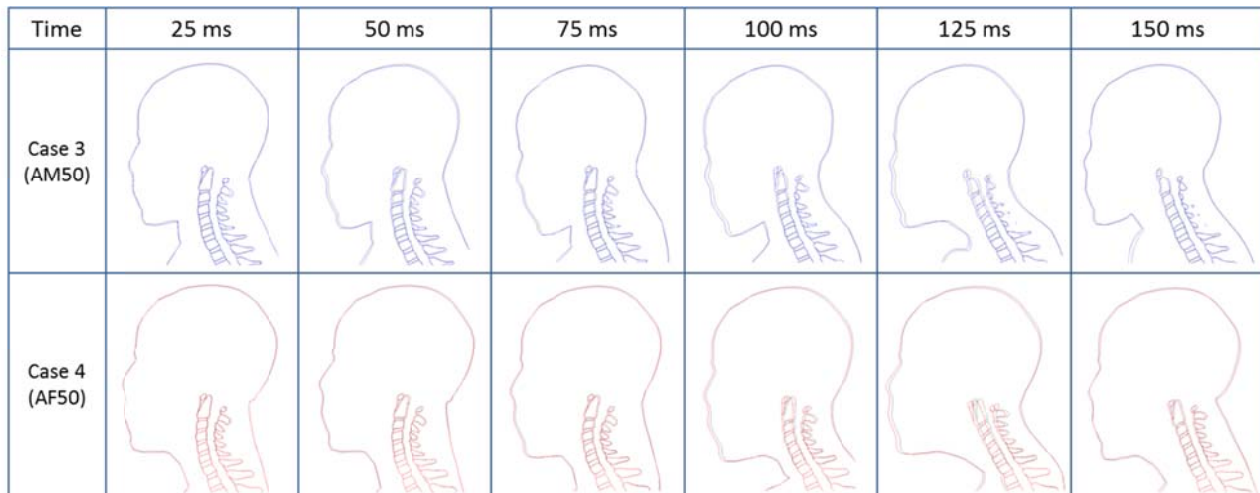


Fig. 17. Head and Cervical Spine Kinematics (Cases 3 and 4).

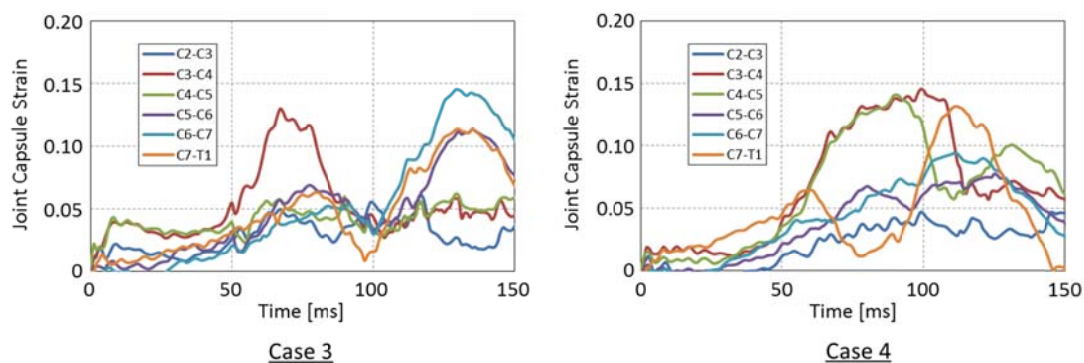


Fig. 18. Time History Curves of Joint Capsule Strains (Cases 3 and 4).

Figure 19 compares the head and head restraint interaction at 100 ms, approximately at the maximum displacement, between Cases 2, 5 and 6. The head motions were shown in the T1 coordinate system. The initial geometries were superimposed in grey colour. The superior end of the cervical spine was close to the initial position in Case 2, while it was forward in Case 5 and rearward in Case 6. The forward position indicates flexion of the cervical spine, while rearward position indicates extension. No significant difference of the cervical spine alignment was noted in Case 2. The cervical spine showed an S-shape mode in Case 5, while it was extended in Case 6. The head contact timings were 54 ms, 48 ms and 72 ms, respectively, lined up in order of the backset. Figure 20 compares the time history curves of joint capsule strain between Cases 2, 5 and 6. The joints with the highest first peak were selected for each case; 0.118 in C4-C5 in Case 2, 0.157 in C3-C4 in Case 5 and 0.124 in C3-C4 in Case 6. The strain peak value was the lowest in Case 2 and was the highest in Case 5 despite that the backset was shortest.

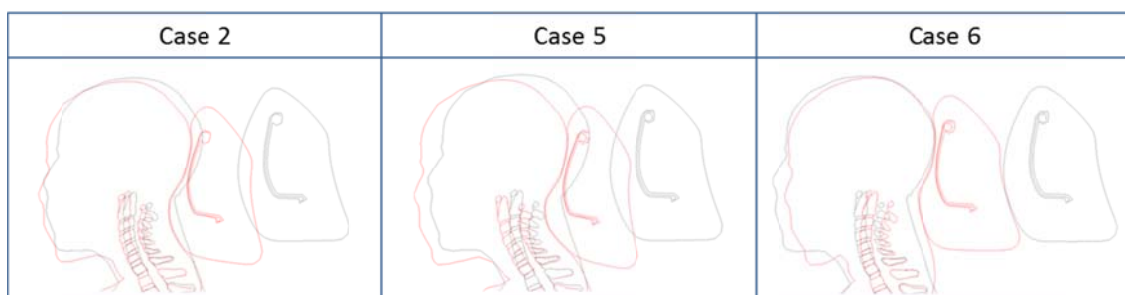


Fig. 19. Head and Head Restraint Interaction at 100 ms (Cases 2, 5 and 6).

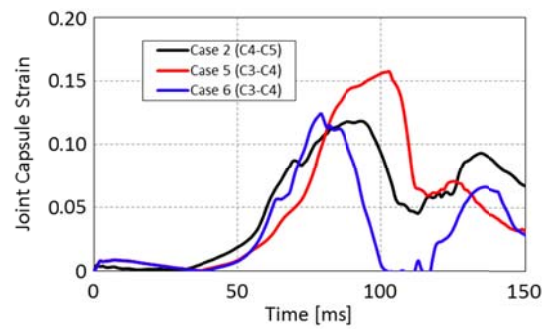


Fig. 20. Time History Curves of Joint Capsule Strains (Cases 2, 5 and 6).

Figure 21 shows the time history curves of joint capsule strain in Cases 7, 8, 9 and 10. The curves of AM50 (Cases 7 and 9) were similar to each other, and those of AF50 (Cases 8 and 10) were similar to each other. The first peak strain values were 0.061 in C7-T1, 0.102 in C3-C4, 0.075 in C7-T1 and 0.117 in C4-C5, respectively. Figure 22 compares the first peak values in Cases 1, 2, 7, 8, 9 and 10. The AF50 model showed higher strain peaks than the AM50 model in all selected cases. The ratios were 1.42, 1.67 and 1.56, respectively. Table II summarises the first peak value of joint capsule strain, joint location and time in all cases.

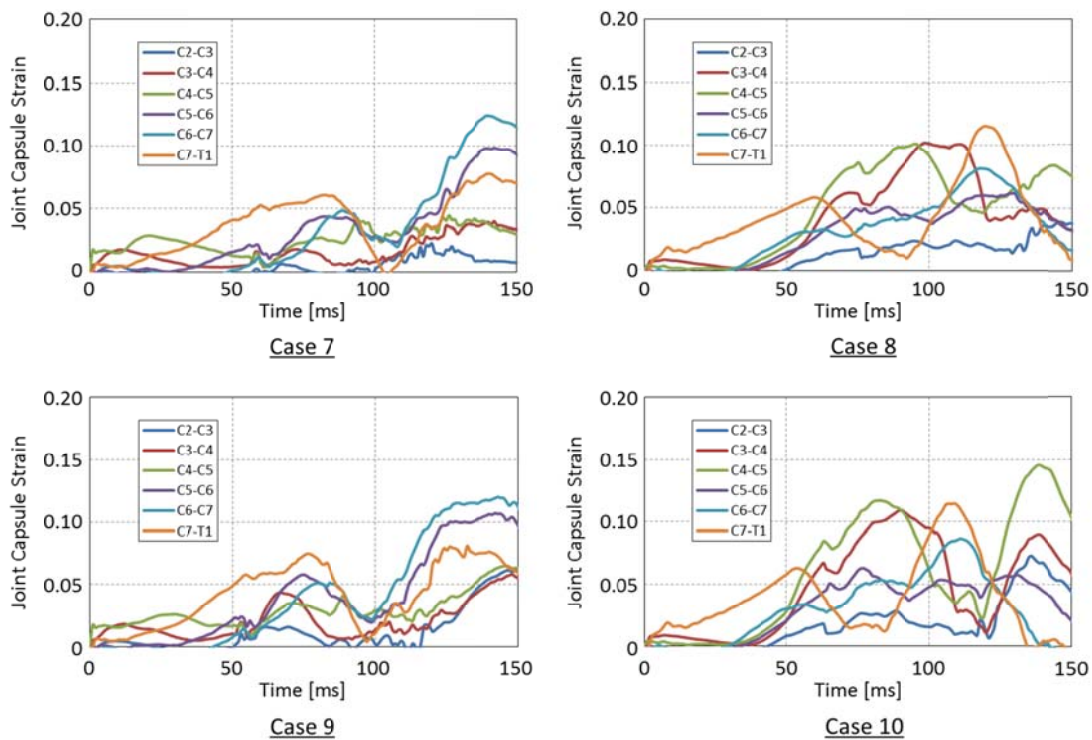


Fig. 21. Time History Curves of Joint Capsule Strains (Cases 7, 8, 9 and 10).

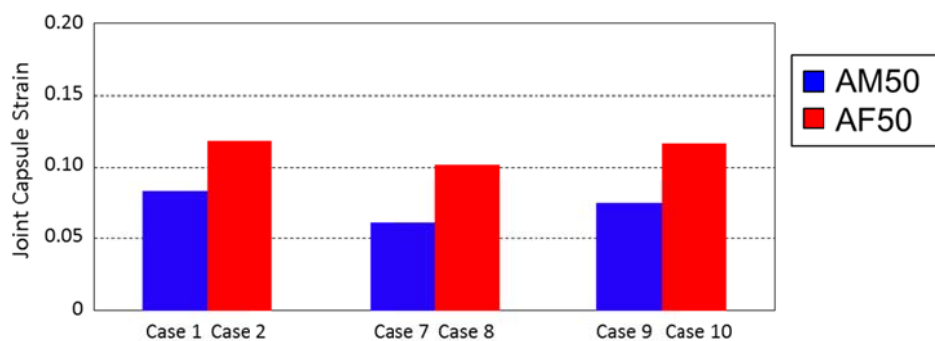


Fig. 22. 1st Peaks of Joint Capsule Strains (Cases 1, 2, 7, 8, 9 and 10).

TABLE II  
JOINT CAPSULE STRAIN – 1ST PEAK VALUE

|                | Case 1 | Case 2 | Case 3 | Case 4 | Case 5 | Case 6 | Case 7 | Case 8 | Case 9 | Case 10 |
|----------------|--------|--------|--------|--------|--------|--------|--------|--------|--------|---------|
| 1st Peak Value | 0.083  | 0.118  | 0.130  | 0.145  | 0.157  | 0.124  | 0.061  | 0.102  | 0.075  | 0.117   |
| Joint          | C7-T1  | C4-C5  | C3-C4  | C3-C4  | C3-C4  | C3-C4  | C7-T1  | C3-C4  | C7-T1  | C4-C5   |
| Time [ms]      | 82     | 93     | 67     | 99     | 103    | 79     | 82     | 98     | 77     | 83      |

#### IV. DISCUSSION

The comparison of joint capsule strains between Cases 1, 2, 3 and 4 suggests that the gender difference in head and cervical spine kinematics was due primarily to the anatomical difference rather than the difference in muscular force. The AM50 model had thicker neck than the AF50 model and its stiffness against extension was higher. When the torso was pushed forward, before the head contacted the head restraint, the AM50 neck generated torque to jerk the head. That force to the occipital condyle rotated the head, causing neck extension. On the other hand, the AF50 neck was less stiff and did not generate enough torque to jerk the head. The head did not rotate much and generated an S-shape motion of the flexible AF50 neck. The superior part of the cervical spine flexed, while the inferior part was extended. The relative rotation between adjacent vertebrae tended to be large at the inflection point. Such a difference can be found in the head and cervical spine kinematics at 75–100 ms in Fig. 13 and in the relative vertebral rotations in Fig. 14. The comparison of the time history curves of relative vertebral rotations (Fig. 14), joint capsule strain (Fig. 15) and C2-T1 distance change (Fig. 16) indicates that the joint capsule strain was caused by both extension and compression of the cervical spine. The negative peaks in C2-T1 distance change coincided well with the first peaks of joint capsule strains, while the relative vertebral rotations had gentle rises. The results suggested that the location of first strain peak was determined by cervical spine kinematics, while the peak value was given by spine straightening.

The muscle tone reduced the neck extension but compressed the neck. Reducing the neck extension lowered the joint capsule strain, while compression raised the strain. As previously shown in the component validation, joint capsule strain could grow just by compressive force owing to the incompressive synovial fluid. The strain rise was thought to be proportional to the force magnitude. That was the reason why the increase of the first strain peak was higher in AM50. The muscle tone increased the strain peak by 57% in AM50 and by 23% in AF50. As a result, the muscle tone bridged the gender difference in terms of joint capsule strain. It should be noted that the head was well supported by the head restraint in the study cases. Although the muscle force was adjusted for reducing the head rotation by 30%, the effect was limited to a small amount when the head was well supported by the head restraint during the second half of impact.

Observing the results of Cases 2, 5 and 6, it was noted that the shortest backset did not necessarily give the lowest value of the first strain peak, while the contact timing was proportional to the backset. The lowest strain peak in Case 2 was owing to the smallest relative position between the head and the torso, as shown in Fig. 19. It was not only the backset but the balance among the backset, head restraint stiffness and seatback stiffness. The backset generated the initial relative displacement between the head and T1. However, the maximum relative displacement was given by the head displacement pushing against the head restraint and the torso displacement pushing against the seatback. The AF50 model showed higher first peak values of joint capsule strain than the AM50 model regardless of the acceleration pulse. The ratios between AM50 and AF50 were close to each other. It indicates that the gender difference observed in the study was not generated by a particular impact condition but was possibly universal in low speed rear impacts.

The study has the following limitations. The gender difference indicated by the simulation model was affected by the conditions assumed in this study. For example, it was assumed that the muscular force for the AF50 model was scaled from those for the AM50 by 0.725. Although the factor was based on the ratio of muscle volumes between genders, actual difference in muscular force between genders might not necessarily appear at the same ratio because the reflex response in a rear impact could be different among individuals. Further study is necessary to better understand the cause of gender difference of WAD in actual rear impacts.

## V. CONCLUSIONS

The gender difference in head and cervical spine kinematics was investigated through rear impact simulations using the THUMS AM50 and AF50 models with the updated cervical spine model. The AF50 model showed higher first peaks of joint capsule strain than the AM50 model in all comparison cases. It was considered that the gender difference was due primarily to the anatomical difference in terms of the neck stiffness. The AF50 cervical spine was likely to exhibit an S-shape mode, while the AM50 cervical spine was extended. The first peaks of joint capsule strain coincided with the shortening of the cervical spine. The muscle tone increased the maximum strain peak in both genders, but bridged the gender difference in strain peak with the head restraint equipped. The shortest backset did not necessarily give the lowest joint capsule strain. The backset generated the initial relative displacement between the head and T1, while the maximum relative displacement was given by the head and torso displacement pushing against the head restraint and seatback. The gender difference was commonly observed under the three different acceleration pulses used in this study.

## VI. ACKNOWLEDGEMENT

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## VII. REFERENCES

- [1] Carlson, A., Siegmund, G. P., Linder, A., Svensson, M. Motion of the Head and Neck of Female and Male Volunteers in Rear Impact Car-to-Car Tests at 4 and 8 km/h. *Proceedings of IRCOBI Conference*, 2010, Hanover (Germany).
- [2] Kullgren, A., Krafft, M. Gender Analysis on Whiplash Seat Effectiveness: Results from Real-World Crashes. *Proceedings of IRCOBI Conference*, 2010, Hanover (Germany).
- [3] Kullgren, A., Stigson, H., Krafft, M. Development of Whiplash Associated Disorders for Male and Female Car Occupants in Cars Launched Since the 80s in Different Impact Directions. *Proceedings of IRCOBI Conference*, 2013, Göteborg (Sweden).
- [4] Linder, A., Olsén, S., Eriksson, J., Svensson, M., Carlsson, A. Influence of Gender, Height, Weight, Age, Seated Position and Collision Site Related to Neck Pain Symptoms in Rear End Impacts. *Proceedings of IRCOBI Conference*, 2012, Dublin (Ireland).
- [5] Carlson, A., Siegmund, G.P., Linder, A., Svensson, M. Motion of the Head and Neck of Female and Male Volunteers in Rear Impact Car-to-Car Impacts. *Traffic Injury Prevention*, 2012, 13:378–87.
- [6] Carlsson, A., Chang, F., et al. EvaRID – A 50th Percentile Female Rear Impact Finite Element Dummy Model. *Proceedings of IRCOBI Conference*, 2012, Dublin (Ireland).
- [7] Carlsson, A., Linder, A., Davidsson, J., Hell, W., Schick, S., Svensson, M. Dynamic Kinematic Responses of Female Volunteers in Rear Impacts and Comparison to Previous Male Volunteer Tests. *Traffic Injury Prevention*, 2011, 12:347–57.
- [8] Sato, F., Nakajima, T., Ono, K., Svensson, M., Brolin, K., Kaneoka, K. Dynamic Cervical Vertebral Motion of Female and Male Volunteers and Analysis of its Interaction with Head/Neck/Torso Behavior during Low-Speed Rear Impact. *Proceedings of IRCOBI Conference*, 2014, Berlin (Germany).
- [9] Yamada, H. Strength of Biological Materials. F.G. Eevan, Ed., *The Williams & Wilkins Company*, Baltimore, 1970.
- [10] Abe, H., Hayashi, K., Sato, M. Data Book on Mechanical Properties of Living Cells, Tissues and Organs. *Springer-Verlag*, Tokyo, 1996.
- [11] Watanabe, R., Katsuhara, T., Miyazaki, H., Kitagawa, Y., Yasuki, T. Research of the relationship of pedestrian injury to collision speed, car-type, impact location and pedestrian sizes using human FE model (THUMS Version 4). *Stapp Car Crash Journal*, 2012, 56:269–321.
- [12] Ng, H., Teo, E., Lee, V. Statistical Factorial Analysis on the Material Property Sensitivity of the Mechanical Responses of the C4-C6 under Compression, Anterior and Posterior Shear. *Journal of Biomechanics*, 2004, 37:771-777.
- [13] Schneider, L., Robbins, D., Pfug, M., Snyder R. Development of Anthropometrically Based Design Specifications for an Advanced Adult Anthropomorphic Dummy Family, Volume 1. University of Michigan Traffic Research Institute (UMTRI); Ann Arbor, MI; 1983-12.

- [14] Ono, K., Kaneoka, K., Wittek, A., Kajzer, J. Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics during Low Speed Rear Impacts. *Proceedings of 41st Stapp Car Crash Conference*, 1997, Florida (USA).
- [15] Iwamoto, M., Min, K., Kimpara, H. Prediction of Head-Neck Motions during Rear-End Impacts using a Human Body FE Model with Muscle Activity. *Proceedings of JSAE Spring Congress*, 2012, Yokohama (Japan).
- [16] Siegmund, G. P., Myers, B. S., Davis, M. B., Bohnet, H. F., Winkelstein, B. A. Human Cervical Motion Segment Flexibility and Facet Capsular Ligament Strain under Combined Posterior Shear, Extension and Axial Compression. *Proceedings of 44th Stapp Car Crash Conference*, 2000, Atlanta (USA).
- [17] Ono, K., Ejima, S., Suzuki, Y., Kaneoka, K., Fukushima, M., Ujihashi, S. Prediction of Neck Injury Risk Based on the Analysis of Localized Cervical Vertebral Motion of Human Volunteers during Low-Speed Rear Impacts. *Proceedings of IRCOBI Conference*, 2006, Madrid (Spain).