Head and Cervical Spine Responses of Post Mortem Human Subjects in Moderate Speed Rear Impacts

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Abstract The objective of this study was to obtain head and cervical spine responses of post mortem human subjects (PMHS) in moderate speed rear impacts while positioned in production seats. Instrumentation used to measure biomechanical responses of the PMHS included both accelerometers and angular rate sensors (ARS). A total of seven sled tests using seven PMHS (males 181.9 ± 3.9 cm of stature and 79.0 ± 4.7 kg of weight) were conducted in several moderate speed rear impact test conditions (FMVSS 202a, JNCAP and 10.5g, 24 km/h). Results show that the intervertebral rotations of the cervical vertebrae were relative flexion rotations although all cervical vertebrae rotated rearward in the global coordinate system. This relative flexion rotation occurred in all three moderate speed conditions and at all intervertebral levels: 4.2 ± 2.8 degrees for C2/C3, 4.0 ± 2.3 degrees for C3/C4, 6.3 ± 3.8 degrees for C4/C5, 6.1 ± 3.5 degrees for C5/C6, and 5.8 ± 3.2 degrees for C6/C7. Although the cervical flexion kinematics observed in this study in production seats are not representative of the traditional neck extension kinematics attributed to whiplash-type cervical spine injuries, results indicated that intervertebral flexion kinematics might also be regarded as an additional potential injury mechanism of the cervical spine in moderate speed rear impacts.

Keywords Cervical spine injury, whiplash, rear impacts, intervertebral kinematics

I. INTRODUCTION

Claims of cervical spine injuries (e.g. whiplash) in rear impact collisions are very common and result in enormous societal cost, with estimates on the order of $8.0 billion annually [1]. Cervical spine injuries have also been identified as a serious problem in Europe and Asia [2-3]. Societal cost in Europe resulting from this type of injury is approximated to be between 5 and 10 billion Euro per year [3]. It is reported that 50% of vehicle collisions resulted in cervical injuries in Japan [2]. Even though cervical spine injuries can result from frontal and rear impact crashes, their risk in rear impact crashes is twice that of frontal impact crashes [4].

Biomechanical responses of human subjects in rear impacts have been studied utilizing volunteers [5-12] and post-mortem human subjects (PMHS) [6, 7, 13-19]. Most of the previous studies have focused on low-speed rear impacts (ΔV less than 17 km/h) and have used a wide range of seat types, including rigid seats [6,13-18] and production seats [7-11]. However, it is important to understand the head and cervical spine responses in moderate speeds (ΔV over 17 km/h but less than 30 km/h) and in realistic test conditions (i.e. using production seats), as biomechanical data collected under moderate speeds and realistic test conditions are important to the design of safer seats and restraint systems that optimize occupant protection not only in low speeds but also in moderate to high speeds. The frequency of the Maximum Abbreviated Injury Scale (MAIS) 1 and 2+ injuries have been found to occur at moderate speeds as often as they occur at low speeds in the NASS CDS database. Moreover, when the analysis is limited to MAIS 3+ injuries, these more serious injuries are more frequent in crashes at ΔV greater than 17 km/h than they are in lower speed crashes [20]. A few studies have examined the human subject response to moderate-speed rear impact [6, 13, 14, 17, 18], although none of them utilized a modern production seat with a yielding seat back and typical padding/upholstery. In 2012, eight unembalmed PMHS were tested in two moderate speeds (8.5g and 17 km/h; 10.5g and 24 km/h) using an experimental seat designed for biofidelity evaluation of rear impact dummies and that was capable of simulating the dynamic seat back rotation of modern vehicle seat backs [20]. Additionally, an instrumentation technique for measuring the

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kinematics of each vertebra in the PMHS cervical spine was developed, validated and used to assess the cervical kinematics of the PMHS in that test series [21]. For the PMHS cervical kinematics in this experimental seat it was found that although the head and each vertebra rotated rearward in the global coordinate system, the head rotated forward with respect to T1 (i.e. neck flexion) and similar relative flexion motion was also observed for the intervertebral rotations of each pair of cervical vertebrae (e.g. C2/C3, C3/C4, …, C6/C7) at all intervertebral levels. This flexion motion is not representative of the neck extension kinematics traditionally thought to cause whiplash-type cervical spine injuries, yet whiplash-like injuries were still observed in those tests. Since it was unclear whether this behavior and resulting injuries were specific to the experimental seat, the objective of the current study was to obtain head and cervical spine responses of post mortem human subjects (PMHS) in moderate speed rear impacts while positioned in production seats.

II. METHODS

Rear impact HYGE sled tests were conducted using three different pulses (FMVSS 202a [1], JNCAP [26] and 10.5g, 24 km/h [21]). A total of seven rear impact sled tests were conducted with seven different PMHS using two different types of production seats (seat A and seat B) in the three speeds. A test matrix is provided in Table I.

TABLE I TEST SEVERITY AND TEST MATRIX

	Pulse name	Max sled acceleration (g)	ΔV (km/h)	Seat type
PMHS1	202a	8.1	16.0	Seat A
PMHS2	JNCAP	10.6	17.5	Seat A
PMHS3	JNCAP	10.8	17.8	Seat B
PMHS4	24kph	10.3	24.8	Seat B
PMHS5	JNCAP	10.7	17.8	Seat B
PMHS6	24kph	10.3	24.7	Seat A
PMHS7	24kph	10.3	24.9	Seat B

Subject selection

The PMHS used for this study were available through the Ohio State University's Body Donor Program and all applicable NHTSA and University guidelines, as well as IRB protocol, were reviewed and followed. Subjects were rejected if they exhibited severely degenerative disc, osteophytes, or previously documented spinal surgery. In order to screen osteoporotic PMHS, the PMHS were scanned using Dual Energy X-ray Absorptiometry (DXA). Seven unembalmed male subjects (74 ± 8 year old) were used in this study, and their average size was comparable to a 50th percentile male with an average weight of 79.0 ± 4.7 and height of 181.9 ± 3.9 cm. Anthropometric data of subjects' head and neck are provided in Fig. 1.

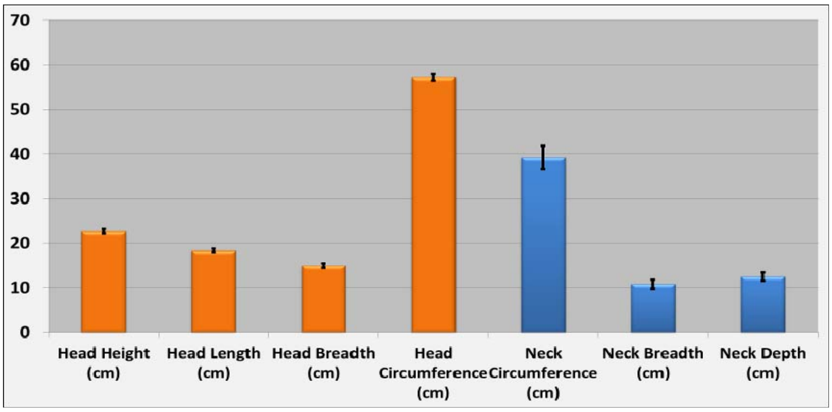


Fig. 1. Average anthropometry measurements of the head and neck (error bars indicate standard deviations)

Subject Instrumentation and sled set up

A PMHS instrumentation scheme was devised based on the instrumentation of the BioRID II ATD [27-29] so that a direct comparison of the ATD to the PMHS could be made as part of a separate biofidelity study. Instrumentation was attached at the head, cervical spine (C2-C7), T1, T8, T12 and S1, but the focus of this study is only on the head, cervical spine and T1 kinematics. For the head instrumentation, six accelerometers and three angular rate sensors (6aω) were installed on an aluminum tetrahedron fixture [23] that was screwed into the PMHS heads shown in Fig. 2A [20-22, 24]. In order to measure the intervertebral kinematics of the cervical spine, three accelerometers and three angular rate sensors (3aω) were installed on the anterolateral aspect of the cervical vertebral bodies using custom wing mounts proposed in a previous study [21] (Fig. 2B). Each instrumentation block was digitized using a FARO arm device so that responses could be properly transformed to SAE J211 coordinate systems and sign conventions. Before seating, each PMHS was dressed in a shirt and pant set made of cotton. The PMHS head was supported by a harness that was attached to a head release system using a cable (approximate 2mm diameter) that passed through a cutter device (Roberts research laboratory, Model G2) shown in Fig. 2C. Target backset (i.e. horizontal distance between the head and the head restraint) and topset (i.e. vertical distance from the center of gravity of the head to the top surface of the head restraint) were 50 ± 5 mm and 80 ± 5 mm, respectively. However, some subjects could not satisfy the backset and topset tolerances due to the nature of the PMHS spine curvature and limited adjustability of the head restraint of the production seats. The backset and topset information is provided in Table II. Strips of conductive tape were placed on both the posterior aspect of the head and the anterior surface of the head restraint to record head-to-head restraint contact times shown in Fig. 2D. A three-point belt was used to restrain the PMHS on the seats, with initial belt tensions of 17.8 N (4lb) for the lap belt and 26.7 N (6lb) for the shoulder belt. The initial seat back angle was approximately 25 degrees from the vertical (Z-axis according to SAE J211). The seat was replaced following each sled test.

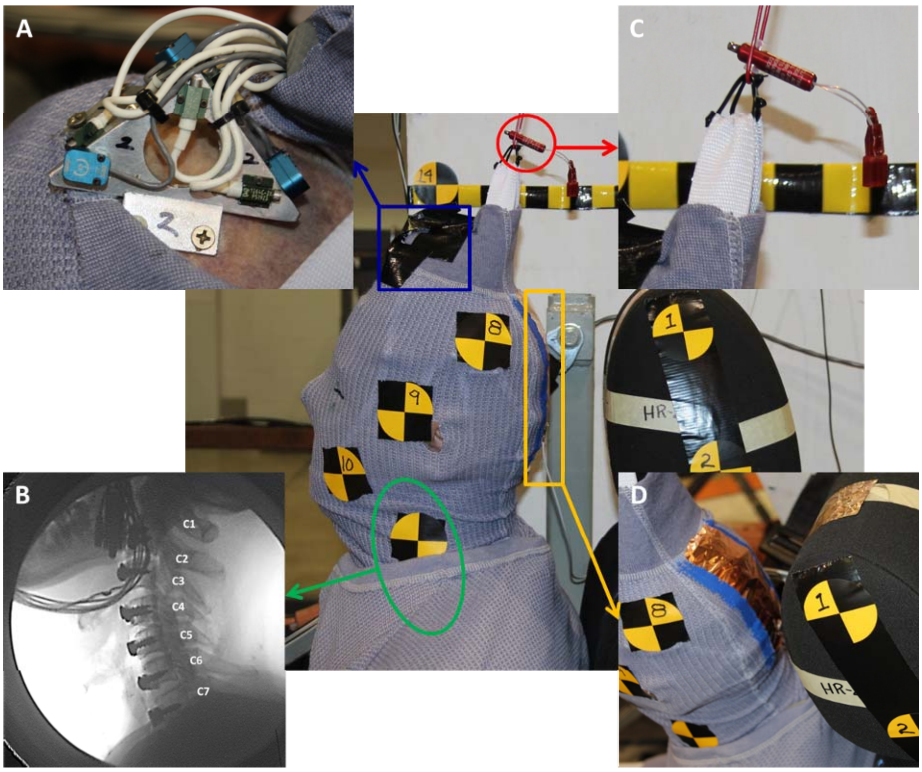


Fig. 2. PMHS instrumentation and set-up:

A: Head instrumentation using 6 accelerometers and 3 ARS on the tetrahedron fixture

B: C-arm picture for the cervical spine instrumentation installed at C2 – C7

C: Head release mechanism

D: Conductive tape for measuring head-to-head restraint contact times

TABLE II BACKSET AND TOPSET INFORMATION
unit: mm

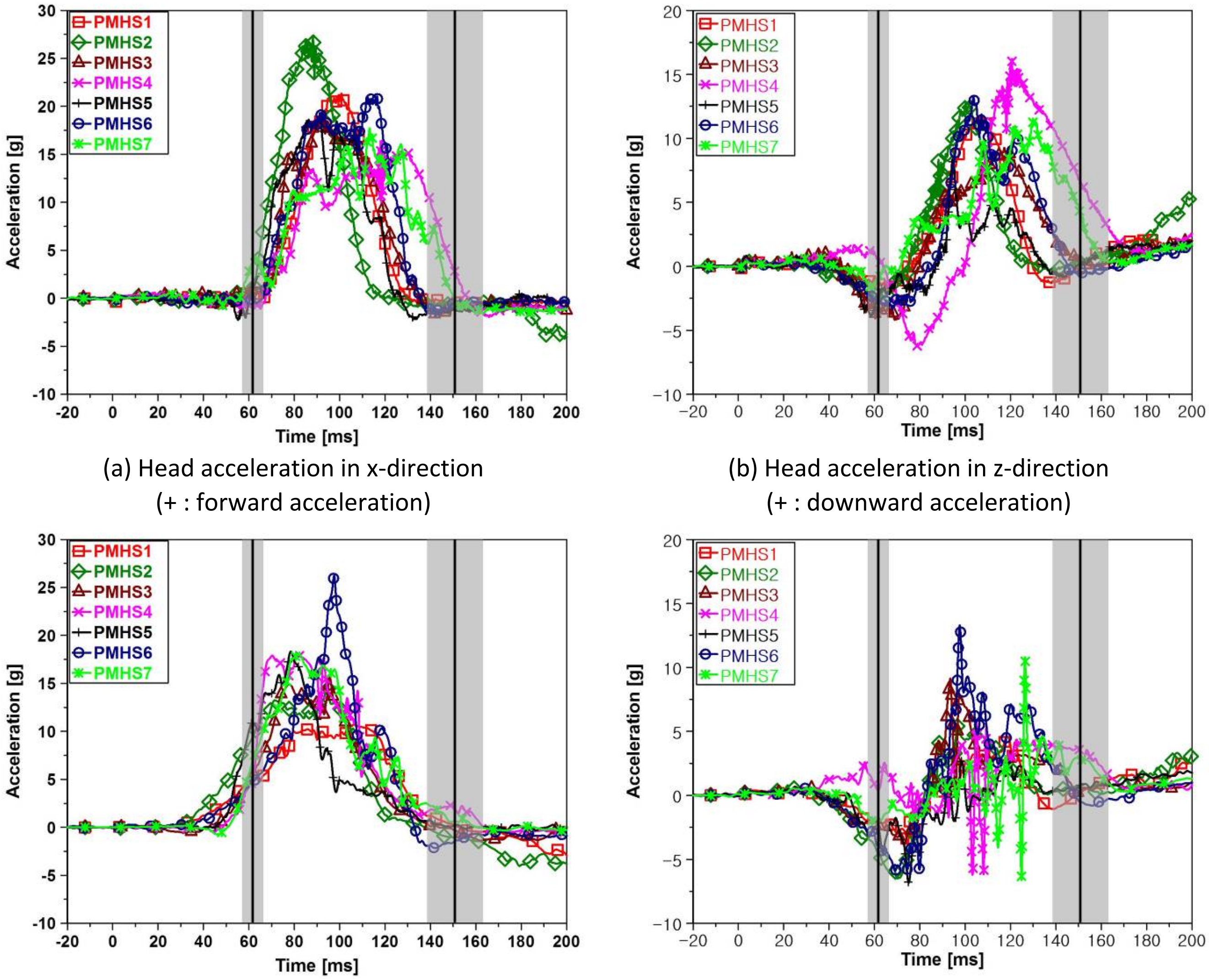
	Backset	Topset
PMHS1	55	112
PMHS2	50	97
PMHS3	75	78
PMHS4	50	65
PMHS5	55	79
PMHS6	55	81
PMHS7	52	79
Mean	56.0	84.4
Standard deviation	8.7	15.3

Data processing

The sampling frequency used in all sled tests was 12,500 Hz and all data obtained from the tests were filtered according to SAE J211. Data measured from the head instrumentation were transformed to the center of gravity (CG) in the body-fixed coordinate system that was defined by digitizing the infraorbital notches and external auditory meati (x-axis forward and z-axis downward according to SAE J211). The head CG and mass properties are presented in Table AI and AII in the appendix. The cervical and T1 instrumentation blocks were digitized to transform the data measured from the instrumentation blocks to the vertebral coordinate system located at the antero-superior edge of each cervical vertebral body shown in Fig. A1 (see appendix). The influence of gravity on the accelerometers was removed [25].

III. RESULTS

The input sled acceleration pulse and velocity for the sled tests are presented in Fig. A2 in the appendix. Whole body kinematics superimposed with cervical spine kinematics for exemplar PMHS are shown in Fig. A3, showing that the cervical curvatures were lordotic initially but became straight around 80 ms in all three speeds. Fig. A3 also shows that the cervical spine showed slightly kyphotic curvature due to the head interaction with the head restraint around 120 ms. The PMHS exhibited ramping up motion at all speeds but the highest ramping up motion and rearward head rotation were observed in the 24 km/h pulse (Fig. A3c). Head and T1 acceleration in the x and z-direction are shown in Fig. 3. Note that vertical lines and shading are included in all plots, where the first vertical solid line indicates mean head contact time with the shaded area of ± one standard deviation, while the second vertical line represents head release time from the head restraint (with shaded ± one standard deviation). More detailed information for head contact time is provided in Table AIII. The head acceleration in the x-direction was close to zero until 60 ms (Fig. 3a), while T1 began moving forward at around 40 ms (Fig. 3c), representing the common head lag phenomenon in rear impact. The head and T1 began accelerating upward around 40 ms, indicative of ramping up motion of the PMHS and spinal straightening (Fig. 3b and 3d). After head restraint contact, the head and T1 began accelerating downward due to the head interaction with the head restraint.



(a) Head acceleration in x-direction
(+ : forward acceleration)

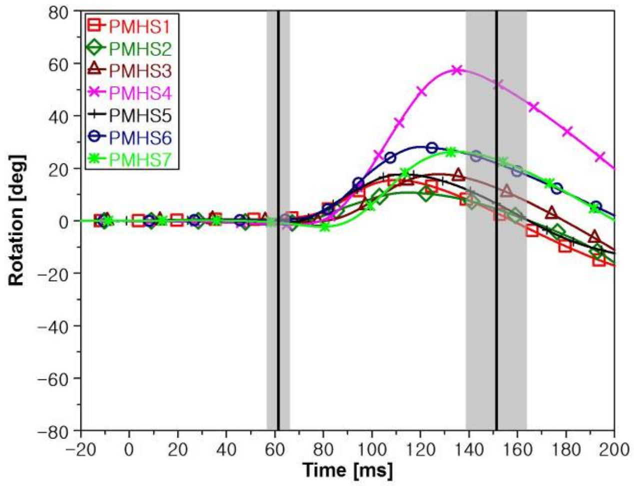
(b) Head acceleration in z-direction
(+ : downward acceleration)

(c) T1 acceleration in x-direction
(+ : forward acceleration)

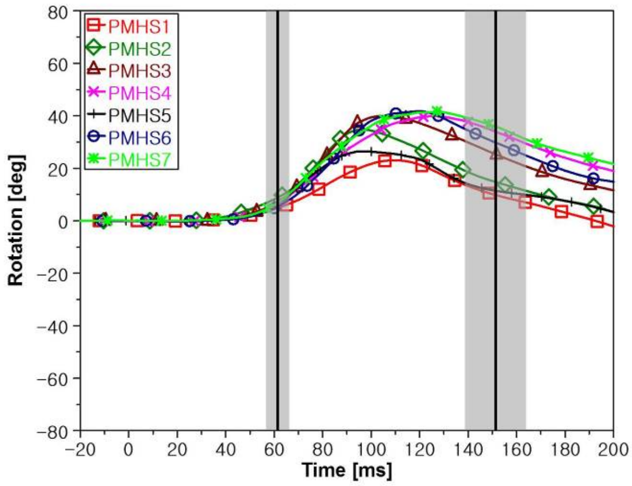
(d) T1 acceleration in z-direction
(+: downward acceleration)

Fig. 3. Head and T1 acceleration in the body coordinate system

The head and T1 rotated rearward with respect to the global coordinate system as shown in Fig. 4a and 4b, while the head rotation relative to T1 was forward rotation, indicating neck flexion during the events shown in Fig. 5. The head tended to stay close to initial position relative to the sled due to inertia, while the thorax translated and rotated rearward because it is coupled to the seat back. The lag due to inertial difference between the head and the rest of the body influenced initial neck flexion. The maximum flexion occurred during the head restraint contact, indicating that the head restraint interaction also affected neck flexion in the rear impacts. Peak acceleration and rotation for the head and T1 of each PMHS can be found in Table AIII. The average maximum rearward rotation of the head was 24.8 ± 15.6 degrees in the global coordinate system, while the head rotated forward relative to T1 (i.e. neck flexion) 22.5 ± 8.4 degrees during the event.



(a) Head rotation about the Y-axis



(b) T1 rotation about the Y-axis

Fig. 4. Head and T1 rotation in the global coordinate system
(+ : rearward rotation)

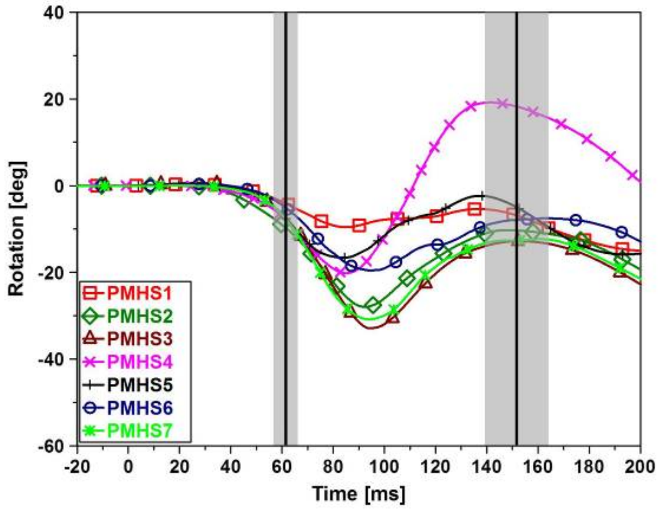
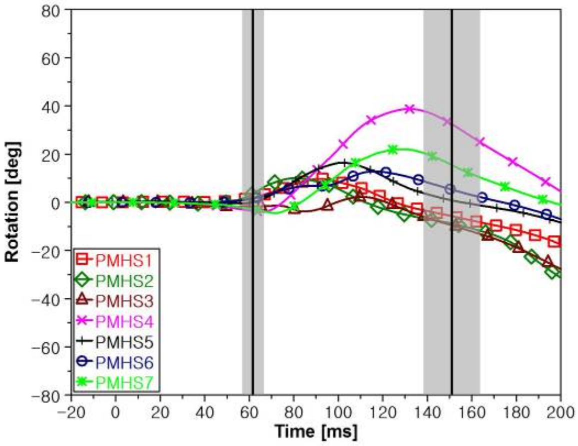
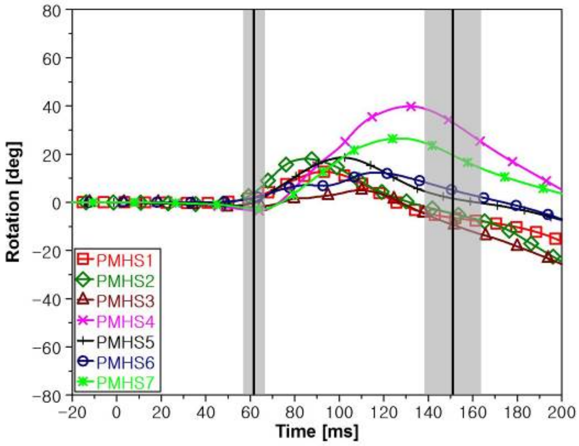


Fig. 5 Head rotation relative to T1 (+: rearward rotation represents neck extension)

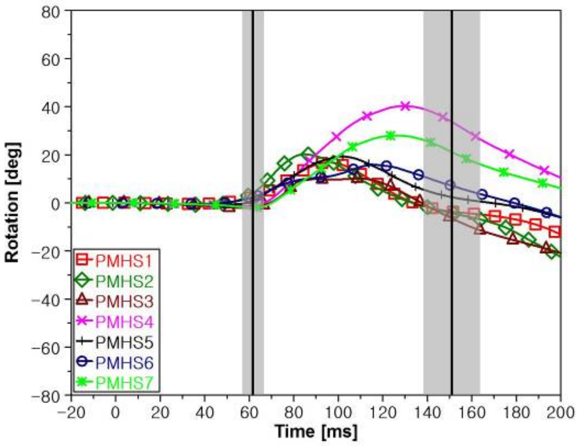
Similarly, each vertebra within the cervical spine rotated rearward in the global Y-axis (Fig. 6), but the intervertebral rotation of each vertebra relative to the vertebra below it was forward rotation (i.e. neck flexion), as shown in Fig. 7. The average maximum rearward rotations of each cervical vertebra in the global Y-axis were 16.0 ± 11.7 degrees for C2, 19.0 ± 11.3 degrees for C3, 21.5 ± 9.9 degrees for C4, 27.0 ± 8.4 degrees for C5, 32.1 ± 7.3 degrees for C6, and 36.2 ± 7.5 degrees for C7. The relative flexion rotation occurred in all three moderate speed conditions and at all intervertebral levels: 4.2 ± 2.8 degrees for C2/C3, 4.0 ± 2.3 degrees for C3/C4, 6.3 ± 3.8 degrees for C4/C5, 6.1 ± 3.5 degrees for C5/C6, 5.8 ± 3.2 degrees for C6/C7, and 2.1 ± 1.3 degrees for C7/T1. It should be noted that one of the ARS on C7 in PMHS3 was broken during the test so that C7, C6/C7 and C7/T1 rotation could not be included in Fig. 6, Fig. 7 and Table AIII.



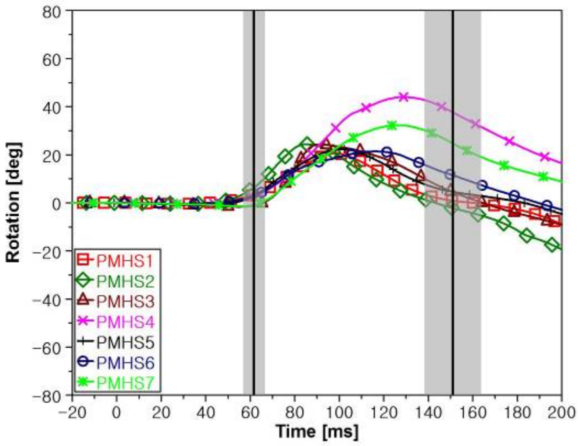
(a) C2



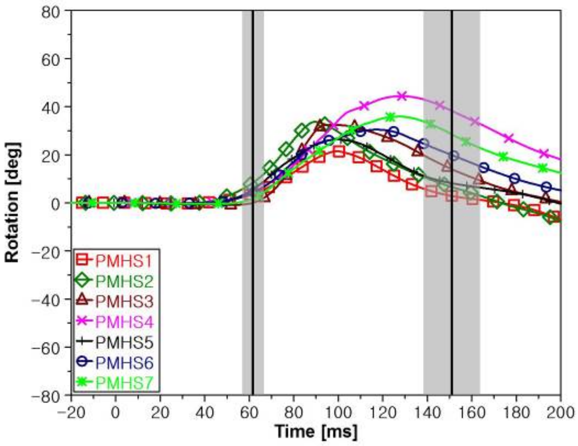
(b) C3



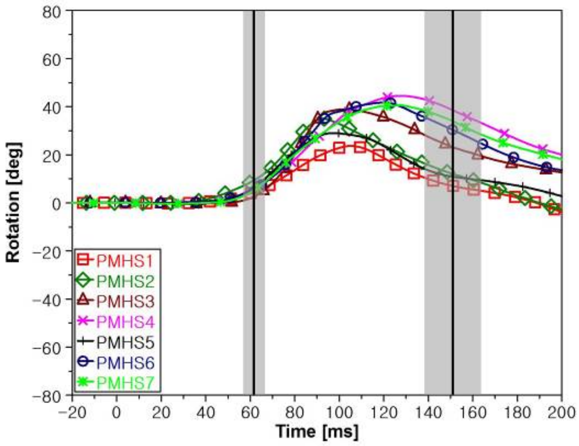
(c) C4



(d) C5



(e) C6



(f) C7

Fig. 6. Cervical rotation in the global Y-axis (+: rearward rotation represents extension)

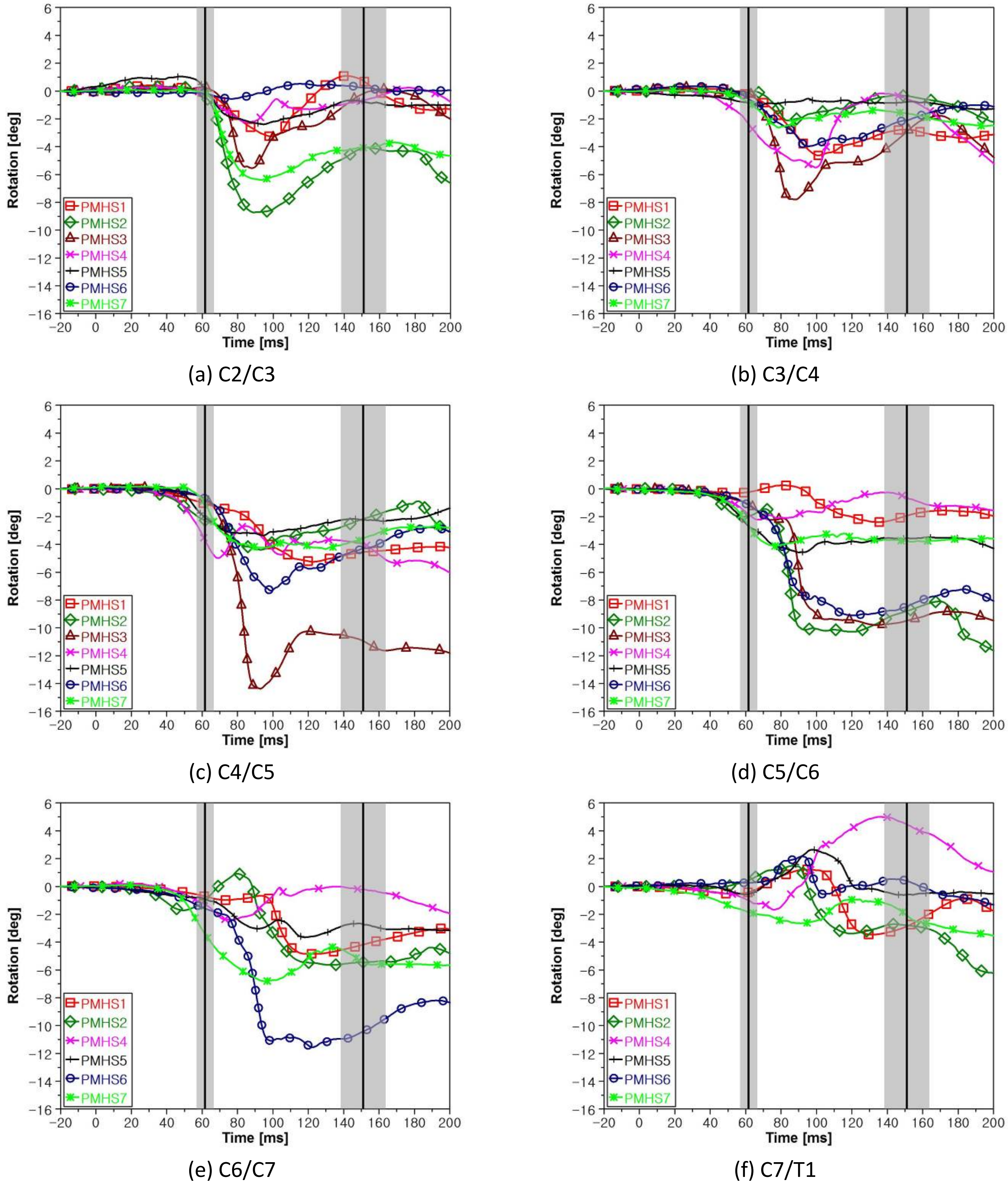


Fig. 7. Intervertebral rotation about the global Y-axis

IV. DISCUSSION

Head kinematics

The average peak head acceleration in the x-direction determined from this study was 21.3g for FMVSS 202a, 21.5g ± 4.9g for JNCAP, and 18.4 ± 2.4g for 24 km/h pulse, which were higher than previous studies using PMHS (7.3 to 10.6g in Mertz and Patrick, 1967; 10.2 to 11.7g in Kallieris et al., 1996; 6.3 to 9.7g Yoganandan et al., 2000; 4.1 to 7.4g in Bertholon et al., 2000; 6.0 to 7.0g in Deng et al., 2000) with the exception of one study (37.1 g for 17 km/h test; 68.8 for 24 km/h in Kang et al., 2012) [6,14-16,18,20]. However, a rigid seat with no head restraint was used in the moderate speed tests conducted in Mertz, Yoganandan and Kallieris studies [6, 14, 18]. The absence of a head restraint explains the low head accelerations in the x-direction. The Kang study focused on moderate speeds (17 km/h and 24 km/h) using an experimental seat with a yielding seat back with an instrumented head restraint in which load cells were installed [20]. Even though the head restraint foam and cushion covered the instrumented head restraint of the experimental seat, the frame and supporting bars of the head restraint were much more rigid than the production head restraint used in this study so the head acceleration was higher in the experimental seat. Head acceleration in the z-direction obtained in this study was 11.6g for FMVSS 202a, 10.1 ± 2.3g for JNCAP, and 13.6 ± 2.4g for 24 km/h pulse, which were similar to three studies that had a similar ΔV but with no head restraint (13.2 to 15.6g in the Mertz study, 13.2 to 15.5g in the Yoganandan study, and 12.7 to 15.3g in the Kallieris study) and one study with a head restraint (12.4 to 23.8g in the Kang study) [6,14,16,18,20]. With regard to global head rearward rotation, the results from the current study (15.5 deg for FMVSS202a, 15.4 ± 4.0 deg for JNCAP, and 37.2 ± 17.5 deg for 24 km/h pulse) were similar to that reported in the Deng (16 to 68 degree) and Kang (35.8 ± 9.8 deg in 17 km/h) studies [16, 20].

T1 kinematics

The average peak T1 acceleration in the x-direction (10.7g for FMVSS 202a, 16.3 ± 1.8g for JNCAP, 20.8 ± 4.7g for 24 m/h pulse) was similar to that measured from Yoganandan (8.5 to 13.5g) and Kang (20.3g in 17 km/h and 25.6g in 24 km/h) [18,20]. However, the average peak T1 acceleration in the z-direction (3.8g for FMVSS 202a, 5.7 ± 3.2g for JNCAP, 10.0 ± 3.8g for 24 km/h pulse) was close to the Yoganandan study (2.1 to 8.1g) but smaller than the Kang study (11.3g in 17 km/h test and 27.5g in 24 km/h test) [18,20]. T1 rotation in this study exhibited an average maximum rearward rotation of 23.0 deg for FMVSS202a, 33.7 ± 6.8 deg for JNCAP, and 41.1 ± 1.1 deg for 24 km/h pulse, somewhat between the Bertholon study (7 to 21 deg) and the Kang study (41.9 ± 6.5 deg in 17 km/h test and 64.5 ± 11.8 deg in 24 km/h test) [15, 20]. A rigid seat was used in the Bertholon study, while an experimental seat that incorporated dynamic seat back rotation was used in the Kang study [15, 20]. The different seat back dynamic properties between the rigid seat used in the Bertholon study, the experimental seat from the Kang study, and the production seat used in this study likely resulted in different rotational kinematics of T1.

Head rotation relative to T1

Hyperextension, large rearward rotation of the head relative to T1, has been considered as a likely predictor of injury in rear impact crashes. One of the current seat pass/fail criteria used in FMVSS 202a is 12 deg of rearward head rotation relative to T1 [1]. However, in this study, the average rotation of the head relative to T1 was forward in all three pulses (9.6 deg for FMVSS 202a, 25.8 ± 8.3 deg for JNCAP, and 23.5 ± 6.4 deg for 24 km/h pulse), and was forward rotation for the entire event up to 200 ms for every test except PMHS4 (24 km/h pulse). Primarily forward head rotation relative to T1 was also observed in the previous Kang study using the experimental seat [20]. Similarly, Philippens et al. (2000) reported that head lag affected the head rotation relative to T1, exhibiting 40 deg of relative forward rotation in the initial phase of the rear impact [17].

Cervical vertebrae rotations

Table III shows global rotations of the cervical vertebrae obtained in the current study compared to studies in the literature. Deng 01 and 02 represent cervical vertebral rotations from 2 PMHS tested with a head restraint in the Deng study [16], while White 01 and 02 are cervical vertebral rotations from 2 PMHS tested with a head restraint in the White study [19]. The rotations of the cervical vertebrae from the current and all previous studies in the literature were rearward in the global coordinate system as shown in Table III. Results from the FMVSS 202a and JNCAP tests in the current study were comparable to those from the White study [19]. Results from the 24 km/h tests in the current study were similar to those from the 17 km/h tests in the Kang study [20].

TABLE III COMPARISON OF GLOBAL VERTEBRAL ROTATIONS MEASURED IN THE CURRENT STUDY COMPARED TO STUDIES IN THE LITERATURE. ROTATIONS ARE REPORTED AS MEAN (SD)

	C2 (deg)	C3 (deg)	C4 (deg)	C5 (deg)	C6 (deg)	C7 (deg)
Deng 01	47.7	48.2	N/A	48.4	43.6	N/A
Deng 02	56.7	56.7	N/A	49.9	34.5	N/A
White 01	12.9	19.5	N/A	40.6	39.7	N/A
White 02	9.6	15	N/A	25.7	18.3	N/A
Kang (17 km/h)	29.2 (6.9)	32.5 (4.9)	34.8 (6.7)	38.2 (6.8)	40.9 (8.1)	42.6 (7.7)
Kang (24 km/h)	59.1 (7.0)	61.8 (6.3)	65.5 (8.2)	70.5 (10.5)	71.3 (12.7)	68.1 (12.9)
Current (202a)	9.7	12.7	16.8	20.6	21.5	23.7
Current (JNCAP)	9.6 (7.1)	13.9 (7.7)	16.5 (5.6)	23.7 (1.1)	30.8 (3.9)	34.3 (5.0)
Current (24 km/h)	24.5 (13.1)	26.2 (13.8)	28.0 (12.2)	32.6 (11.3)	36.9 (7.0)	42.3 (2.0)

Cervical intervertebral rotations

Relative intervertebral rotations have been reported using a high-speed X-ray system in low-speed rear impact conditions (5.0-9.8g and 5-16 km/h) using full body PMHS [16]. Table IV shows intervertebral rotation obtained in the current study as compared to Deng and Kang studies [16, 20]. Flexion intervertebral rotations were observed in each study shown in Table IV, but the current study and the Kang study showed intervertebral flexion rotation at all intervertebral levels (Table IV and Fig. 7). For the Deng study, relative forward rotation (i.e. flexion) was found at C2/C3 for the no head restraint condition, while C2/C3 and C3/C4 exhibited relative forward rotation for the head restraint condition [16]. It should be noted that a rigid seat was used in the Deng study [16] so there was no rotation of the seat back, while the experimental seat used in the Kang study [20] and the production seats in the current study both exhibited rotation of the seat back. The influence of a rotating seat back is evident in the different polarity at the lower cervical vertebral levels (C4/C5 and C6/C7); that is, the Deng study showed relative rearward rotation (i.e. extension), while the current study and Kang study exhibited flexion at the lower cervical vertebral levels (Table IV). In the rigid seat configuration, the upper torso and T1 are quickly coupled to the rigid seat back and move forward quickly, while the head tends to stay close to an initial position so that neck extension occurs. However, in the experimental and production seats, the upper torso compresses the seat back and induces seat back rotation, while the head stays close to an initial position, resulting in neck flexion. Additionally the head restraint stops head motion, while the upper torso keeps moving rearward with respect to the sled due to seat back rotation and deformation, which also contributes to neck flexion. Geigl et al. (1994) also reported that the plastic deformation/rotation of the seat back of a production seat caused large flexion for the relative kinematics of the head to C4 and C4 to C7 [7].

Head and neck biomechanical responses have been studied extensively in rigid seats and without head restraint, since the primary focus of the previous research was to isolate, understand and ultimately prevent hyperextension injuries. The results from those studies improved safety systems by driving the design of modern yielding seat backs and head restraints that can reduce the risk of the hyperextension injuries. However, the fact that whiplash-like injuries were observed in this study (as shown in Table AIV) and the Kang study, despite the fact that head-neck kinematics were dominated by flexion, may indicate that whiplash-type injuries can also occur in flexion and seat back rotation may have the confounding effect of increasing the risk of these neck flexion injuries in rear impact. It should be noted that the neck flexion observed in rear impact (i.e., global rearward rotation of the head and cervical spine with relative intervertebral flexion) is distinctly different than neck flexion that occurs in frontal crashes and is likely a result of a completely different injury mechanism. Future studies are underway to investigate this phenomenon further and generate injury criteria for flexion-based injuries in rear impact. Also, seat back and head restraint designs should be further investigated with a focus on optimizing safety over a wide range of rear impact speeds and with consideration to mitigating cervical spine injuries as a result of both neck extension and neck flexion.

TABLE IV COMPARISON OF RELATIVE INTERVERTEBRAL ROTATION MEASURED IN THE CURRENT STUDY COMPARED TO STUDIES IN THE LITERATURE

	Test severity	Head restraint	C2/C3	C3/C4	C4/C5	C5/C6	C6/C7
Deng et al. 2000	5-9.8g & 5-16 km/h	No	-6.0 (2.0)	8.0 (3.0)	9.0 (4.0)	9.0 (6.0)	N/A
Deng et al. 2000	5-9.8g & 5-16 km/h	Yes	-5.0 (1.0)	-7.0 (5.0)	4.0 (1.0)	9.0 (4.0)	N/A
Kang (17 km/h)	8.5g & 17 km/h	Yes	-6.3 (4.1)	-5.4 (4.1)	-8.6 (5.5)	-6.4 (4.4)	-5.3 (3.4)
Kang (24 km/h)	10.5g & 24 km/h	Yes	-6.6 (3.0)	-6.9 (4.3)	-11.7 (6.0)	-8.7 (5.0)	-8.0 (4.2)
Current (202a)	8.1g & 16.0km/h	Yes	-3.3	-4.6	-5.2	-2.4	-4.8
Current (JNCAP)	10.7g & 17.7 km/h	Yes	-5.5 (3.2)	-3.7 (3.6)	-7.4 (6.1)	-8.2 (3.2)	-4.6 (1.4)
Current (24 km/h)	10.3g & 24.8 km/h	Yes	-3.1 (3.0)	-4.0 (1.4)	-5.7 (1.5)	-5.1 (3.6)	-6.9 (4.6)

PMHS Injuries

All PMHS injuries found in this study are documented in Table AIV. None of these injuries could be detected by imaging or palpation, and could only be identified in autopsy. These injuries do not represent physically diagnosable injuries in a live occupant after a rear impact crash, but likely represent the occult minor whiplash-type injuries that can cause neck pain due to soft tissue damage. The injuries were very similar as those found in the Kang study [20], in which large forward head rotation relative to T1 was also recorded.

Limitations

A total of 7 rear impact sled tests were conducted using 7 PMHS. Since rear impact head and cervical spine responses from PMHS are limited in the literature, this study should help to better understand head and cervical spine responses in rear impacts. Since our target speeds were more severe than low speed tests that can utilize volunteers, PMHS were chosen as test specimens. PMHS cannot account for neck muscle activation so some caution needs to be taken when the data obtained from this study are applied to the real world. However, at the speeds considered in this study it has been shown that the effect of the lack of muscle activation is minimal [20], particularly because the data are intended to be applicable to a live unaware occupant (which is the basis for the design of existing rear impact dummies).

The PMHS used in this study, as in most PMHS studies, were elderly (over 70 years old). Even though these PMHS were screened to ensure they had acceptable Bone Mineral Density (T-score > -2.5), it is possible that age-related differences in the properties of the cervical spine had an influence on the cervical spine kinematics as well as the potential for intervertebral subluxations to occur.

Subject-to-subject variation is naturally part of every PMHS test series. For the cervical spine in particular, subjects have different neck lengths, intervertebral disc heights, spinal curvatures, and varying levels of initial local degeneration, all of which could contribute to differences in kinematics and injury potential. There is no clear way to control or adjust for all of these factors, so these differences likely manifest as between-subject variance in kinematic response or injury. Some of this variance was accounted for by attempting to control the backset (50 ± 5 mm) and topset (80 ± 5 mm) for each PMHS, but even this was not always possible due to limited adjustability of the head restraints.

Three accelerometers and three ARS were attached to each cervical vertebra. Even though this instrumentation technique provided valuable intervertebral kinematic information, adding mass to the vertebrae could change the nature of the head and cervical spine responses. The mass of the instrumentation including cable and mounts was 32 grams. Bertholon et al. (2000) reported that their cervical vertebral instrumentation with a weight of 50 grams was about 25% of the mass of one vertebral level (i.e. vertebra with soft tissues) [15]. Based on this, the proposed instrumentation herein was about 15% of the mass of one vertebral level, and even less if the effective mass of each vertebral level is considered due to the resistance to motion of the vertebral connections during these moderate rear impacts. However, future work will be carried out to investigate this mass effect on the head and cervical spine responses.

V. CONCLUSIONS

The intervertebral rotations of the cervical vertebrae were relative flexion rotations although all cervical vertebrae rotated rearward in the global coordinate system. This relative flexion rotation occurred in all three moderate speed conditions and at all intervertebral levels. The cervical flexion kinematics observed in this study in production seats are not representative of the traditional neck extension injuries, but results from this study indicated that intervertebral flexion kinematics may possibly be regarded as an additional potential injury mechanism of the cervical spine in moderate speed rear impacts.

VI. ACKNOWLEDGEMENT

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VIII. APPENDIX

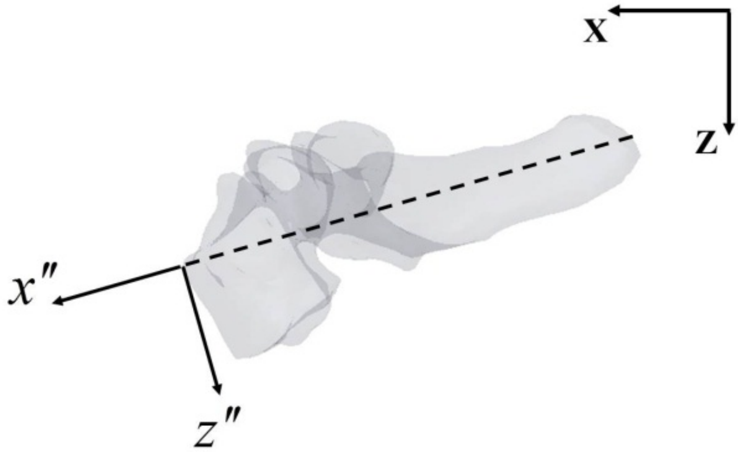
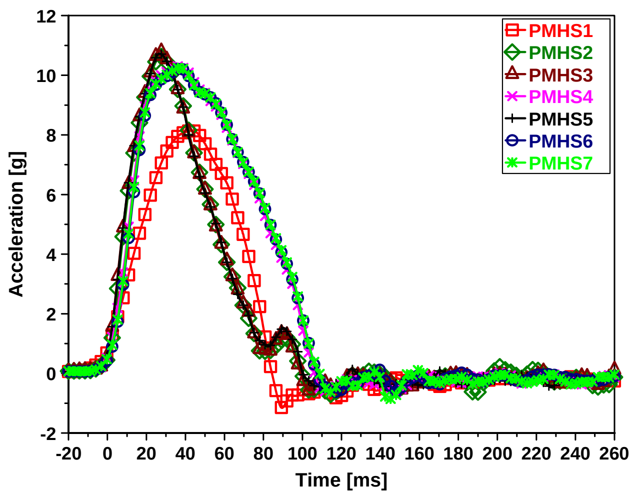
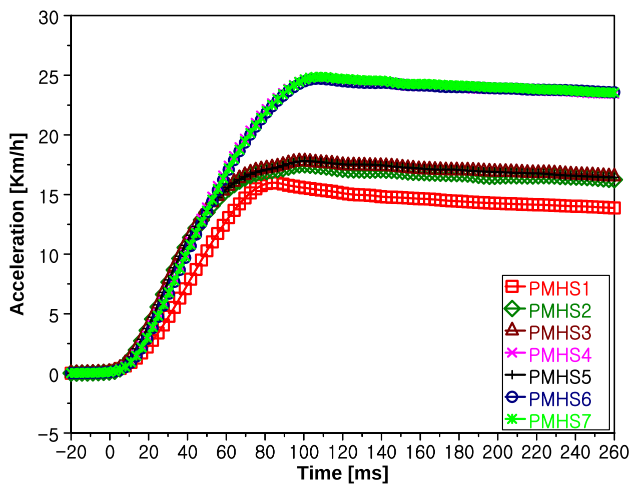


Fig. A1. Anatomical coordinate system of the cervical vertebrae and T1

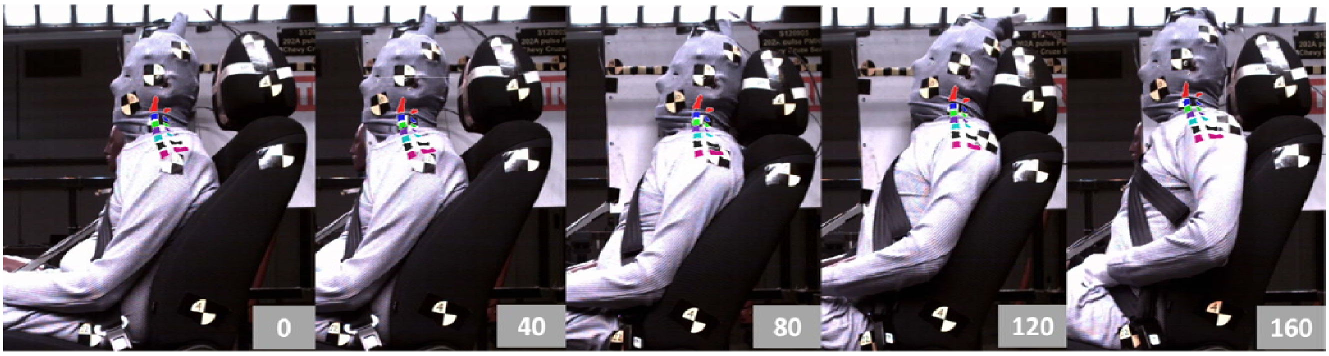


(a) sled acceleration

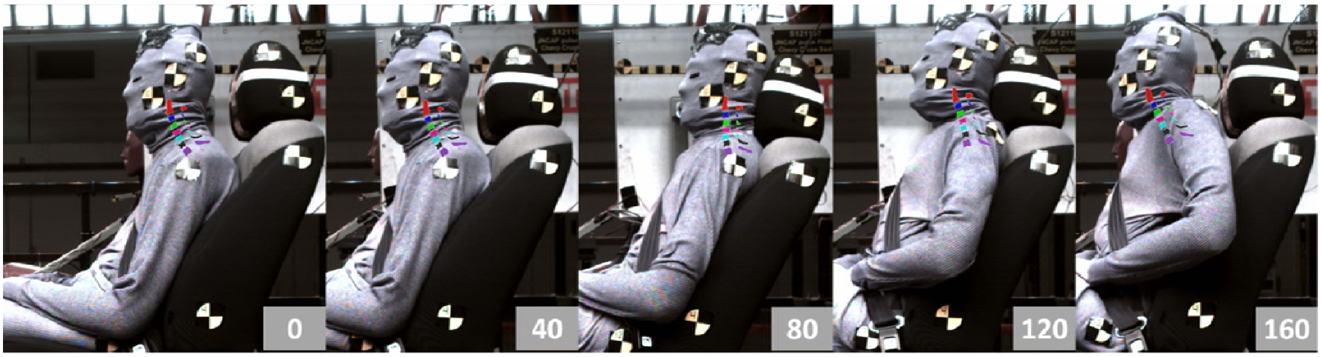


(b) sled velocity

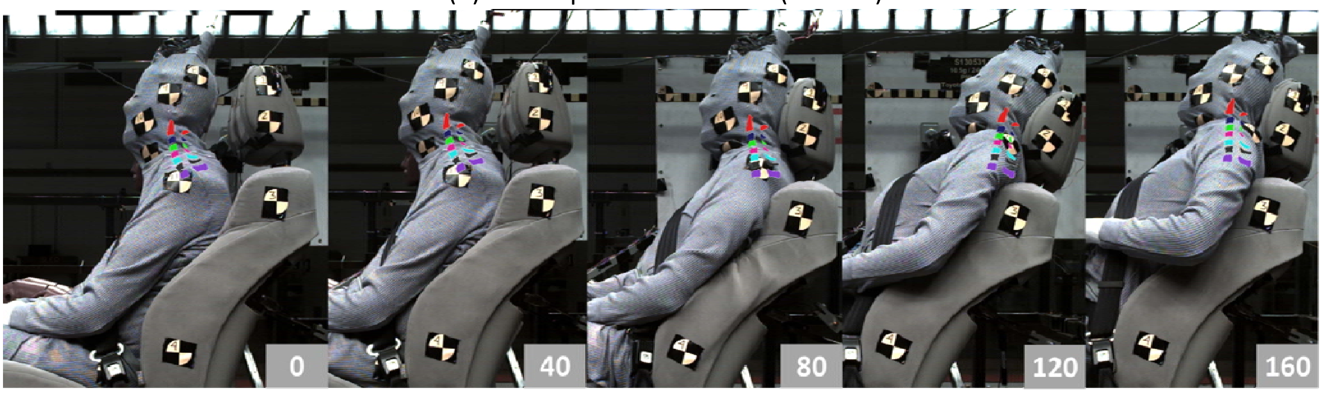
Fig. A2. Sled pulses for FMVSS 202a, JNCAP, and 24 km/h



(a) FMVSS 202a pulse with seat A (PMHS1)



(b) JNCAP pulse with seat A (PMHS2)



(c) 24 kph pulse with seat B (PMHS7)

Fig. A3. Sequential images of the PMHS overall kinematics and cervical spine kinematics (0 – 160 ms)

TABLE AI HEAD WEIGHT AND MASS MOMENTS OF INERTIA
UNIT: KG AND KG·M^2

	PMHS1	PMHS2	PMHS3	PMHS4	PMHS5	PMHS6	PMHS7	Mean	SD
Head Wt	4.23	4.14	4.01	3.75	3.77	3.83	4.00	3.96	0.19
Head Ixx	0.0187	0.0195	0.0203	0.0187	0.0192	0.0191	0.0130	0.0184	0.0024
Head Iyy	0.0265	0.0236	0.0198	0.0176	0.0178	0.0180	0.0190	0.0203	0.0034
Head Izz	0.0150	0.0154	0.0134	0.0124	0.0125	0.0131	0.0179	0.0142	0.0020

TABLE AII MEASURED HEAD CG AND OC LOCATION (RELATIVE TO ORIGIN OF HEAD COORDINATE SYSTEM – FRANKFORT PLANE)
UNIT: CM

	PMHS1	PMHS2	PMHS3	PMHS4	PMHS5	PMHS6	PMHS7	Mean	SD
Head CGx	1.52	0.6	1.11	0.9	-0.68	0.29	0.43	0.60	0.70
Head CGz	-2.53	-2.8	-2.89	-2.19	-2.05	-2.45	-2.47	-2.48	0.30
Head OCx	-1.26	-0.75	-1.75	-1.38	-1.52	-1.13	-1.03	-1.26	0.33
Head OCz	1.83	1.62	1.89	3.35	4.01	1.8	3.18	2.53	0.96

TABLE AIII PEAK VALUES FOR KINEMATICS
unit: ms for contact time, g for acceleration, and degree for rotation

	PMHS 1	PMHS 2	PMHS 3	PMHS 4	PMHS 5	PMHS 6	PMHS 7	All Speeds		JNCAP		24kph	
	FMVSS 202	JNCAP	JNCAP	24kph	JNCAP	24kph	24kph	Mean	SD	Mean	SD	Mean	SD
Head													
contact time start	66.4	61.9	66.6	63.3	52.5	64.8	55.0	61.5	5.6	60.3	7.2	61.0	5.3
contact time end	149.7	135.1	150.9	170.7	136.4	147.9	165.0	150.8	13.3	140.8	8.8	161.2	11.9
acceleration x (+)	21.3	27.1	18.5	16.4	18.8	21.0	17.7	20.1	3.5	21.5	4.9	18.4	2.4
acceleration z (-)	-1.8	-2.7	-4.2	-6.3	-3.9	-3.4	-1.7	-3.4	1.6	-3.6	0.8	-3.8	2.3
acceleration z (+)	11.6	12.7	9.3	16.2	8.4	13.2	11.4	11.8	2.6	10.1	2.3	13.6	2.4
rotation y (+)	15.5	10.7	17.7	57.4	17.7	28.1	26.2	24.8	15.6	15.4	4.0	37.2	17.5
T1													
acceleration x (+)	10.7	15.0	15.6	18.1	18.3	26.2	18.0	17.4	4.7	16.3	1.8	20.8	4.7
acceleration z (-)	-3.8	-6.5	-3.8	-6.2	-6.8	-5.9	-6.3	-5.6	1.3	-5.7	1.6	-6.1	0.2
acceleration z (+)	4.6	5.4	9.1	5.9	2.7	13.3	10.8	7.4	3.8	5.7	3.2	10.0	3.8
rotation y (+)	23.0	34.7	39.9	39.8	26.4	41.7	41.7	35.3	7.7	33.7	6.8	41.1	1.1
Head-to-T1													
rotation y (-)	-9.6	-27.9	-32.9	-20.2	-16.6	-19.5	-30.8	-22.5	8.4	-25.8	8.3	-23.5	6.4
rotation y (+)	0.4	0.0	0.7	19.2	0.2	0.4	0.0	3.0	7.1	0.3	0.3	6.5	11.0
Cervical rotation													
C2 rotation y (+)	9.7	10.0	2.2	38.7	16.4	12.8	22.0	16.0	11.7	9.6	7.1	24.5	13.1
C3 rotation y (+)	12.7	18.1	5.0	39.9	18.6	12.3	26.5	19.0	11.3	13.9	7.7	26.2	13.8
C4 rotation y (+)	16.8	20.2	10.1	40.2	19.3	15.8	28.0	21.5	9.9	16.5	5.6	28.0	12.2
C5 rotation y (+)	20.6	24.4	24.3	44.0	22.4	21.4	32.3	27.0	8.4	23.7	1.1	32.6	11.3
C6 rotation y (+)	21.5	33.7	32.4	44.4	26.3	30.4	35.9	32.1	7.3	30.8	3.9	36.9	7.0
C7 rotation y (+)	23.7	35.1	38.9	44.5	29.0	41.8	40.6	36.2	7.5	34.3	5.0	42.3	2.0
Intervertebral rotation													
C2C3 rotation y (-)	-3.3	-8.7	-5.5	-2.3	-2.4	-0.6	-6.4	-4.2	2.8	-5.5	3.2	-3.1	3.0
C2C3 rotation y (+)	1.1	0.3	0.3	0.3	1.0	0.5	0.1	0.5	0.4	0.5	0.4	0.3	0.2
C3C4 rotation y (-)	-4.6	-2.1	-7.8	-5.5	-1.1	-4.0	-2.6	-4.0	2.3	-3.7	3.6	-4.0	1.4
C3C4 rotation y (+)	0.2	0.1	0.3	0.1	0.0	0.3	0.1	0.2	0.1	0.1	0.2	0.2	0.2
C4C5 rotation y (-)	-5.2	-4.4	-14.4	-5.3	-3.4	-7.3	-4.3	-6.3	3.7	-7.4	6.1	-5.7	1.5
C4C5 rotation y (+)	0.0	0.0	0.1	0.0	0.3	0.0	0.2	0.1	0.1	0.1	0.1	0.1	0.1
C5C6 rotation y (-)	-2.4	-10.3	-9.8	-2.2	-4.6	-9.1	-4.1	-6.1	3.5	-8.2	3.2	-5.1	3.6
C5C6 rotation y (+)	0.2	0.0	0.0	0.0	0.0	0.0	0.0	0.1	0.1	0.0	0.0	0.0	0.0
C6C7 rotation y (-)	-4.8	-5.6	N/A	-2.4	-3.7	-11.6	-6.8	-5.8	3.2	-4.6	1.4	-6.9	4.6
C6C7 rotation y (+)	0.0	0.9	N/A	0.2	0.0	0.0	0.1	0.2	0.4	0.5	0.6	0.1	0.1
C7T1 rotation y (-)	-3.5	-3.4	N/A	-1.7	-0.6	-0.5	-2.7	-2.1	1.3	-2.0	2.0	-1.6	1.1
C7T1 rotation y (+)	1.2	1.5	N/A	5.0	2.6	2.2	0.0	2.1	1.7	2.1	0.8	2.4	2.5

TABLE AIV DOCUMENTATION OF INJURIES

PMHS#	Injury Description
PMHS1	Subluxation at C4/C5 - right facet joint/capsule
PMHS2	Subluxation at C5/C6 - interspinous ligaments, ligamentum flavum, facet joints/capsules, disc Subluxation at C6/C7 - disc
PMHS3	Subluxation at C4/C5 - interspinous ligaments, ligamentum flavum, facet joints/capsules, disc Subluxation at C5/C6 - interspinous ligaments, disc
PMHS4	Subluxation at C3/C4 and C4/C5 - interspinous ligaments, ligamentum flavum, facet joints/capsules, disc, anterior longitudinal ligament Subluxation at C7/T1 - disc
PMHS5	Subluxation at C6/C7 - interspinous ligaments, ligamentum flavum, facet joints/capsules, disc
PMHS6	Subluxation at C5/C6 and C6/C7 - interspinous ligaments, ligamentum flavum, facet joints/capsules, disc
PMHS7	Subluxation at C2/C3, C6C7 and C7/T1 - interspinous ligaments, ligamentum flavum, facet joints/capsules, disc