An Evaluation Method of Cervical Spinal Injury for Car Passenger in Dynamic Rollover Using a Human FE model

Yasuhiro Dokko
Shigeo Tobaru
Honda R&D Co., Ltd.
Japan

Kazuki Ohashi
Honda Technofort Co., Ltd.
Japan

ABSTRACT

Although the electronic stability control devices have reduced the number of dynamic rollover accidents, it still occupies non-negligible portion of the traffic accidents with fatality and severe injuries. The principal body region of fatal or severe injury in dynamic rollover is cervical spine, while there have been no recognized injury criteria of cervical spinal injuries using existing ATDs for such a loading condition. In this study, the authors tried to establish the method to evaluate cervical spinal injury of the car passengers in dynamic rollover using a human FE model. The human FE model that the authors had developed to be capable of predicting whole body kinematics and the injuries on thorax, lumbar spine, and lower extremities of car occupant in frontal and side impact was adopted as a baseline model. Since the cervical spine part of the model had been constructed by jointed rigid bodies, it could not be used to predict injury level under loading. Therefore, the model was modified to be capable of injury prediction. First, each vertebral body of the cervical spine was modified to deformable, and the deformable intervertebral disk (IVD) was inserted between each pair of vertebral bodies. Second, each isolated vertebra or IVD models were exposed to static compression in the same conditions as the experiments from the literature to find the critical stress corresponding to the fracture or rupture in the experiments. Next, the kinematics and these critical stress values were validated against the whole body inverted drop tests from the literature. Finally, the critical stress values were examined to be available in several different angles of impact in two series of head-neck drop tests from the literature. In the whole body inverted drop, the kinematics of the cervical spine was well replicated by the model and the critical stress values could well divided the impact velocities with or without injuries. In the head-neck drop with different angles of impact, the model could well predict injurious or non-injurious conditions of the tests. In addition, existing anthropomorphic test devices (ATDs) were examined if their neck structures and corresponding injury criteria could be used for evaluating cervical spinal injuries in rollover compared with the human model. It was found that there were large differences between the predicted injury by the modified human model and those by ATDs' output based on the injury assessment reference value (IARV). As a result, the human FE occupant model modified to have deformable vertebral bodies and IVDs instead of jointed rigid bodies appeared to be capable of predicting cervical spinal injury in dynamic rollover. On the other hand, it could be mentioned that further investigation on ATD neck structure and/or injury criterion is necessary to establish a physical evaluation method for occupant protection in dynamic rollover.

INTRODUCTION

As a countermeasure for occupant protection of motor vehicle rollover accident, many kinds of policies, such as improvement in wear rate of seat belt, enhancement of roof strength, and adoption of inflatable restraints for ejection mitigation by FMVSS226, have been introduced until now. Furthermore, fatality rate in rollover accident of SUV has decreased by widespread equipment of the Electrical Stability Control System in recent years. However, the rollover accidents forms about 30 percent of all the fatal accidents of passenger vehicle s in the U.S. [1], that means the reduction of fatalities in rollover accidents is still one of the big issues. The main causes of death in rollover accidents are resulted from ejection. By applying the above FMVSS 226, it is expected to have an effect on ejection mitigation at the time of a rollover accident. On the other hand, when the head is impacted by the inside of the roof of a vehicle during a rollover accident, injuries tend to occur in the head and/or neck. In such a case, the rate of occurrence of failure in the head and neck is high. Even those occupants wearing seatbelts that did not eject outside the vehicle have been injured. Although the countermeasures against these injuries are expected, criteria and dummies that evaluate the measures do not exist. Existing ATDs are the tools to evaluate injuries caused by frontal collisions or side impacts. However, injury severity levels of cervical spine caused by impact input from multiple directions like rollover accidents cannot be evaluated because there is no criterion. In this study, the evaluation method of the injury criteria of the cervical spine was examined using the Human FE model.

METHODOLOGY

The human FE model that the authors had developed to be capable of predicting whole body kinematics and the skeletal injuries on thorax, lumbar spine, and lower extremities of car occupant in frontal and side impact was adopted as a baseline model.[3]-[7] Figure 1 shows the overview of the baseline model. Since the cervical spine part of that model had been constructed by rigid bodies connected by spring elements representing intervertebral disks (IVDs), it could not be used to predict injury level by stress or strain on each element under loading. Therefore, the part of the model was modified to be capable to to that.

Figure 1 Overview of the Baseline Model

Modification of Cervical Spine Model

First, each vertebral body model of the cervical spine was modified from rigid body to deformable model with shell and solid elements for cortical and trabecular bones respectively. And the deformable solid elements for IVD was inserted between each pair of vertebral bodies. Fig.2 shows the cervical spine models of the baseline model (left) and the modified model (right).

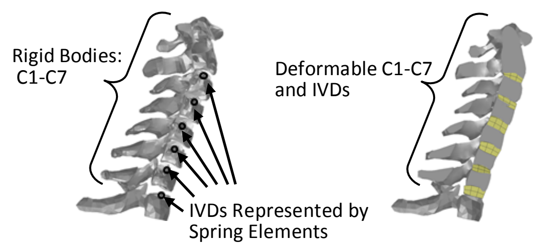


Figure 1. the cervical spine models of the baseline model (left) and the modified model (right)

From one of the anatomies[8], it was found that the cylindrical bodies have extensive cancellous interior with a thin shell of compact bone, while pedicles, articular and transverse processes are mainly compact bone. Based on this knowledge, the thickness of the shell elements of anterior and posterior surface and upper and lower endplates was set thin (0.5mm) and that of other parts including pedicles and processes was set thick (2.0mm) as shown in Fugure 2.

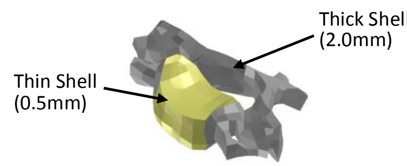


Figure 2. Assignment of Thickness for Vertebral body (C6 as an example)

All the ligaments surrounding the cervical spine were modeled by elastic membrane elements like Sato et al. [9]. Since, in this study, the focus was whether any injury occurs or not, it was thought enough to see the stress level over vertebral bodies and IVDs if more or less than those of critical levels. Therefore, it was decided that either vertebral body or IVD could be treated as of simple elastic materials. The elastic moduli of the lumbar spine model from Dokko et al.[4] were adopted for them. The explicit FE solver PAM-CRASH^{TM}[10] was used.

Determination of Critical Stress

Second**,** each isolated vertebra or IVD model was exposed to (quasi) static compression in the same conditions as the experiments from Sonoda[11] as shown in Figure 3. Sonoda[11] showed maximum forces of C3 through C7 for twenty-two Japanese PMHSs from twenties to seventies, from which averaged maximum forces of twenties through fifties were calculated. Sonoda [11] also showed an averaged maximum force of IVDs for forties through fifties. As Dokko et al. [3] did, these forces were scaled to be those for AM50%ile body size by the scale factor of 1.1, resulting in 1.21 for cross section, considering the body size of old Japanese and AM50%ile. Table 1 shows these derived numbers.

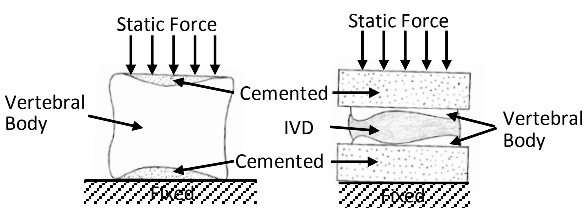


Figure 3. Static Compression of Lumbar Vertebral body (left) and IVD (right)

Table 1. Averaged Maximum Forces of C3 through C7 and IVD from Sonoda [11] and those after Scaling

	Averaged Maximum Force (N)	Scaled x1.21 (N)
C3	3,484	4,216
C4	3,680	4,453
C5	3,646	4,411
C6	3,827	4,631
C7	3,856	4,666
IVD	3,136	3,795

Under the same loading condition, the maximum stress over the shell elements corresponding to each force level was determined as the critical stress. Figure 4 shows an example for C4. Because of lack of the data for C1 and C2, average of C3 through C7 was put for them. The derived critical Stresses are shown in Table 2.

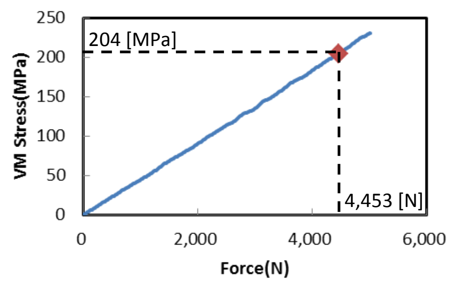


Figure 4 Determination of the Critical Stress Corresponding to the Maximum Force (Example of C4)

Table 2. Determined Critical Stress for Vertebral Bodies

#	Critical Stress [MPa]
C1	218
C2	218
C3	248
C4	204
C5	218
C6	210
C7	208

Table 3. Determined Critical Stress for IVDs

#	Critical Stress [MPa]
C2-C3	15
C3-C4	15
C4-C5	13
C5-C6	12
C6-C7	13
C7-T1	12

Validation of the Modified Model and Critical Stresses

The kinematics and the critical stress values derived above were validated against the whole body inverted drop tests from the litereature. Roberts et al.[12] performed the series of full body inverted drop tests as shown in Figure 5 to see the kinematics and injuries around the cervical spine in low (2.0 m/s) and high impact velocity (4.4 m/s) using five male PMHSs of 47 through 71 y.o. with neary AM50%ile body sizes. In addition, the four other results from Kerrigan et al.[13] of the similar condition but medium velocities were supplementaly adopted. Table 4 shows the PMHS physical information and test matrix of those two series of the tests.

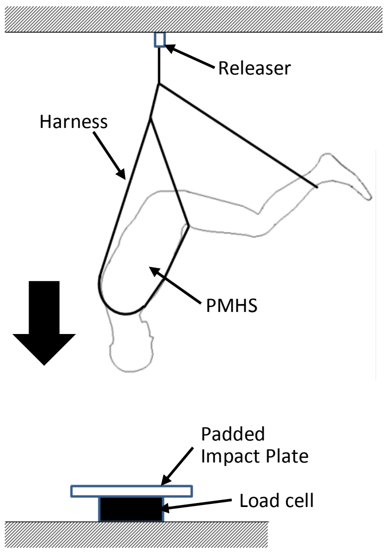


Figure 5. Full Body PMHS Inverted Drop Test

Table 4. Physical Information and Test Matrix for PMHS from Roberts et al. [12] and Kerrigan et al. [13]

Subject #	Age (y.o.)	Height (cm)	Mass (kg)	V1 (m/s)	V2 (m/s)	V3 (m/s)
582	71	178	68	4.4		
534	71	172	93	2.0	4.4	
606	62	180	51	2.0	2.0	4.4
610	48	172	62	4.4		
693	47	178	64	4.4		
516	89	155	54	3.1		
552	82	170	78	3.0		
631	71	178	69	3.6		
553	60	170	57	3.5		

Note: V1-V3 means the impact velocity in first to third tests
Upper five subjects are from Roberts et al. [12] and other four from Kerrigan et al. [13]

Finally, the critical stress values were examined to be available in several different angles of impact in two series of head-neck drop tests from Nightingale et al [14] and Toomey et al. [15]. In both series, soft tissue around cervical spine was removed, T1 was fixed into the rigid-like pot, and the mass of carriage including a load cell and a pot was set 16kg representing the effective body mass under T1. Nightingale et al. [14] performed twenty-two head/neck drop tests as shown in Figure 6 varying the conditions, i.e., rigid or padded impact surface with four different anterior/posterior angles. Because of lack of detailed information on the pad to specify to the model, only ten cases with rigid impact surface were picked. Table 5 shows the test matrix picked in this study. Impact velocity was around 3.2m/s.

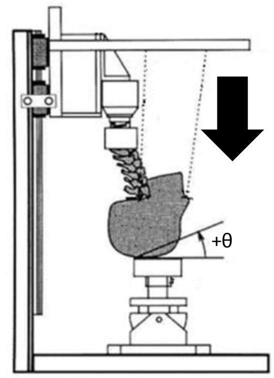


Figure 6. Test Set-up of Head-neck Drop from Nightingale et al. [14]

Table 5. Test Matrix Picked from Nightingale et al. [14] Impacting onto Rigid Surface

Test #	PMHS Age (y.o.)/Sex	Θ (deg)	Impact Velocity (m/s)
N05-R+30	36/M	+30	3.2
N18-R+15	-/M	+15	3.3
D41-R+15	69/M	+15	3.1
I32-R+15	78/M	+15	3.2
N26-R+0	65/M	0	2.4
N24-R+0	62/M	0	3.2
N22-R+0	71/M	0	3.3
N11-R-15	55/M	-15	3.1
N13-R-15	35/F	-15	3.3
UK3-R-15	62/M	-15	3.1

On the otherhand, Toomey et al. [15] performed the series of five tests in similar condition to Nightingale et al. [14] but with either lateraly angled impact surface or laterally tilted neck as shown in Figure 7. Table 7 shows the test matrix of them.

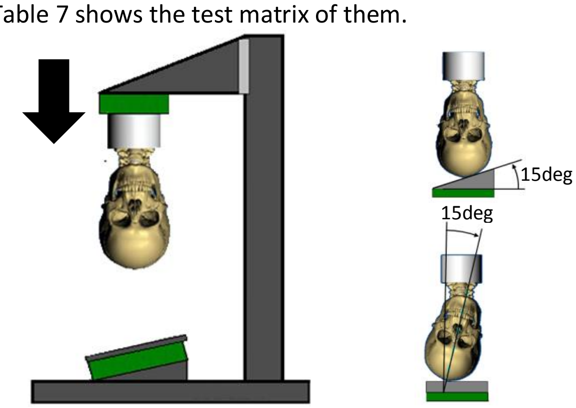


Figure 7. Test Set-up of Head-neck Drop from Toomey et al. [15] with Two Conditions; Laterally Angled Impact Surface (upper right) and Angled Neck (lower right)

Table 6 Test Matrix from Toomey et al. [15]

Test #	PMHS Age [y.o.]	PMHS Height [cm]	PMHS Mass [kg]	Impact Velocity [m/s]	Condition Laterally angled
1	76	178	80	3.0	'Surface'
2	80	193	91	3.1	'Surface'
3	77	173	73	3.3	'Surface'
4	81	183	82	2.9	'Neck'
5	56	175	80	3.3	'Neck'

Note: 'Surface' means impact onto the angled surface. 'Neck' means tilted neck.

RESULTS

Either full body inverted or head/neck drop tests were replicated by the human FE model, results of which were compared with the tests.

Full Body Inverted Drop

Figure 8 shows the human FE model replicating the full body inverted drop tests. As same as the PMHSs used in the tests, upper extremities including clavicles and scapulae were removed. Initial velocity of each case was given to the whole body. From the results, first, overall kinematics around cervical spine was checked. Figure 9 shows the comparison between the high-speed Xray image of the subject 582 from Roberts et al. [12] and the corresponding status from the human model simulation. The simulation result shows the characteristic motion of the cervical spine, i.e., extension in upper and flexion in lower portion, similar to that seen in the test. Next, to confirm the kinematics and responses from the human model representing those from the tests, head and T1 vertical accelerations and head impact force onto the impact plate were compared with those of the test results. Figure 10 through Figure 12 show them. For the tests in 2.0m/s or 4.4m/s, it was possible to develop the corridors of 1SD by three or five data sets for each condition, while, for 3.0m/s or 3.5m/s, only two data sets for each were not enough to do that. Therefore, comparisons were to those two data sets as they were for 3.0m/s and 3.5m/s. Looking at these comparisons, it was found that, in higher velocity, i.e., 3.5m/s and 4.4m/s, the peaks of T1 vertical accelerations from the simulation were higher than those of the tests. It may be caused by elastic modeling of cervical spine without fracture that makes the responsive forces increase linearly while fracture makes it drop in PMHSs, resulting in such a diffirence in T1 acceleration. In higher velocity, it was also found that the second peaks from the simulation were greater than those of the tests. In the tests, assuming the restraint by three point seat belt, the secondary strap was provided to prevent the lower body bearing on the neck, while the simulation did not consider that. It might cause such difference in the second peak of impact forces. Considering those limitation, the principal responses from the human model simulation look satisfactorily representative of those from PMHS tests.

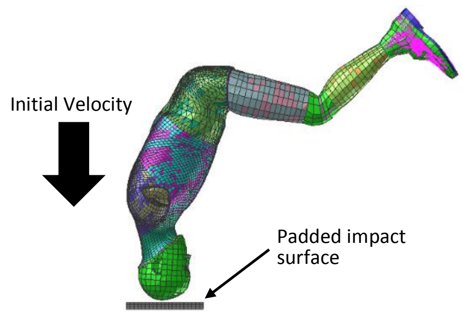


Figure 8. The Human FE Model of Full Body Inverted Drop

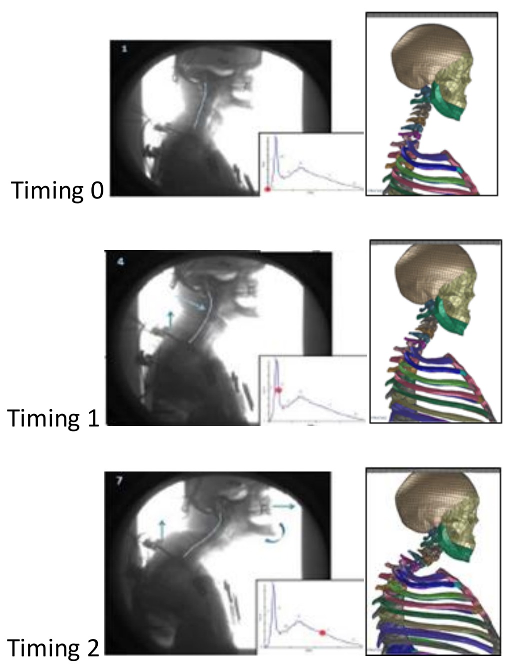


Figure 9. Comparison of Overall Kinematics around Cervical Spine between the test (left) and the Model (right)

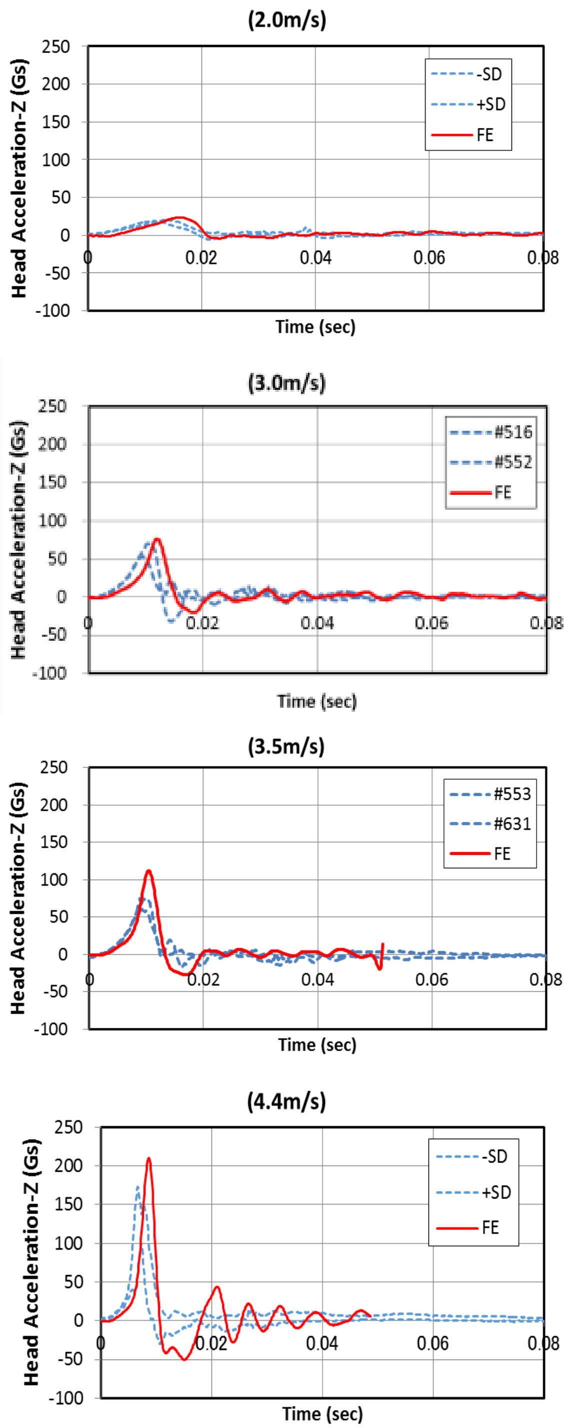


Figure 10. Comparisons of Head Vertical Acceleration between Human Model and PMHSs in Inverted Drop Test

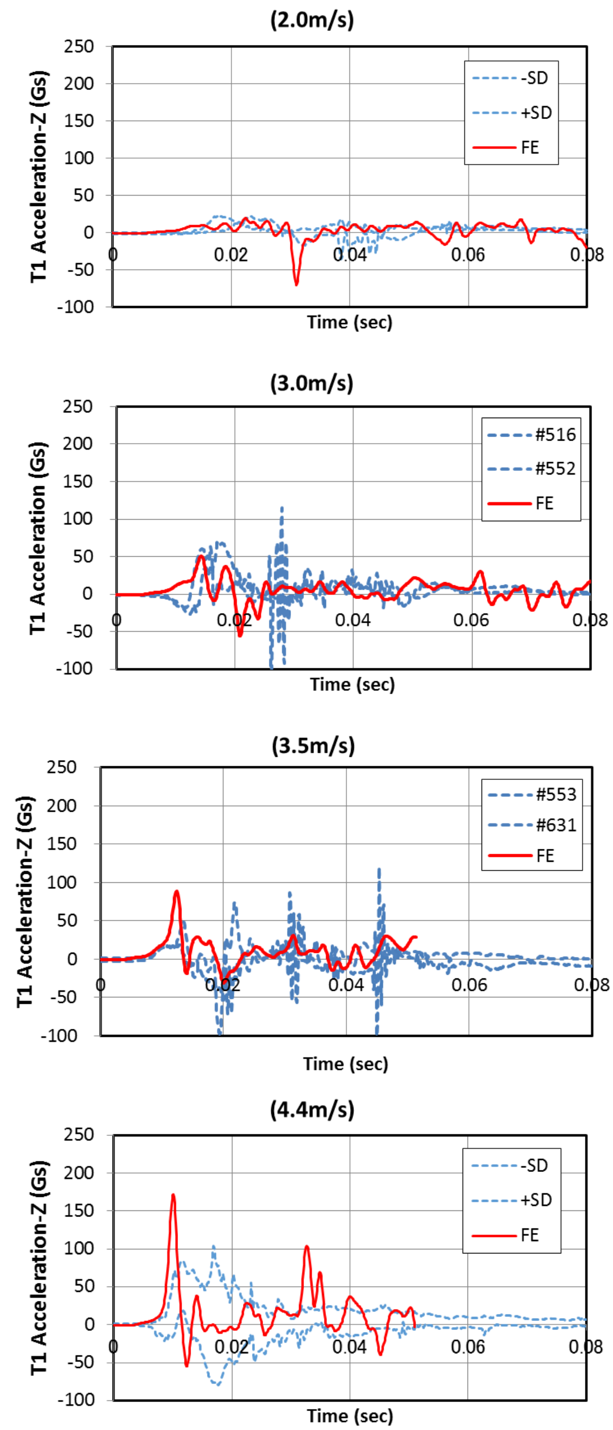


Figure 11. Comparisons of T1 Vertical Acceleration between Human Model and PMHSs in Inverted Drop Test

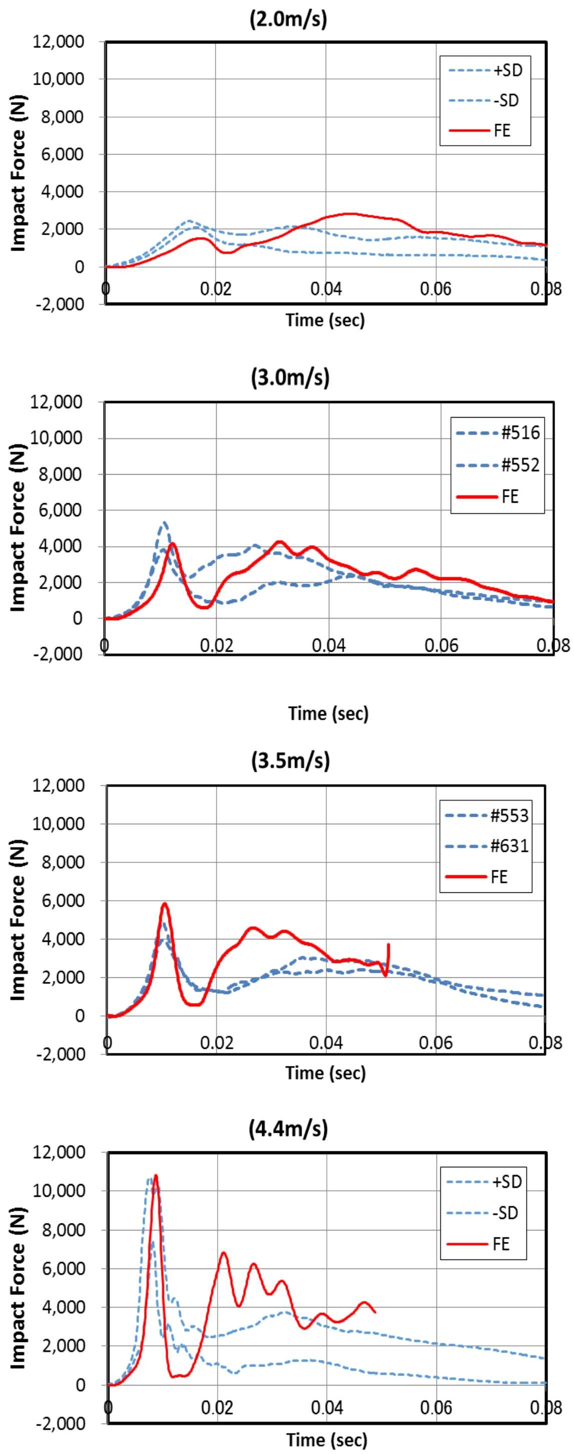
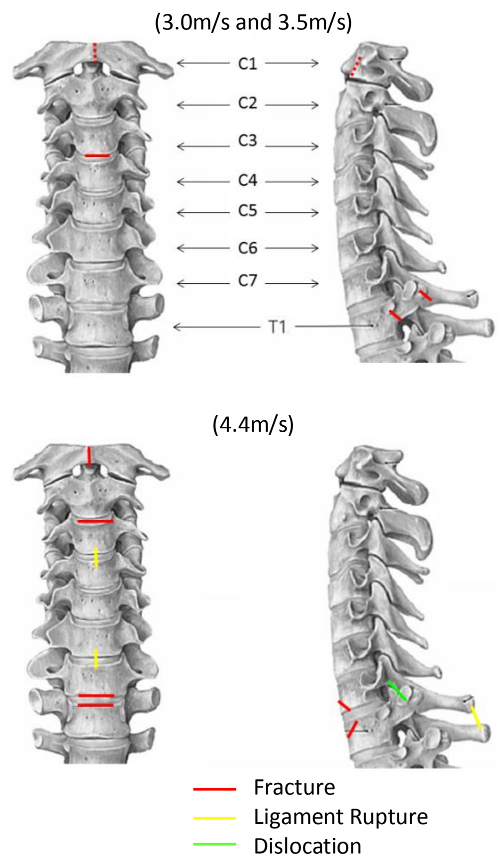
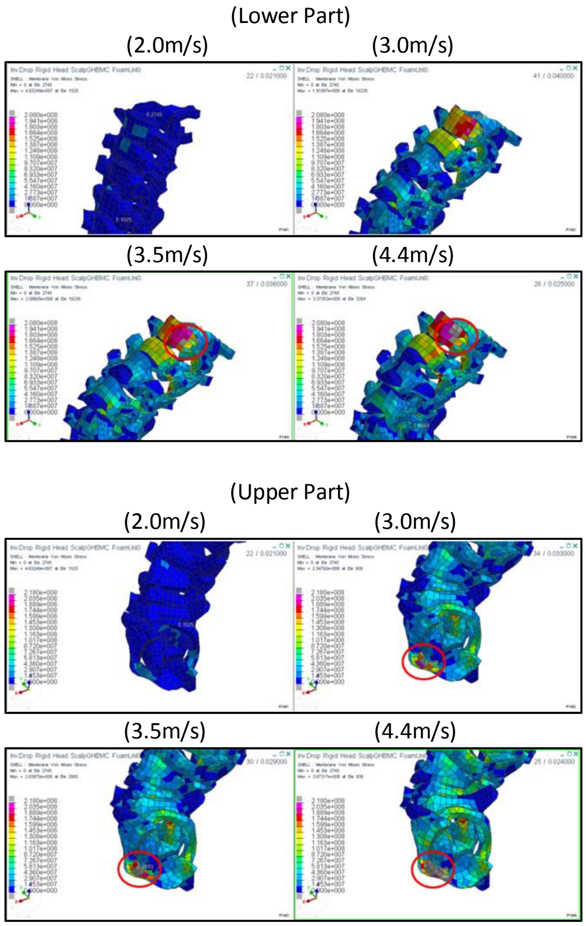


Figure 12. Comparisons of Impact Force between Human Model and PMHSs in Inverted Drop Test

For injury prediction, Roberts et al. [12] and Kerrigan et al. [13] described the injuries diagnosed after the tests, which were depicted in Figure 13. There was no injury in any tests in 2.0m/s. Maximum von-Mises stress from the human model simulation was checked if greater than the critical stress determined in the previous section for each condition as shown in Figure 14. The elements in C1 and C7 in the velocity higher than or equal to 3.0m/s showed higher von-Mises stresses than the critical stresses. Injuries in the tests and predicted from the model for each case are listed in Table 7. From them, it would be mentioned that the human model used with the critical stresses could predict the occurrence of fracture in full body inverted drop condition.



Fugure 13. Diagnosed Injuries in the Full Body Inverted Drop Tests



Circles show the elements of von-Mises stress greater than the critical stress.

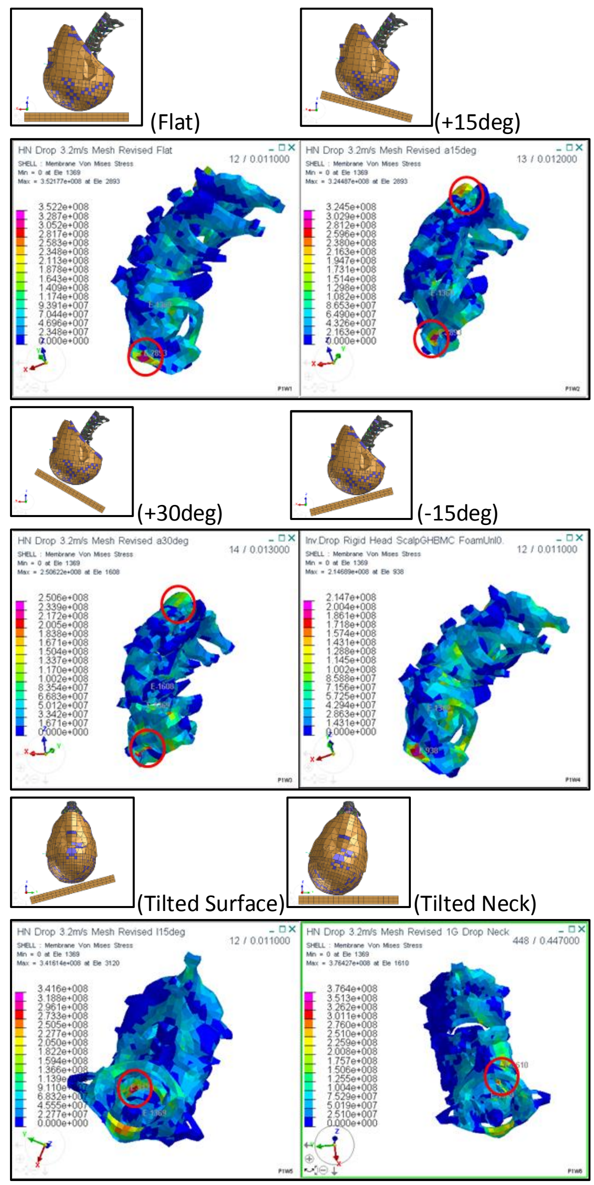
Figure 14. Predicted Fractures from the Human Model Simulation of Full Body Inverted Drop

Table 7. List of Injuries Diagnosed in the Full Body Inverted Drop Tests and those Predicted by the Human Model Simulation

Vel. (m/s)	PMHS ID	Fracture	Ligament	Dislocation	Predicted Fracture
2.0	534	-	-	-	-
	606	-	-	-	
	606	-	-	-	
3.0	516	-	-	-	C1
	552	C7	-	-	
3.5	553	C1	-	-	C1, C7
	631	C4	-	-	
4.4	534	C3, C7, T1	C7-T1	-	C1, C7
	582	C1	-	-	
	606	C6, C7	C6-7, C7-T1	-	
	610	C7	C3-4, C7-T1	C7-T1	
	693	-	C7-T1	C6-7	

Head/neck Drop

Further validation of the model with the critical stresses was tried if they were applicable for other conditions as shown in Table 5 and Table 6 including laterally asymmetric loading conditions. As same as full body inverted drop cases, predicted injuries from the simulation of each condition were compared with those described in Nightingale et al.[14] and Toomey et al.[15]. Figure 15 depicts the von-Mises stress contour at the timing of its maximum for each condition highlighting the elements with higher von-Mises stresses than the critical stresses. Injuries in the tests and predicted from the model for each case are listed in Table 8. The condition with fracture was well predicted and the tendency of more fractures at left side was represented by the model.



Circles show the elements of von-Mises stress greater than the critical stress.

Figure 15. Predicted Fractures from the Human Model Simulation of Head/neck Drop

Table 8. List of Injuries in the Head/neck Drop Tests and those predicted by the Human Model Simulation

Seriese	ID	Angle (deg)	Fracture	IVD	Ligament	Dislocation	Predicted Fracture
Nightingale et al.	N05-R+30	+30	C3	C3-4	C3-4ALL, C4-5ALL	-	C1,C2
	N18-R+15	+15	C1, C2	C2-3	C2-3ALL	C6-7	C1,C2,C3,C7 ,
	D41-R+15	+15	-	-	-	-	
	I32-R+15	+15	-	C5-6	C5-6 Cap&ALL	-	
	N26-R+0	0	-	-	-	-	C1,C3,C6,C7
	N24-R+0	0	C1, C2	-	-	-	
	N22-R+0	0	C1	-	-	-	
	N11-R-15	-15	-	-	-	-	-
	N13-R-15	-15	-	-	-	-	
	UK3-R-15	-15	-	-	-	-	
Toomey et al.	1	15(Surface)	-	-	-	-	C1, L-side:C2,C6,C7 R-side:C2,C3
	2	15(Surface)	T1	-	-	T1-T2	
	3	15(Surface)	L-side: C1, C4, C5, C6 R-side C5	-	-	-	
	4	15(Neck)	C4	-	C3-C4	-	C1, L-side:C2,C3,C4,C6,C7
	5	15(Neck)	L-side: C5, C6 R-side: C6	-	-	-	

DISCUSSION

The possibility of the prediction of cervical spine fracture by the partly modified human FE model with the determined critical stresses was aforementioned. On the other hand, in the development of motor vehicles, it is still necessary to evaluate the performance for occupant protection by physical, not on computer, test using an ATD. In this study, such two existing ATDs as Hybrid III and THOR were examined via their FE models if they were possible to be used for injury evaluation in rollover. The FE models used in this section were as follows.

- Humanetics H3-50th v8.0.1 [16]
- Humanetics THOR-50th Metric v1.3 [17]

The explicit FE solver was LS-DYNA® [18]. Both models were applied with the same loading conditions as the full body inverted drop test in previous sections as shown in Figure 16. From the results, upper neck force time histories of 2.0m/s are shown in Figure 17. Even in such a low velocity that no injury occurred in any PMHS tests as aforementioned, an upper neck force of either Hybrid III or THOR FE model indicated higher value than the IARV for neck compression force. This inconsistency should be discussed considering two points. First is the structural difference of the cervical spine between the human and ATDs. The stiff and straight-shaped cervical spine of ATDs produces higher axial force in axial loading, while less stiff and curved multi-segmented human cervical spine should ease the force by its flexibility in deformation. Second is no muscle tense in PMHS. As is well known, the IARV for neck compression, 4,000N for AM50 ATD, was based on the reconstruction of the injurious accident in tackling drill for football by Mertz et al.[19]. It is no doubt a football player is in his maximum muscle tense when he charges the target, that makes his neck stiffer than loose state, resulting in higher loading on cervical spine. At present, within the author's knowledge, it is not clear whether tense or loose state of the neck is likely to suffer cervical spinal injury in the same level of loading.

Further investigation is necessary in both points above to establish a physical evaluation method for cervical spinal injury for dynamic rollover, that is, structure of ATD and/or injury criterion.

(Hybrid III)

(THOR)

Figure 16. Hybrid III (upper) and THOR (lower) Models in Inverted Drop Condition

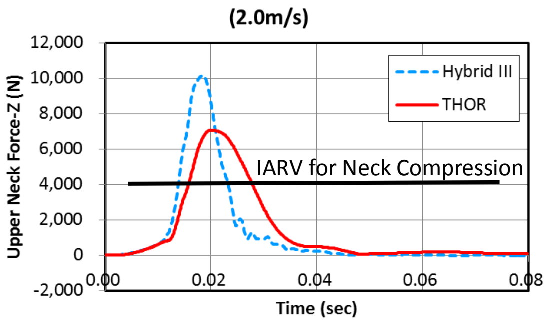


Figure 17. Upper Neck Vertical Forces from ATD Models in Inverted Drop (2.0m/s)

CONCLUSIONS

The human FE occupant model was modified to have deformable vertebral bodies and IVDs instead of jointed rigid bodies for cervical spine, which resulted in capable of predicting cervical spinal injury in dynamic rollover condition by comparing stress level among cervical spine to the determined critical stress. On the other hand, large differences were found between the injury prediction by the modified human model and ATD models in the same loading condition. It has become clear that further investigation on ATD neck structure and/or injury criterion is necessary to establish a physical evaluation method for occupant protection in dynamic rollover.

ACKNOLEDGEMENT

The authors would thank to ESI Japan for providing the material parameters of the pad tuned for the drop test results

REFERENCES

[1] Traffic Safety Fact 2014, NHTSA
[2] Ridella S. A., Eigen A, Kerrigan J., Crandall J.R. 2009. "An Analysis of Injury Typr and Distribution of Belted, Non-ejected Occupants Involved in Rollover Crashes", AAAM Vol. 53
[3] Dokko Y., Ito O., Ohashi K. 2009, "Development of Human Lower Limb and Pelvis FE Models for Adult and the Elderly", SAE 2009-01-0396
[4] Dokko Y., Kanayama Y., Ito O., Ohashi K., 2009, "Development of Human Lumbar Spine FE Models for Adult and the Elderly", SAE-01-0382
[5] Ito O., Dokko Y., Ohashi K., 2009, "Development of Adult and Elderly FE Thorax Skeletal Models", SAE 2009-01-0381
[6] Ito Y., Motozawa Y., Dokko Y., Ohashi K., Mori F., 2012, "Kinematics Validation of Age-Specific Restrained 50th Percentile Occupant FE Models in Frontal Impact", SAE 2012-01-13
[7] Dokko Y., Yanaoka T., Ohashi K., 2013, "Validation of Age-specific Human FE Models for Lateral Impact", SAE 2013-01-1242
[8] Williams P. L., 1995, "Gray's Anatomy, Thirty-Eighth Edition", Churchill Livingstone
[9] Sato F., Antona J., Ejima S., Ono K., 2010, "Influence on Cervical Vertebral Motion of the Interaction between Occupant and Head Restraint/Seat, based on the Reconstruction of Rear-End Collision Using Finite Element Human Model", 2010 IRCOBI Conference Proceedings
[10] "Explicit Solver Reference Manual, PAM-CRASH", 2009, ESI Group
[11] Sonoda T., 1962, "Studies on the Strength for Compression, Tension and Torsion of the Human Vertebral Column", Journal of Kyoto Prefectural University of Medicine, Vol. 71, pp659-702 (1953) (in Japanese)
[12] Roberts C., Kerrigan J., 2015, "Injuries and Kinematics: Response of the Cervical Spine in Inverted Impacts", ESV 15-0432
[13] Kerrigan J.R., Foster J.B., Sochor M., Forman J.L., Toczyski J., Roberts C.W., and Crandall J.R., 2014, "Axial Compression Injury Tolerance of the Cervical Spine: Initial Results", Short Communications from AAAM's 58th Annual Scientific Conference, Traffic Injury Prevention, 15:sup1, S238-S269, DOI: 10.1080/15389588.2014.956646
[14] Nightingale R.W., McElhaney J.H., Camacho D.L., 1997, "The Dynamic Response of the Cervical Spine: Buckling, End Conditions, and Tolerance in Compressive Impacts", 41st STAPP Car Crash Conference Proceedings, SAE-973344
[15] Toomey D.E., Mason M.J., Hardy W.N., Yang K.H., Kopacz J.M., 2009, "Exploring the Role of Lateral Bending Postures and Asymmetric Loading on Cervical Spine Compression Responses", Proceedings of the ASME 2009 International Mechanical Engineering Congress & Exposition, IMECE2009-12911
[16] "User's Manual, Hybrid III 50th Percentile Male Dummy, LS-DYNA Model Version 8.0.1", 2013, Humanetics Innovative Solutions, Inc.
[17] "THOR-50TH Dummy Model LS-DYNA, Release Version 1.3", 2016, Humanetics Innovative Solutions, Inc.
[18] "LS-DYNA® KEYWORD USER'S MANUAL, Version 971 R6.0.0", 2012, JSOL Corporation

[19] Mertz H.J., Hodgson V.R., Thomas L.M., Nyquist G.W., "An Assessment of Compressive Neck Loads under Injury-producing Conditions", 1978, The Physician and Sports Medicine 6