

HEAD LINEAR AND ROTATIONAL ACCELERATIONS AND CRANIOCERVICAL LOADS IN LATERAL IMPACT

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ABSTRACT

The objective of the study was to determine lateral impact-induced three-dimensional temporal head center of gravity linear and angular accelerations, and craniocervical forces and moments from post mortem human subject (PMHS) sled tests and compare with the European side impact dummy, ES-II, responses. Anthropometrical data were obtained and x-rays were taken. Specimens were seated on a sled, restrained using belts, and lateral impact acceleration was applied. Injuries to craniocervical structures were identified using pre- and posttest radiography and computed tomography. A similar testing protocol with the exception of biological evaluations was adopted for ES-II dummy tests. Lateral head linear accelerations were lower in the dummy than PMHS. Head rotational accelerations were suggestive of mild traumatic brain injury with potential for loss of consciousness during in vivo situations. Forces and moments at the craniocervical junction were lower in the dummy than PMHS with injuries confining to soft tissue structures. Relatively severe lateral impact pulse produces minor craniocervical trauma.

Key Words: biomechanics, injuries, side impact, head and neck,

BIOMECHANICAL RESEARCH studies using post mortem human subjects (PMHS) are often used to determine mechanisms of load transfer, injuries, and associated metrics for response quantifications. As applied to the lateral impact mode, previous studies using sled equipment have focused primarily on chest and pelvis injuries, mechanisms, tolerances, and comparison with side impact dummies such as SID, ES-II, and WorldSID (Been et al., 2004; Yoganandan et al., 2004; Yoganandan and Pintar 2005; Yoganandan and Pintar 2005). A recent article focusing on the survey of PMHS data from side impact tests identified paucity of information for craniocervical structures (Yoganandan et al., 2006). The International Standards Organization based its recommendations for lateral neck bending from low acceleration (7 g) human volunteer studies which included no injuries, and data modified from an unpublished PMHS test conducted in France at an acceleration of 12.2 g (ISO 1999). In order to mimic real-world human responses, it is important for the dummy head to be "at the right place at the right time" during the loading event, i.e., neck responses are critical. From a motor vehicle perspective, craniocervical data are important as side impact dummies are continually subjected to improvements due to public awareness for safety and changing technological advancements such as side airbags including curtains. These data are also needed in other areas, e.g., aviation environment. In addition, three-dimensional (3-D) data are needed because the human neck exhibits intervertebral coupling in the lateral bending mode (Yoganandan et al., 1998). Consequently, the present research was designed with an emphasis on 3-D craniocervical biomechanical data from side impact sled

tests. Specifically, the objective of the current study was to determine 3-D head linear and angular accelerations and craniocervical forces and moments. In addition, tests were conducted with the European side impact dummy, ES-II, and responses were compared between the two models.

METHODS

Unembalmed PMHS were obtained, screened for HIV and Hepatitis A, B, and C, and anthropomorphic data were obtained. Pretest radiographs were obtained to identify various body and musculoskeletal components. This included the head, neck, chest, thorax, upper and lower extremities, and pelvis. Antero-posterior and lateral and functional x-rays of the craniocervical complex were examined in addition to overall evaluations. Specimens were dressed in tight-fitting leotards and positioned on a custom-designed seat rigidly fixed to the platform of a sled to simulate lateral impact. The seatback was oriented at an angle of thirteen degrees from the vertical, and the horizontal seat pan was unpadded. The seatback was slotted to allow mounts for accelerometers for the dorsal spine and sacrum. Specimens were seated upright with the Frankfort plane horizontal, legs stretched parallel to the mid-sagittal plane, and normal curvature and alignment of the thoracolumbar spine maintained without pretorso rotation. Belts were used to restrain the torso and extremities, and were routed at the level of bilateral axilla, twelfth rib, pelvis, and extremities connecting the anatomic regions to the seat. A rigid armpit fixture located bilaterally was used to direct the restraint system. Figure 1 shows the schematic of the restraining system. A head-neck support and release mechanism was used to maintain the normal forward looking upright initial position of the craniocervical complex. The head was secured using bilaterally placed parietal and frontal-parietal nylon cables and connected to the release mechanism balancing its weight. At the initiation of impact, the release mechanism activated, and the specimen sustained the acceleration input in the lateral mode.

Accelerometers were used on the sled, and head, upper and lower thoracic spines, and pelvis. A pyramid-shaped nine-accelerator package (PNAP) was fixed to the head (Yoganandan et al., 2006). The 3-2-2-2 design of PNAP consisted of three sets of biaxial accelerometers (Entran model EGE-73B, Fairfield, NJ) mounted to the triangular base of the pyramid and one triaxial accelerometer mounted at its vertex. The three base arrays constituted the biaxial 2-2-2 components, and the vertex contained the triaxial component of the 3-2-2-2 configuration. Figure 2 shows a schematic of the specimen along with the accelerometer instrumentation and PNAP. Positions and orientations of accelerometers were determined using a 3-D coordinate measuring device. Retroreflective photographic markers were placed on the head, first and twelfth thoracic vertebral spinous processes, and sacrum, and at specific predetermined locations on the sled and buck. Each set was such that a minimum of three individual non-collinear markers was positioned to describe 3-D motions. Motions were captured at a rate of 1-kHz using a nine-camera system (Vicon systems, Lake Forest, CA).

Following the test, specimens were palpated, a clinical-type examination of craniocervical structures was performed, functional x-rays were obtained, and a detailed autopsy was conducted. Head-neck complexes were isolated by dissecting at the craniocervical junction, and the entire ligamentous cervical columns were preserved. Radiographic assistance was used during isolation. Radiographs and close-up computed tomography images were obtained. All pre- and posttest evaluations were conducted by the same personnel. Inertial properties of the head were obtained using accepted techniques (Pintar et al., 2005).

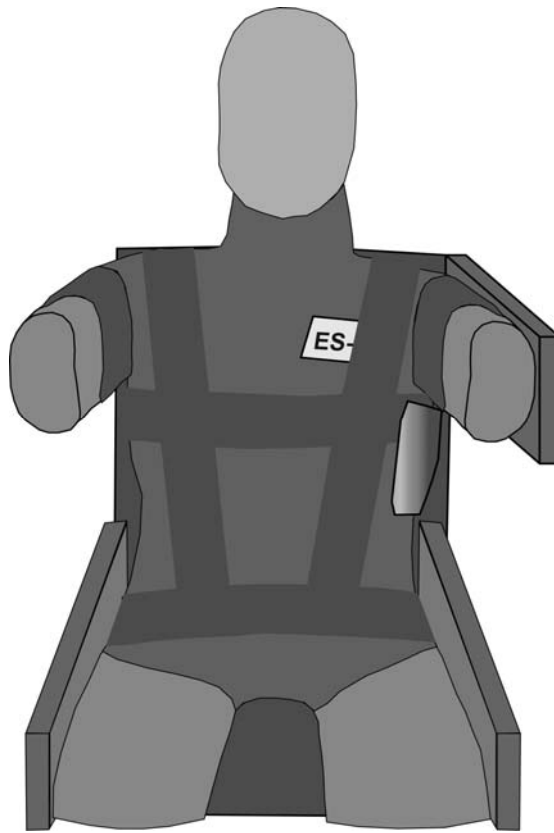
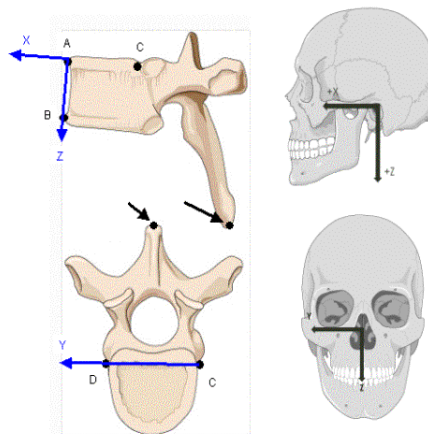
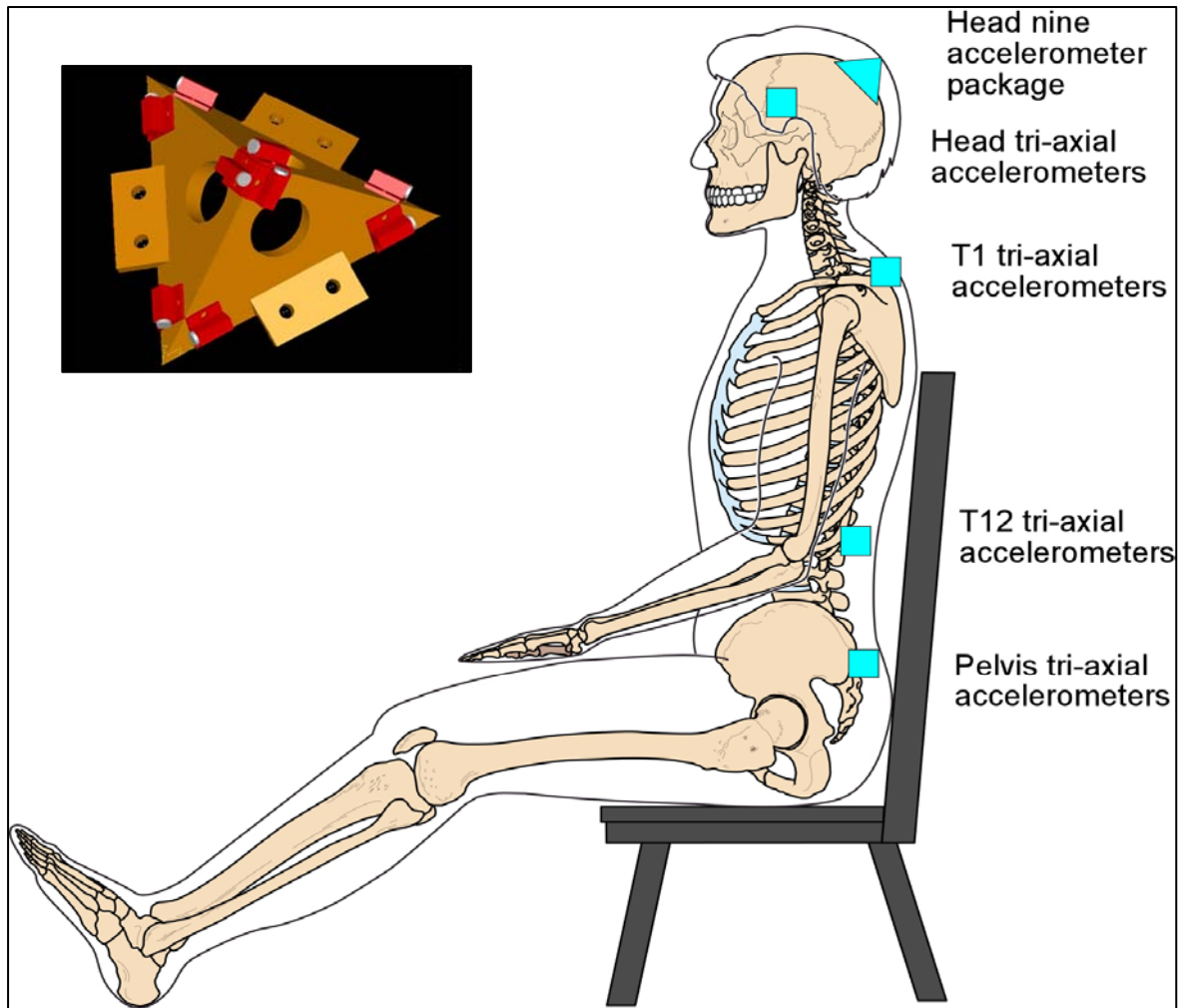


Figure 1: Schematic showing the restraint system used in the test.

The right handed Cartesian coordinate system of reference was adopted in the study. It is shown in figure 3 for the head and T1 vertebra. For the head, the origin was at its center of gravity (cg) with the x-axis parallel to the Frankfort plane from the mid point of left and right porion to the midpoint of left and right infraorbital notches; y-axis was parallel to the Frankfort plane from left to right porion; and z-axis was perpendicular to the Frankfort plane along the superior to inferior direction. For the upper thoracic spine, the origin was at the anterior superior tip of the vertebral body of T1; x-axis was the cross product of y-axis and normal vector from the mid superior-anterior aspect of T1 vertebra to its inferior-anterior aspect; y-axis was the normal vector from the left lateral superior aspect of T1 vertebra to its right lateral superior aspect; and z-axis was the cross product of the x- and y-axes.

Twenty-three channels of accelerometer-based data were collected at a rate of 12.5 kHz and processed according to SAE J211 specifications. Kinematics of the head, first and twelfth thoracic vertebra, and pelvis were obtained from high-speed marker images. Retroreflective marker data were combined with data from anatomic landmarks on the head, first thoracic vertebra, and sled to produce a kinematic model of the impact event. Temporal head angular and cg linear accelerations, and forces and moments at the occipital condyles in 3-D were computed using geometric and inertial data and dynamic equations of equilibrium. Figure 3 shows a flow chart of the experimental protocol.



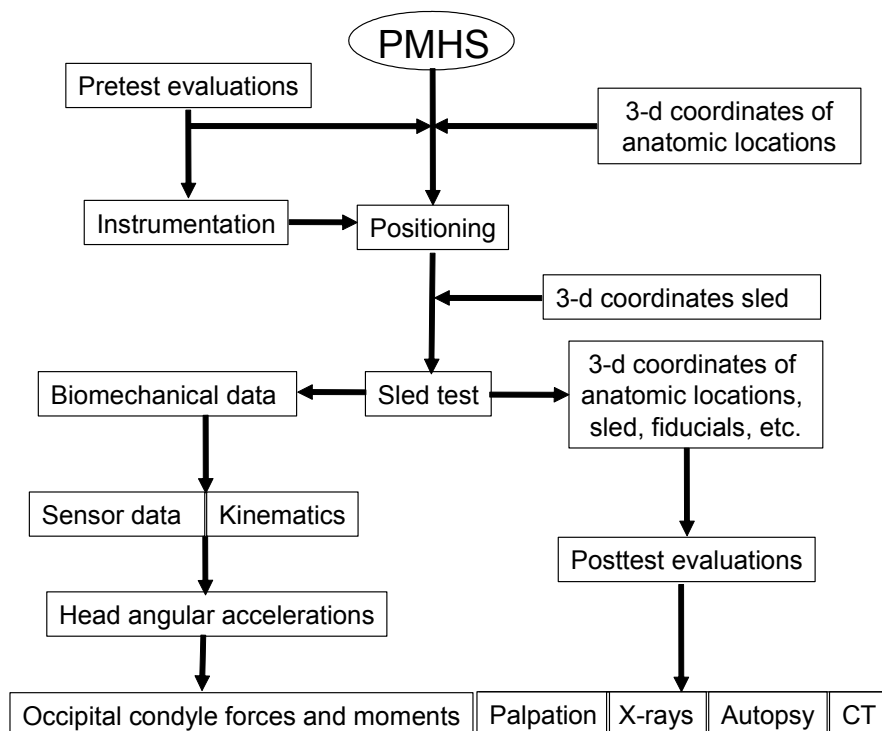


Figure 4 - Flow chart illustrating the testing protocol.

Parallel tests were conducted with ES-II dummy. While positioning was identical with PMHS experiments, instrumentation consisted of accelerometers for measuring head, upper and lower thoracic spine, and pelvis accelerations, and load cells for measuring forces and moments at the upper neck. This resulted in 31 channels per test. Data processing was also similar to PMHS tests. The head and neck were inspected following each test for visible signs of damage. In addition, head and neck bolts were inspected. Neck loads were directly measured using load cells. To compare ES-II results with moments generated at the occipital condyles from PMHS tests, data from the six-axis load cell placed inside the dummy head were transformed according to standard procedures using geometrical data. The computation only affected lateral and transverse (axial or torque) bending moments. The test matrix included five tests. The sequence consisted of testing ES-II dummy first, followed by the first PMHS experiment, and another ES-II dummy test. Head angular data were not gathered in this dummy test. A second pair of ES-II and PMHS experiments was conducted. The test matrix is shown in table 1. Results are compared between the two pairs of PMHS and ES-II dummy (third and fifth) experiments.

Table 1: Test Matrix and outcome.

Series	Test ID	Surrogate	Outcome
1	1	ES-II	Frangible head-neck, results not used
	2	PMHS	Results used in data analysis
	3	ES-II	Results used in data analysis
2	4	PMHS	Results used in data analysis
	5	ES-II	Results used in data analysis

RESULTS

Confining the paper to the objectives of the study, results are described with specific reference to the craniocervical region. The anthropometric data were such that the average age was 47 ± 2.8 years, stature was 187 ± 1.2 cm, and body mass was 78 ± 11.4 kg. Typical sled acceleration-time responses used in both PMHS and ES-II tests are shown in figure 5. The average sled acceleration was 13.5 g with a change in velocity of 12.4 m/s. Because only two tests were conducted using PMHS and ES-II surrogates, head linear and angular acceleration and craniocervical force and moment plots are presented by averaging data from these two tests. In order to facilitate comparison, each plot shows a comparison of PMHS and ES-II data on the same scale. Since head contact with the seat structure occurred in the second PMHS test, averages were obtained prior to contact. Head cg linear accelerations along the x-, y-, and z-directions are shown in figure 6 for the two models. Peak head cg lateral linear accelerations for PMHS were greater (27.9 g) than ES-II (17.5 g). Peak head cg accelerations in the transverse plane also showed a similar pattern with PMHS responding with greater magnitudes (mean: 35.3 g) than ES-II (22.6 g). The dummy responded with the lowest fore-aft accelerations (5.8 g). Head rotational accelerations for both models are shown in figure 7. In the frontal plane, both models responded with similar peak magnitudes (PMHS: 2906 rad/sec/sec, ES-II (2869 rad/sec/sec). Rotational accelerations were lower in the other two planes (Figure 7). Craniocervical forces and moments for both models are shown in figure 8 and 9, respectively. Peak tensile forces and lateral bending moments at the occipital condyles were greater for PMHS (1370 N and 87.4 Nm) than ES-II (891 N and 49.7 Nm). Lateral shear forces and were also greater in PMHS than ES-II (1081 versus 732 N). Fore-aft shear forces were minimal (< 500 N) in both models. The torque was greater in PMHS (20.2 Nm) than ES-II (12.8 Nm). Fractures were not identified to any bony structure of the craniocervical complex. C4-C7 right facet diastases occurred in one specimen and mild joint changes from the axial to fourth vertebral levels occurred in the other specimen. Brain tissues were not evaluated.

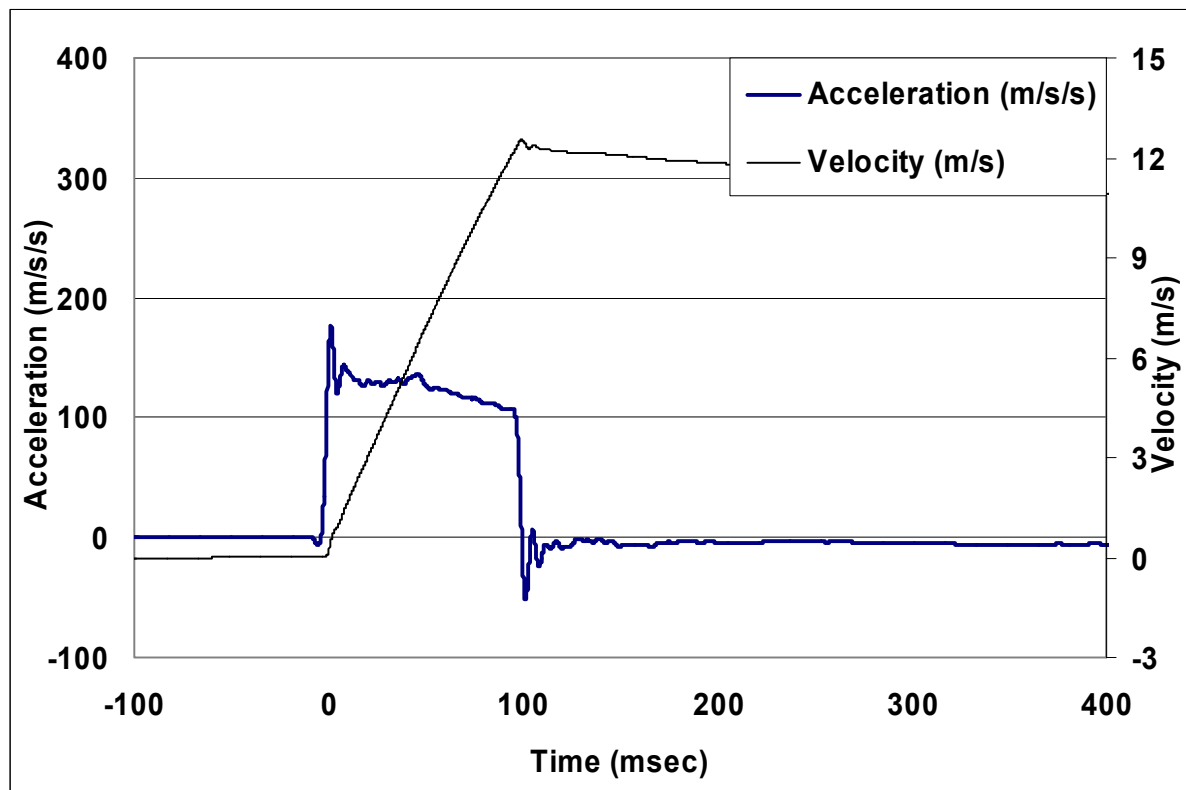


Figure 5: Typical acceleration and velocity profiles used in the study.

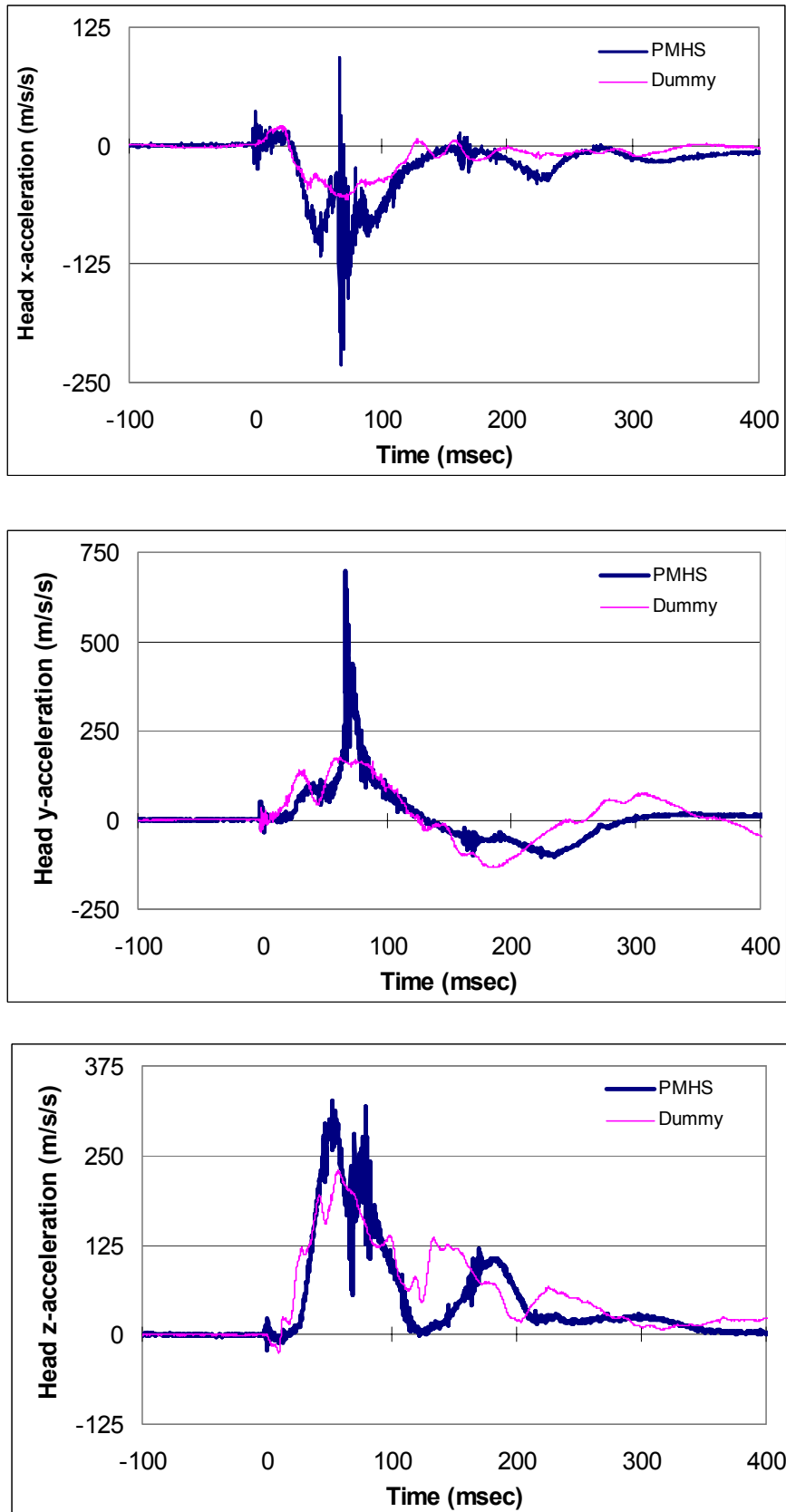


Figure 6 - Comparison of head linear acceleration between PMHS and dummy tests. Refer to text for details. All three components are shown.

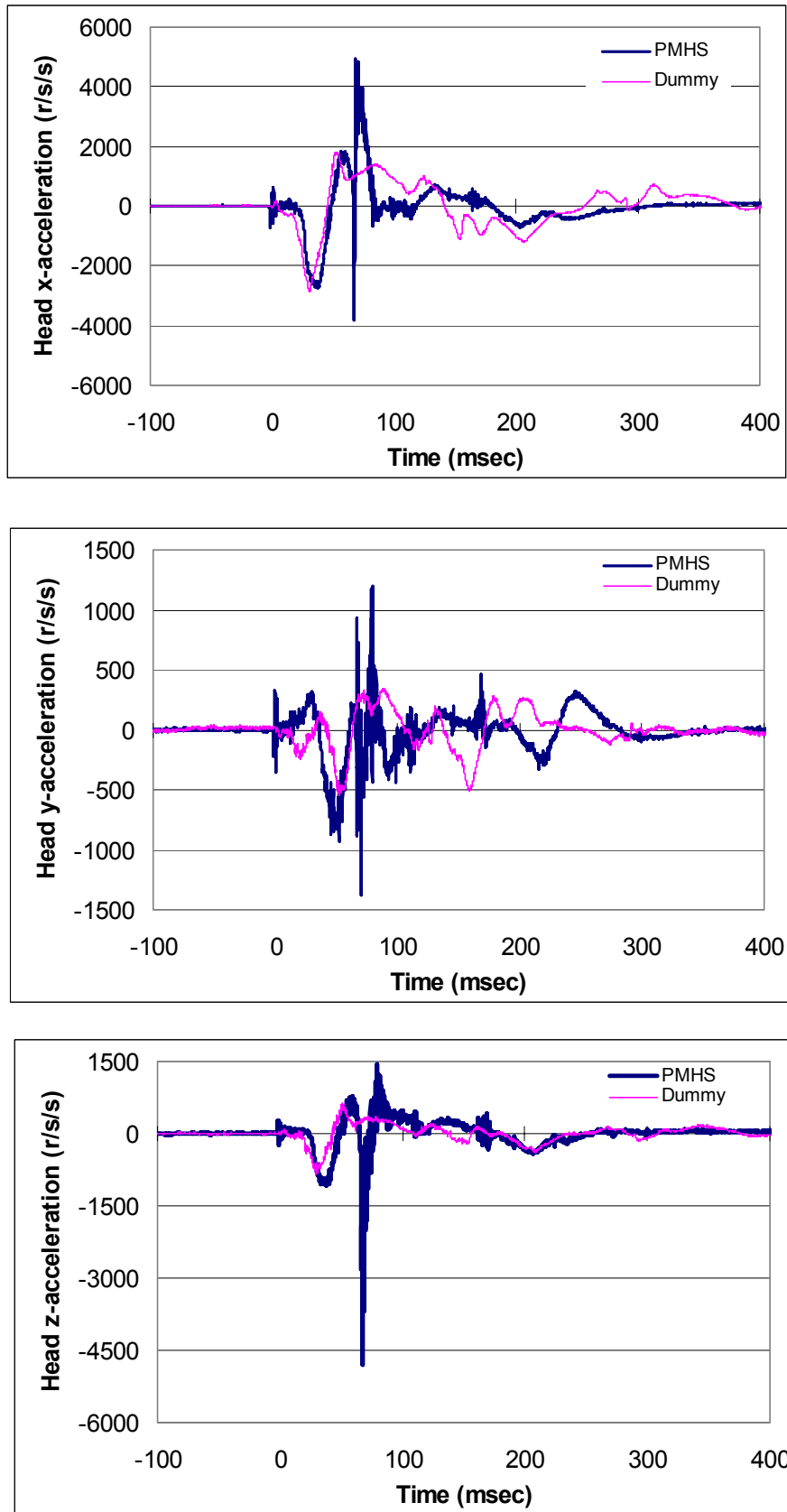


Figure 7 - Comparison of head angular acceleration between PMHS and dummy tests. Refer to text for details. All three components are shown.

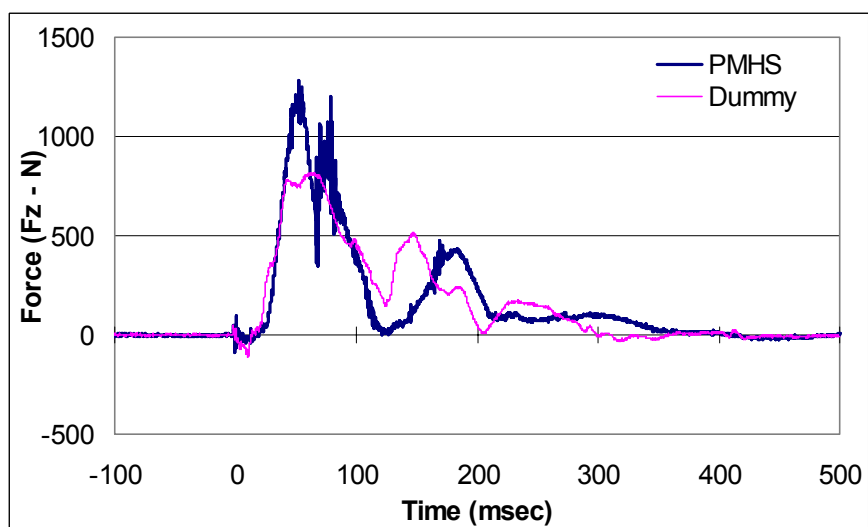
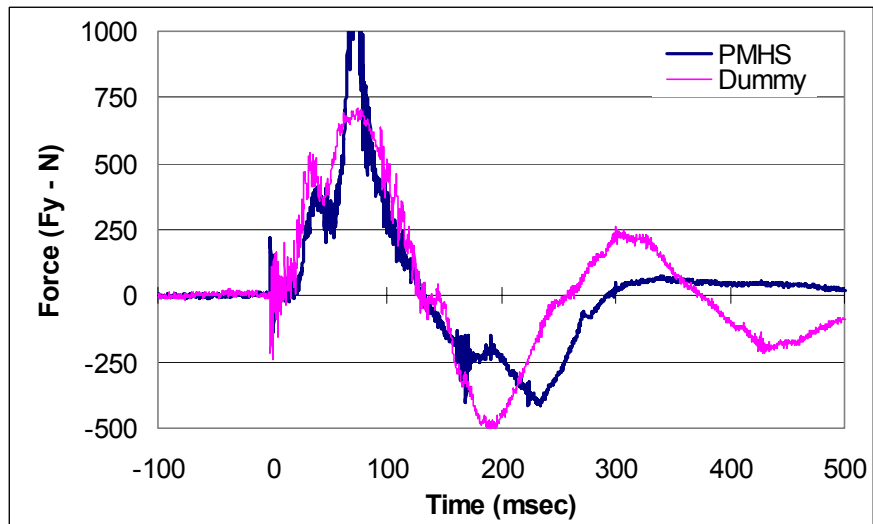
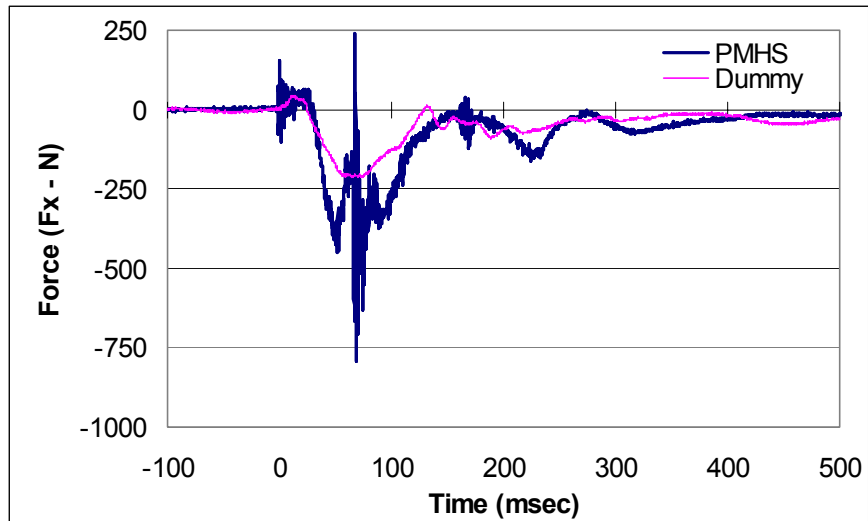


Figure 8 - Comparison of occipital condyle forces between PMHS and dummy tests. Refer to text for details. All three components are shown.

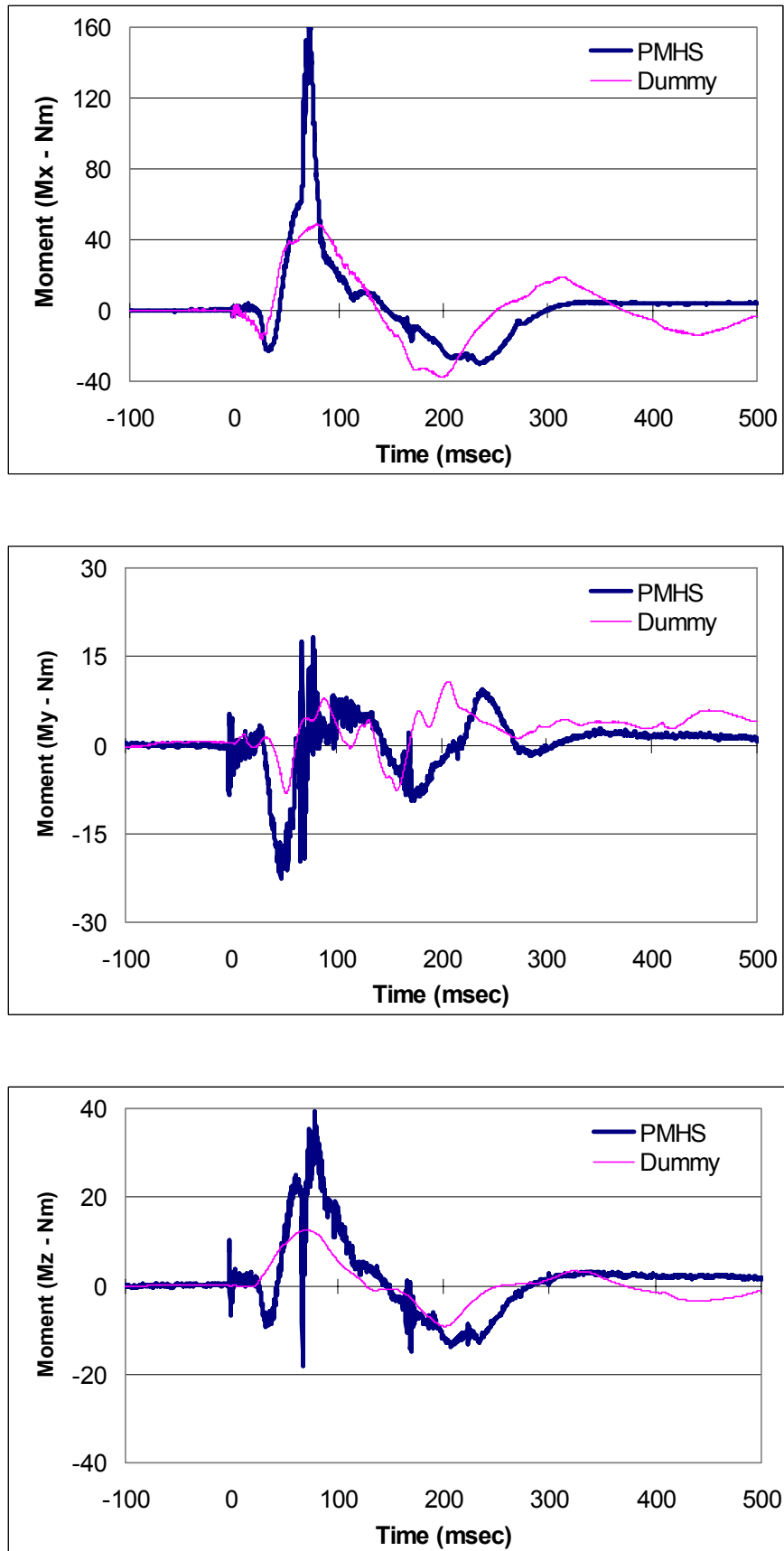


Figure 9 - Comparison of occipital condyle moments between PMHS and dummy tests. Refer to text for details. All three components are shown.

DISCUSSION

Different types of side impact dummies are available for lateral impact evaluations. The current US FMVSS standards (214 and NCAP) specify the use of SID, representing the 50th percentile male occupant. Since the development of this dummy in 1979, BIOSID, EuroSID (later termed ES), and WorldSID, all representing the 50th percentile male occupant are used in crashworthiness evaluations. SID-IIs, representing the small-sized female occupant, is also used in crashworthiness assessments. However, based on recent rulemaking activities, US standards appear to be poised to gradually phase-in ES-IIre, a version of ES. This dummy closely conforms to its predecessor ES-II except for the rib extension component (Kuppa et al., 2003). In the present study, ES-II was selected because this dummy is specified in the European standards ECE-95 for crashworthiness evaluations and used in aviation studies (Soltis et al., 2003). Therefore comparisons are appropriate between PMHS and the selected side impact dummy. To evaluate biofidelity responses of other dummies, it would be necessary to repeat the current series of tests and compare with PMHS responses determined in this study. Since dummy constructions, are different, especially at the craniocervical junction, present PMHS data may drive side impact dummies towards a more biofidelic design.

The human cervical spine responds differently in frontal and lateral impacts; flexion-extension and side bending behaviors show nonuniform characteristics (Yoganandan et al., 1998). Craniocervical structures respond to flexion-extension loading essentially in a symmetrical manner with little contributions from off-axis components. Rotations in the axial and coronal planes are of the second order type in frontal impacts. Therefore, under forward bending of the head-neck, two-dimensional analysis often suffices for crashworthiness evaluations. In contrast, rotations of cervical intervertebral joints in the transverse plane are coupled with rotations in the frontal plane. In other words, lateral bending and axial torque in the spine are coupled, and vice versa. This results in non-planar response even under pure lateral impact acceleration. From an anatomical perspective, uncovertebral joints and uncinat processes of the adult human neck vertebrae in association with the changing anatomy of apophyseal joint articulations contribute to the non-planar response (Kumaresan et al., 1997). Eccentric head weight even on a normally aligned, neutrally positioned spine induces additional non-planar motions in the cervical column. Therefore, it is important to account for these off-axis motions to evaluate craniocervical dynamics in lateral impacts. This dictates the use of 3-D high-speed continuous motion analysis system.

A review of literature revealed initial attempts to extract 3-D data from previous tests. Wismans and Spenny analyzed human volunteer tests conducted at the United States Naval Biodynamics Research Laboratory (NBDL) during the years 1976-1980 and developed performance criteria for mechanical necks in lateral flexion (Ewing et al., 1977; Ewing et al., 1978; Wismans and Spenny 1983). In a later study, using selected NBDL information from 1980 and 1973 databases, the analysis was extended to include other impact modes (Ewing and Thomas 1973; Ewing et al., 1975; Wismans and Spenny 1984). Wismans et al., continued analyses of tests conducted between 1980 and 1981 with the aim of developing a mechanical neck for dummies with omni-directional biofidelity, and a two-pivot model was reported (Wismans et al., 1986). The authors concluded that the model is only applicable for low velocity human volunteer impacts with conditions used during NBDL setup, and additional information is needed from PMHS tests for higher exposure levels. Differences were found when these data were compared with results from five PMHS tests (Wismans et al., 1987). Errors existed in the instrumentation system used in NBDL tests; specifically, T1 vertical displacements did not match due to thoracic rotations. The authors underscored the need to insure accurate test methodology and use 3-D photogrammetry techniques for data reduction. The present study adopted these principles.

In a later analysis, analytical procedures were developed for correcting artifacts in thoracic rotations from NBDL data (Thunnissen et al., 1995). Philippens et al. reanalyzed NBDL data using these procedures and developed head-T1 response corridors (Philippens et al., 2004). While thoracic rotation artifacts were corrected, even this effort resorted to other assumptions. For head inertial properties, Philippens et al. also used the same procedures adopted in earlier analyses (Wismans et al., 1987; Thunnissen et al., 1995; Philippens et al., 2004). Because of this approximation, Philippens et al. indicated that occipital loads might have up to 15% error even disregarding the change in head kinematics due to altered mass (Philippens et al., 2004). Increased head mass due to instrumentation affects head kinematics, and hence, occipital loads. In a more recent study, Pintar et al. also underscored the importance of obtaining accurate data during the experimentation process for accurate 3-D analysis and determination of occipital condyle loads (Pintar et al., 2005). Been et al., combined kinematic and sensor data to obtain occipital moments using an earlier version of the WorldSID dummy (Been et al., 2004). Thus, while researchers have developed approaches to analyze previously gathered data, and recognized various elements needed for 3-D analysis, to our best knowledge, this is the first study to use synchronized 1000 frames/second digital images and gather appropriate data during the experimentation process to determine 3-D kinematics of PMHS in crashworthiness research. This included the identification of the importance of mass and moment of inertia parameters, and the present tests used specimen-specific values for these variables. Therefore, data obtained from the present testing paradigm, although motion responses are not presented because of page constraints, are more realistic and accurately represent true responses.

Another issue is the accurate determination of angular accelerations of the head. From a theoretical perspective, an array 3-2-2-2 of accelerometers is necessary and sufficient to compute head rotational accelerations (Padgaonkar et al., 1975). Applying equations to obtain rotational accelerations from a 3-2-2-2 accelerometer array is relatively straightforward as long as the bookkeeping of the sensor position and orientation is maintained. This is easier in theory than practice, and at least one investigation admitted after the test that “two accelerometer channels were interchanged” to yield “more reasonable” results (Cheng et al., 1982). Because a validated device housing the array was not available in archival literature, a recent study identified critical criteria for PMHS use (Pintar et al., 2005). The criteria included lightweight, low profile, rigid fixation, and above all, accurate locations of the placement on the nine accelerometers in addition to systematic data acquisition, bookkeeping, and processing. To determine angular accelerations with high level of confidence, a recently designed and fabricated device that was validated with extensive tests, was used in the current study. The use of PNAP was deemed to be an integral component of the experimental design, and with this recognition, accurate head cg linear and angular accelerations were obtained, resulting in realistic craniocervical forces and moments. These processes enabled a direct comparison of computed multiaxial forces and moments from PMHS tests with ES-II dummy tests. As stated earlier, load cell-measured moment data in ES-II dummy tests were modulated to account for the equivalent occipital condyle location before comparing with similar data from PMHS tests.

A comparison of responses between the two models demonstrated similarities and differences. Similarities included patterns in biomechanical data, e.g., shapes of head angular and linear accelerations, and occipital condyle forces and moments. Greater tensile forces in PMHS than ES-II occurred due to decoupling issues and increased upper thoracic accelerations in the former surrogate. ES-II is more coupled in the head-neck region than PMHS, and additional decoupling of PMHS head-neck structures has been reported in other loading modes. For example, Yoganandan et al. showed that PMHS craniocervical structures are more decoupled than the Hybrid III dummy head-neck under axial compressive loading (Yoganandan et al., 1989). The decoupling occurs in PMHS structures due to ligamentous attachments between occipital-atlanto-axial complexes devoid of intervertebral discs that span and provide enhanced coupling in the subaxial cervical spine.

Head-neck kinematics due to inertial effects from the increased head weight compared to the neck induces axial distractive forces in the acting ligamentous structures and contributes to multiplanar motions and increased tensile forces at the upper neck junctions. Thus, the transfer of the external load within craniocervical structures in PMHS occurs in a complex and coupled mechanism with the involvement from soft tissue components. This was evidenced in functional radiographs and CT images by the identification of abnormalities of soft tissue structures without accompanying bony fractures. Although head-neck junctions incorporate distinctly different designs between the two dummies, i.e., Hybrid III and ES-II, the present study may have identified a need to improve the biofidelity of ES-II in this loading mode.

An interesting observation from PMHS tests is the level of angular acceleration and absence of skull fractures. It is well known that PMHS is not the best surrogate to determine brain injury because of issues such as tissue degradation. Consequently, direct correlations with in vivo data are not appropriate from this study. However, it is possible to compare angular acceleration data with injury thresholds suggested in literature. Gennarelli et al. synthesized various studies by considering diffuse brain injuries to be a continuous spectrum spanning from mild, classical, and severe concussions to mild, moderate, and severe diffuse axonal injuries, with associated concussion grades of 1 ranging from 1 to 5 (Ommaya and Gennarelli 1974; Gennarelli et al., 2003). Mild to moderate concussions were associated with angular acceleration levels ranging from 2.8 to 5.6 krad/sec/sec. While full statistical analysis from the present dataset is not possible owing to sample size limitations, peak angular accelerations correspond to low-level brain injuries, belonging to the mild traumatic type. Although not reported in the results section of this paper, head injury criteria values were well below 1000 in all tests. These findings may be important in real world environments if crashworthiness evaluations are to be based on angular acceleration levels especially for lower severity mild to moderate traumatic brain injuries. The present study may have laid a foundation for further research in this area.

As indicated elsewhere in this paper, it is important for the dummy to have a biofidelic response for the head and neck in order to better replicate the human. Neck response influences head kinematics because it supports the head mass and connects head and torso structures. In a data analysis work from 15 human volunteer tests, Wismans et al. underscored the need to obtain data at high lateral impact acceleration levels (Wismans et al., 1986). Yoganandan et al. during a review of side impact sled PMHS tests failed to identify published craniocervical biomechanical data (Yoganandan et al., In Press). During the development of lateral impact response requirements to assess the dummy biofidelity, recognizing that ISO used one PMHS test which produced neck fractures, the present study offers additional information (ISO 1999). Data from the current series of PMHS tests that produced no bony injuries to any part of the craniocervical structure will be of value to assess dummy responses. Parallel tests with ES-II and comparing responses have shown the applicability of this method.

A limitation of the present study is the sample size and use of one type of side impact dummy. Consequently, results should not be generalized to the entire population. From this perspective, this study serves as a first step in the determination of head linear and angular kinematics and forces and moments at the craniocervical junction in lateral impacts. Another issue is the use of a custom-designed rigid seat and constraining torso and pelvis motions. This approach was used in the experimental design to insure maximum kinematic transfer of the lateral acceleration to craniocervical structures during the impact event, and hence, can be regarded as a worst-case scenario. Conventional real world belt systems often allow relative motions between the torso and pelvic regions that may result in altered head-neck kinematics and forces and moments at the occipital condyles. Additional tests relaxing these constraints are necessary to delineate the biomechanics. These topics are considered further research.

SUMMARY

Sled tests were conducted using PMHS and ES-II models to determine 3-D head linear and angular accelerations and craniocervical forces and moments. Peak head cg lateral linear accelerations for PMHS were greater than ES-II dummy while the shape of the pulse was similar between the two models. Transverse and fore-aft accelerations were also lower in ES-II. Maximum head angular accelerations in the frontal plane were similar between the models. Peak tensile forces and lateral bending moments and lateral shear forces and torque at the occipital condyles were greater in PMHS than ES-II. These data, albeit from a limited dataset are valuable in establishing response corridors for side impacts and evaluating side impact dummies used in crashworthiness and safety-engineering studies.

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