

Head and Neck Responses of Post Mortem Human Subjects in Frontal, Oblique, Side and Twist Scenarios

Yun-Seok Kang, Jason Stammen, Kevin Moorhouse, John Bolte IV

Abstract Head and neck responses of anthropomorphic test devices and computational human body models should be validated in different impact modes, e.g., frontal, oblique, side, and twist. The main objective of this study is to create biomechanical response targets of the head and neck of post mortem human surrogates using a controlled mini-sled system in various impact scenarios. A mini-sled was designed to dynamically test a post mortem human subject head-neck complex. A six axis load cell was attached at the T3 level of the spine to measure the reaction loads at the upper thoracic spine for the frontal, oblique and side impacts, while T1 was attached to the load cell for the twist test. The post mortem human subject head, C3, and C5 were instrumented using accelerometers and angular rate sensors to capture head and cervical kinematics. Five post mortem human subjects were tested in frontal (5), oblique (2), side (2), and twist (1) scenarios at a nominal mini-sled velocity of 14 km/h for frontal, side, and oblique impacts and 1,800 deg/s for the twist scenario. Biomechanical responses of the head and lower neck were measured in various impact conditions. Biomechanical targets were created for future biofidelity evaluation for anthropomorphic test devices and computational human body models.

Keywords PMHS head and neck, biomechanical target, cervical kinematics, lower neck loads.

I. INTRODUCTION

Cervical spine injuries occur commonly in different impact modes, and societal costs due to the injuries are high in motor vehicle crashes (MVC) [1-7]. Upper cervical injuries tend to be the most serious type of neck injury so the neck injury criterion, Nij, calculated at the upper cervical vertebrae was developed and has been used for frontal crashes [16]. However, survivors with neck injuries in frontal and rear MVCs often sustain injuries in the lower portion of the cervical spine [6-7]. With the exception of very young occupants (8 years or less), where the majority of neck injuries are in the OC/C1/C2 region [47], children and adults are more likely to sustain lower cervical spine and cervicothoracic junction injuries than upper cervical injuries [6].

In order to investigate neck injury tolerance and criteria, many studies have been conducted using human volunteers or post mortem human surrogates (PMHSs) in different impact directions [3-5][7-13]. Upper neck criteria exist in Federal Motor Vehicle Safety Standard (FMVSS) No. 208, partly because upper neck injuries tend to be the most serious type of neck injury in frontal crashes, but also because upper neck loads, i.e., forces and moments at occipital condyle, in PMHS testing are relatively easy to calculate using the inertial properties and kinematics of the head alone. Due to the difficulty of accurately measuring lower neck loads, many studies have focused primarily on upper neck loads of human volunteers and PMHSs by using inverse dynamics approaches [5][17-27]. However, lower neck injuries are also important in that they have been commonly observed in both epidemiological [14] and experimental studies using PMHSs in frontal, side, and rear impacts [3-5][7-8][15]. If lower neck (cervicothoracic junction) biomechanical data could be obtained along with an injury risk function for the lower neck, it could be used in conjunction with the work that has been conducted on upper neck loads to evaluate the biofidelity of anthropomorphic test devices (ATDs) and to validate and improve finite element human body models (HBMs). The addition of a lower neck criterion would improve the overall protection of the spine, neck, and head in motor vehicle occupants [5][7][10].

A few studies reported lower neck loads that were determined using inverse dynamics, for which the neck

Y. Kang, PhD is an Assistant Professor in the Injury Biomechanics Research Center at the Ohio State University in Columbus, OH, USA (e-mail: yunseok.kang@osumc.edu, tel: +1 614 366 7584, fax: +1 614 292 7659). J. Stammen, PhD and K. Moorhouse, PhD are at the National Highway Traffic Safety Administration (NHTSA), USA and J. Bolte, PhD is in the Injury Biomechanics Research Center at the Ohio State University, USA.

was regarded simply as a massless rigid link [7][13][31-35]. Another study reported that the massless inverse dynamics technique for the lower neck calculation could create unexpected errors on the force and moment data [28]. To our knowledge, although computational models have been used to quantify neck muscle tension [49,50], no studies have measured PMHS neck muscle tension forces separately from the vertebral column loads and used those combined muscle/vertebral loads to calculate the lower neck loads in different impact directions. Head and neck responses of ATDs and computational human body models should be evaluated in different impact modes, e.g., frontal, oblique, side, and twist, since three-dimensional head and neck motions are recorded even in unidirectional laboratory, sled, and crash tests [29]. The objective of this study is to create biomechanical responses of the head and lower neck of PMHS in various impact directions.

II. METHODS

Post Mortem Human Surrogates (PMHS)

The PMHSs 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 Institutional Review Board (IRB) protocol, were reviewed and followed. Five fresh-frozen male PMHSs (age 49 ± 16 years) were scanned using dual energy X-ray absorptiometry (DXA) and computed tomography (CT) to ensure that there were no severely degenerated discs, osteophytes, or hardware due to previous spinal surgery on the cervical and upper thoracic spine. PMHS age, area bone mineral density (aBMD) score (lumbar T-score), height, weight, and impact modes are shown in Table I. Detailed anthropometric data and inertial properties of the head and neck can be found in Tables AI and AII (Appendix A). The inertial properties of the head and neck were measured using a repeatable and reliable device validated by Self et al. (1992) [48]. Dissection information for the head and neck for the measurement of the inertial properties was provided in the previous study [28]. The mass moment of inertia of the head and neck was measured using an internal inverted torsional pendulum (Moment of Inertia Instrument Models XR – 50 and GB-3300AX, Space Electronics LLC, Berlin, CT) [48]. PMHS1 and PMHS2 were tested in three modes (frontal, oblique, and side), PMHS3 was tested in both frontal and twist modes, and PMHS4 and PMHS5 were tested in the frontal mode only.

TABLE I
PMHS INFORMATION
(F: FRONTAL IMPACT, O: OBLIQUE IMPACT, S: SIDE IMPACT)

	Age	aBMD Lumbar T-Score	Height (cm)	Weight (kg)	Cause of death	Test mode
PMHS1	67	+1.1	184.5	71.0	Pancreatic cancer	F,O,S
PMHS2	57	-2.0	175.0	64.0	Lung cancer	F,O,S
PMHS3	54	-1.9	175.3	74.1	Non-Small cell lung cancer	F,T
PMHS4	25	-2.6	177.8	73.4	Metastatic synovial sarcoma	F
PMHS5	40	+0.4	175.3	67.0	Metastatic melanoma	F
Mean	49	-1.0	177.6	69.9	N/A	N/A
SD	16	1.6	4.0	4.3	N/A	N/A

Test Setup

A mini-sled was designed to dynamically test a PMHS head-neck complex in various impact directions, e.g., frontal, oblique, side, and twist, at a nominal velocity of 14 km/h (frontal, side, and oblique) and 1,800 deg/sec (twist) shown in Fig. A1. This nominal velocity was determined by mimicking T1 acceleration in the x direction from the full body frontal PMHS test conducted in study [7]. The target T1 acceleration used for the mini-sled input in this study is shown in Fig. 5b in study [7]. Figure 1 shows an instrumentation overview (1a) and general setups for the mini-sled in frontal (1b), oblique (1c), and side impact (1d) modes. A custom-sized elliptical ring (Fig.1 A) was used for attaching upper thoracic structures (1st rib, clavicles, muscles and skin) in an anatomic configuration (Table BI), to account for the contribution of these structures to the kinematics of the head-neck complex. Information about detailed PMHS dissection was provided previously [28]. The ring was fixed to the

mini-sled with turnbuckles (Fig.1 B) in line with six uniaxial load cells (Fig.1 C) that measure passive axial muscle forces during the event (muscle force data shown in Appendix B). The turnbuckles provide adjustability of initial neck muscle tension. The initial muscle tension was adjusted to 40 – 50 N, in order to maintain 75 – 100 N of initial neck pre-load when including head weight, to be consistent with previous studies [18-19]. A six axis load cell (Fig.1 D) that was originally designed for the Hybrid III 3YO dummy lumbar spine (Humanetics Innovation Solutions, Farmington Hills, MI, USA) was attached at the T3 level of the vertebral column to measure the reaction loads at the upper thoracic spine. The T3 vertebra was affixed within a custom cup (Fig.1 E) using Bondo® Body Filler (Bondo Corporation, Atlanta, GA USA). The used Bondo and potting cup masses were measured and used to inertially compensate the measured T3 loads. When the T3 vertebra was potted, the T1 vertebra was tilted forward to be 24 ± 5 degrees based on previous studies [11,12]. The head was carefully released by gravity then lifted slowly until the head angle (e.g Frankfort plane) became 0 ± 5 degrees so that natural cervical spine curvature can be maintained without applying any possible abnormal forces on the neck (Fig. B1 in Appendix B). The PMHS head was instrumented using six accelerometers and three angular rate sensors (ARS) installed on an external tetrahedron fixture (t6a ω) [30] (Fig.1 F). In order to capture kinematics of the upper and lower cervical spine, three accelerometers and three ARS (3a ω) were installed at both the C3 and C6 vertebra as described in previous studies [4][28] (Fig.1 G). Each instrumentation mount was digitised using a FARO arm device (Faro Arm Technologies, Lake Mary, FL, USA) to transform data from the origin of the instrumentation to desired coordinate systems. The PMHS head was supported by a harness that was attached to a solenoid release system. The release system was activated prior to mini-sled motion.

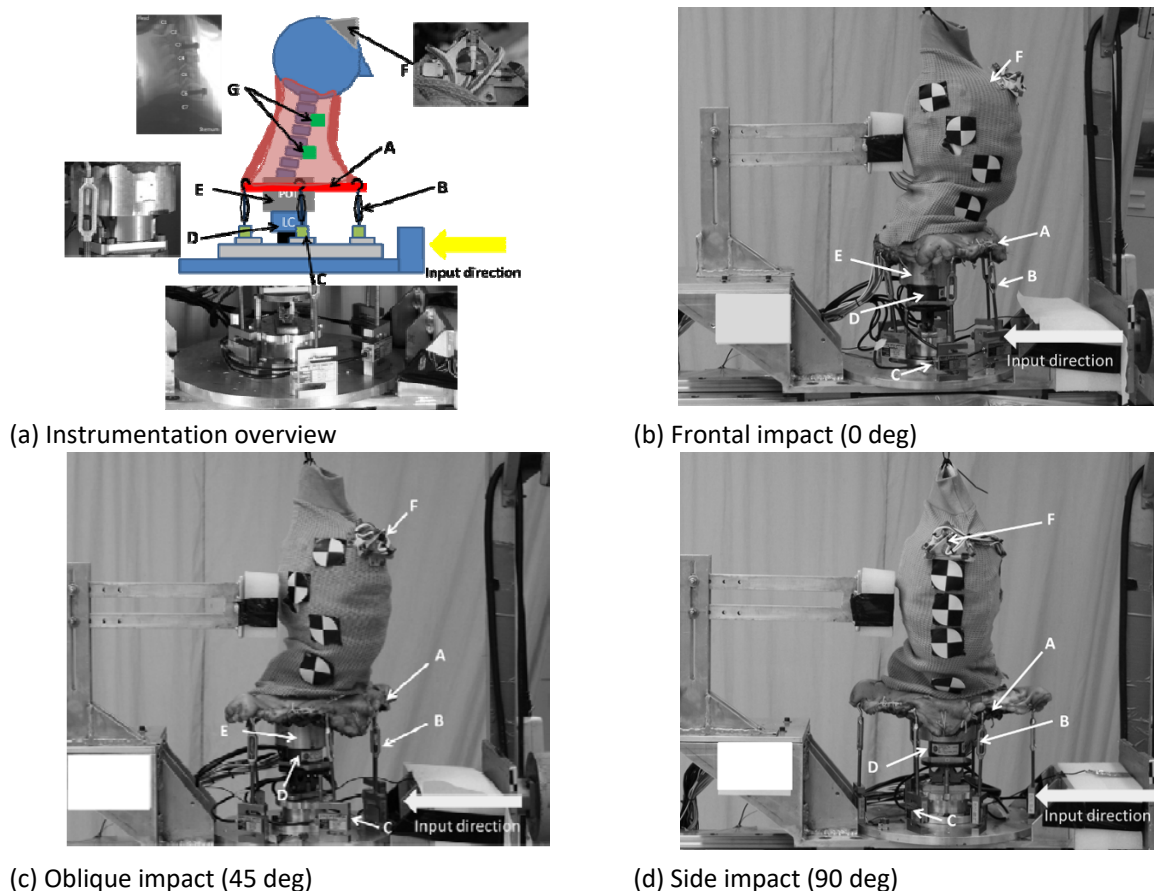


Fig.1. Mini-sled set up for frontal, oblique and side impacts.

A: Custom-sized elliptical ring; B: Turnbuckles for initial muscle tension; C: Uniaxial load cells (6) for measuring muscle tension; D: Six axis load cell at T3; E: Custom potting cup for fixing T3 to the sled; F: Head instrumentation (t6a ω); G: C3 and C6 instrumentation (3a ω).

For the neck twist test, the mini-sled fixture was modified to create rotational input. Figure 2 shows the neck twist test setup with PMHS head-neck complex. The rails for the mini-sled were removed, and the mini-sled

fixture was clamped to the test table. In order to create an isolated neck twist condition, the T2 and T3 were removed. The T1 was affixed within the custom pot (Fig.2 H). The head was attached to the stationary fixture using head holders (Fig.2 A). The mastoid process (Fig.2 B) was aligned to the anterior-superior edge of the T1 vertebra (Fig.2 D) as shown in the lateral view (Fig. 2a). Two of the six uniaxial load cells (Fig.2 C) previously shown in Figure 1 were used to measure torsion applied to the ring (Fig.2 E). A threaded rod (Fig.2 I) that was screwed into the uniaxial load cell was welded to the ring such that the load cell could be attached to the ring rigidly (Fig. 2b). The six axis load cell (Fig.2 F) used previously was attached to the potting cup (Fig.2 H) to measure vertebral torsion at T1. A hinge joint was installed at the centre of the disk such that one dimensional torsion was generated. A moment arm (Fig.2 G) was attached to the disk, and the ram pushed the moment arm to generate rotational input (Fig. 2a). In this test condition, the head was stationary while the lower neck rotated clockwise from a top view.



(a) Lateral view

(b) Posterior oblique view

Fig.2. Twist test set up with PMHS head neck complex.

A: Head holders; B: Mastoid process; C: Uniaxial load cells (2) for measuring muscle resistance force; D: Anterior-superior edge of T1; E: Custom-sized elliptical ring; F: Six axis load cell at T1; G: Input moment arm; H: Custom potting cup for fixing T1 to the sled; I: Threaded rod that connects the load cell to the ring (welded to the ring).

Data Processing

The sampling frequency used in all testing was 20 kHz and all data obtained from the tests were filtered according to SAE J211. Data measured from the head instrumentation were transformed to the head CG in the body-fixed coordinate system that was defined by digitising the infraorbital notches and external auditory meati (x-direction forward and z-direction downward according to SAE J211). The C3 and C6 data were transformed to the vertebral coordinate system used in a previous study, see Kang et al., 2016 Appendix A, [28]. Lower neck loads were calculated by combining T3 vertebral loads, i.e., six axis load cell, with muscle passive axial loads, i.e., uniaxial load cells. A free body diagram and equations for the lower neck were presented in the previous study (see Fig. 5 and Eqs 16-18 in Kang's study) [28]. Detailed information for the defined coordinate system was also provided in the previous study [28]. Biomechanical targets for the head, C3, and C6 kinematics and neck loads (both upper and lower) were created in the frontal impact tests. Biomechanical targets for frontal impacts (n=5) were created using a previously published method [3]. Due to small sample size in side (n=2), oblique (n=2), and neck twist (n=1) modes, individual response curves were provided in this study instead of biomechanical targets. For the neck twist mode, a lower neck moment vs. rotation response that considers both vertebral and muscle moments and input rotational kinematics was quantified. No head kinematics were recorded in the twist mode since the head was stationary, i.e., no head motion.

III. RESULTS

Frontal Impacts

Biomechanical targets for head linear acceleration and lower neck force in the x and z directions are shown in Figs. 3 and 4, respectively. Head rotation and lower neck moment about the y axis are presented in Fig. 5. C3 and C6 rotations are provided in Fig. A2 (see Appendix A). Lower neck moment with respect to the head rotation about the y axis is shown in Fig. A5 (a).

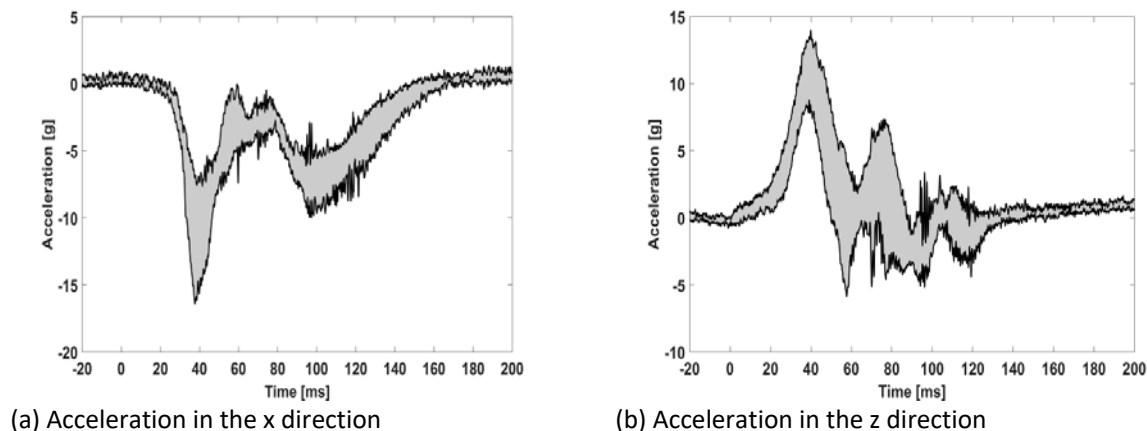


Fig. 3. Head linear acceleration (frontal impact).

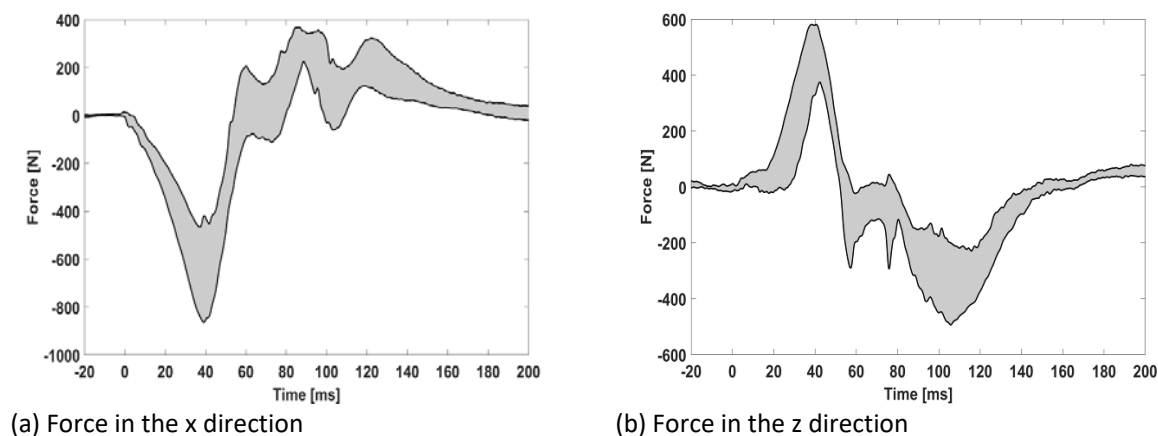


Fig. 4. Lower neck force (frontal impact).

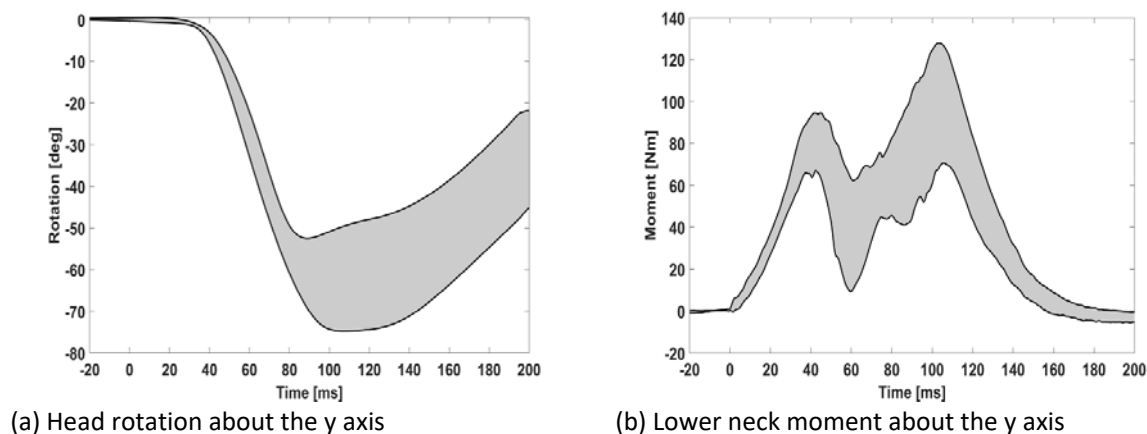
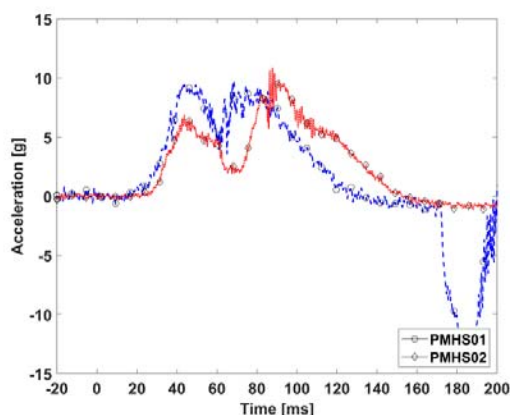


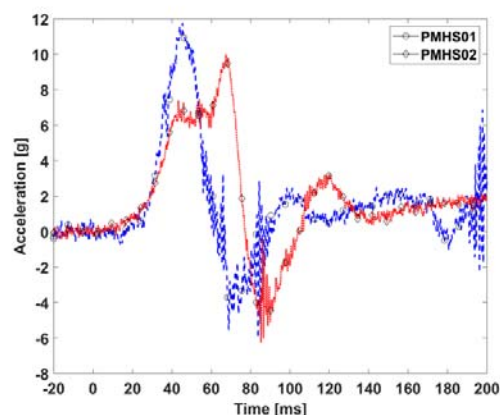
Fig. 5. Head rotation and lower neck moment (frontal impact).

Side Impacts

Biomechanical responses for head linear acceleration and lower neck forces in the y and z directions are shown in Figs. 6 and 7, respectively. Head rotation and lower neck moment about the x axis are shown in Fig. 8. C3 and C6 rotation about the x axis are presented in Fig. A3. Lower neck moment with respect to the head rotation about the x axis is shown in Fig. A5 (b).

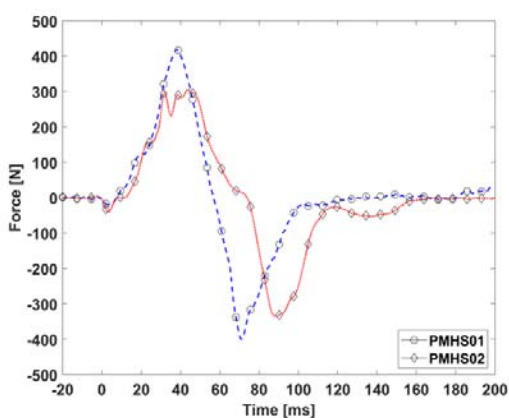


(a) Acceleration in the y direction

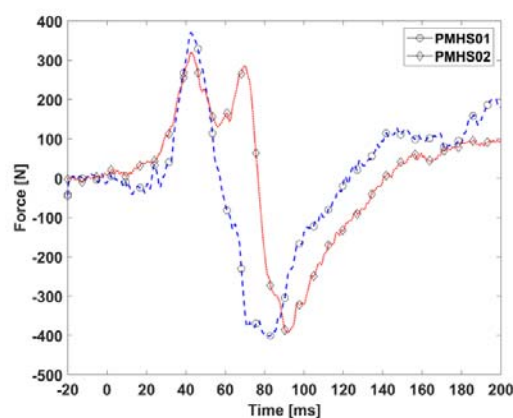


(b) Acceleration in the z direction

Fig. 6. Head linear acceleration (side impact).

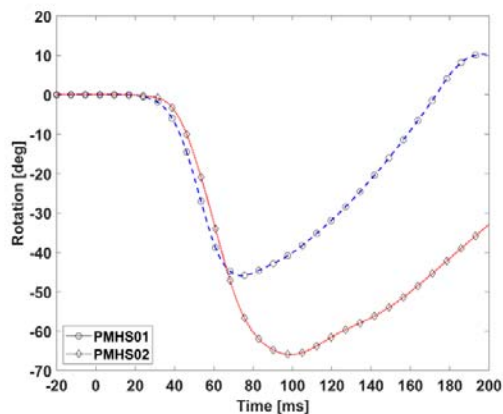


(a) Force in the y direction

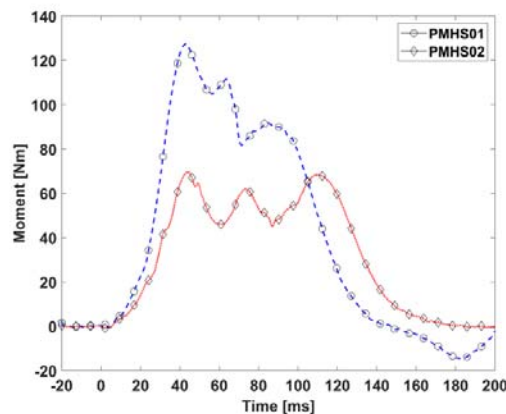


(b) Force in the z direction

Fig. 7. Lower neck force (side impact).



(a) Head rotation about the x axis

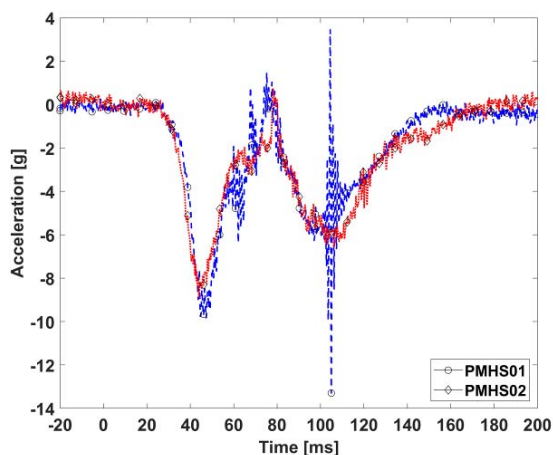


(b) Lower neck moment about the x axis

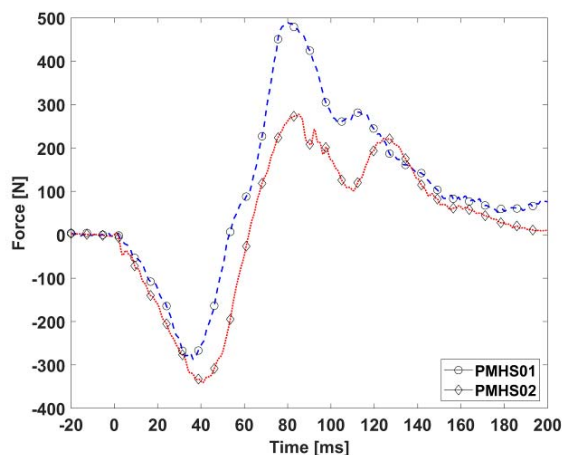
Fig. 8. Head rotation and lower neck moment (side impact).

Oblique Impacts

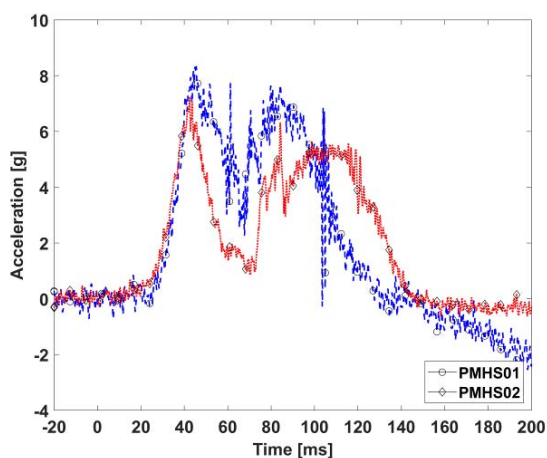
For the oblique condition, three dimensional (x, y and z) head acceleration and lower neck forces are shown in Fig. 9. Three dimensional head rotation and lower neck moments are shown in Fig. 10. C3 and C6 rotations are provided in Fig. A4. Lower neck moments with respect to the head rotations are presented in Fig. A6.



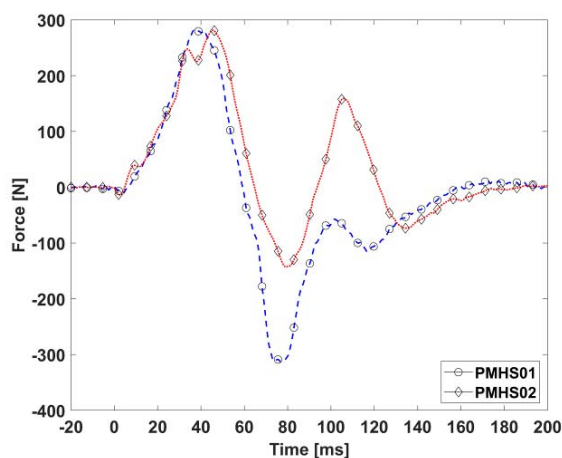
(a) Acceleration in the x direction



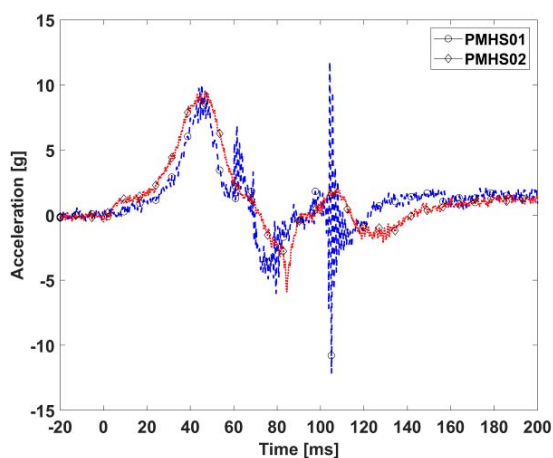
(b) Force in the x direction



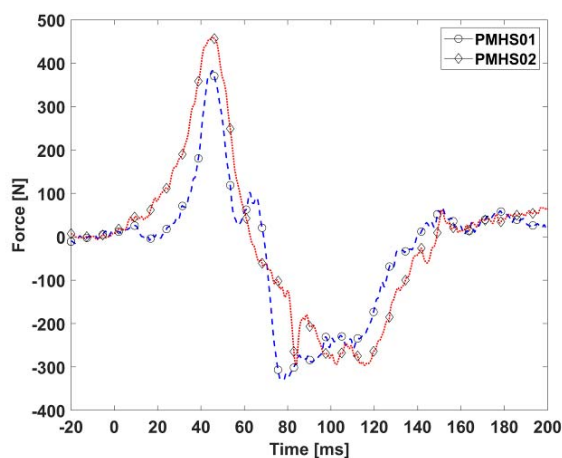
(c) Acceleration in the y direction



(d) Force in the y direction

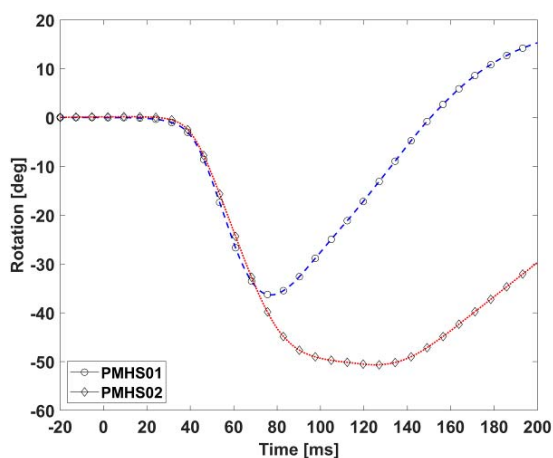


(e) Acceleration in the z direction

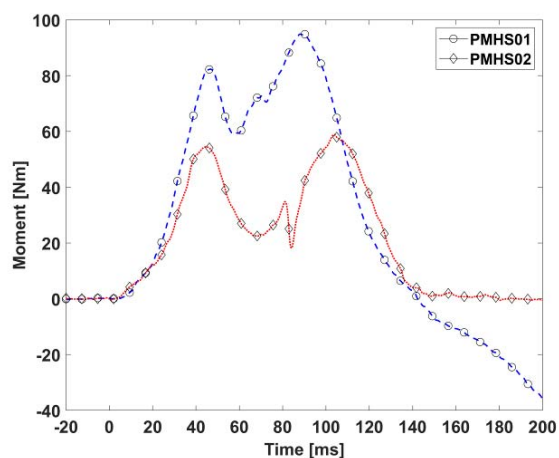


(f) Force in the z direction

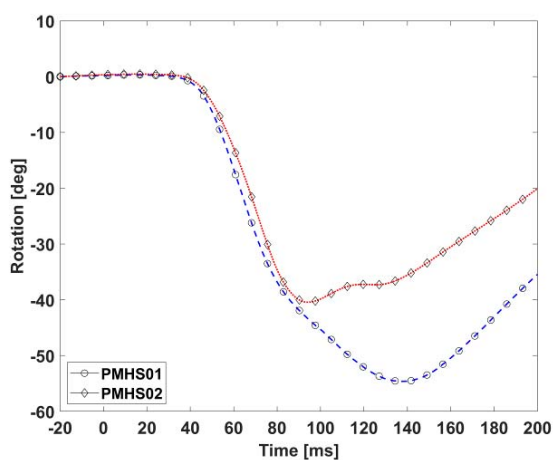
Fig. 9. Head linear acceleration and lower neck force (oblique impact).



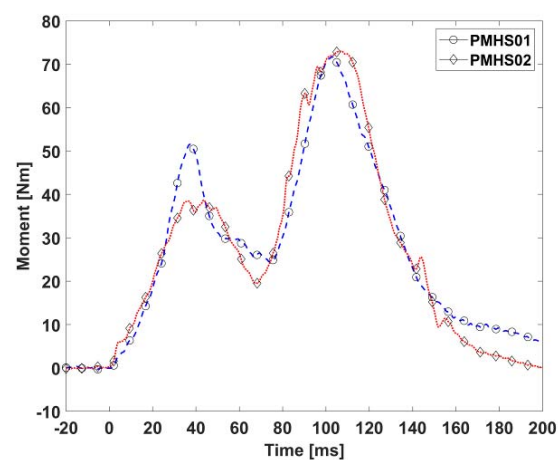
(a) Head rotation about the x axis



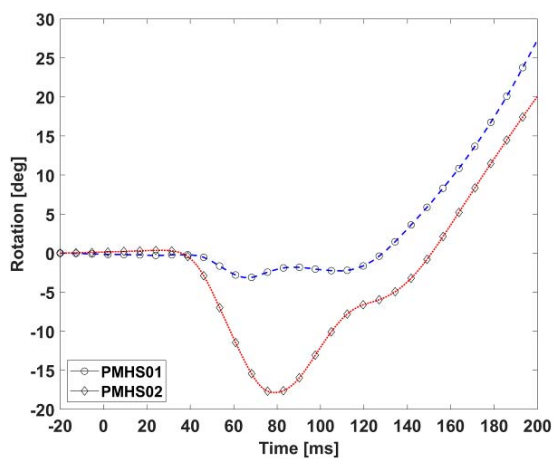
(b) Lower neck moment about the x axis



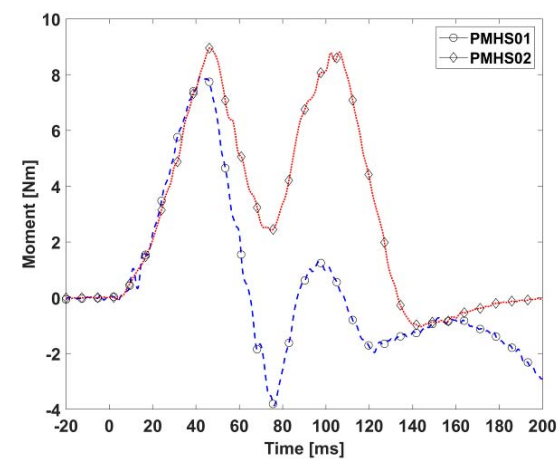
(c) Head rotation about the y axis



(d) Lower neck moment about the y axis



(e) Head rotation about the z axis

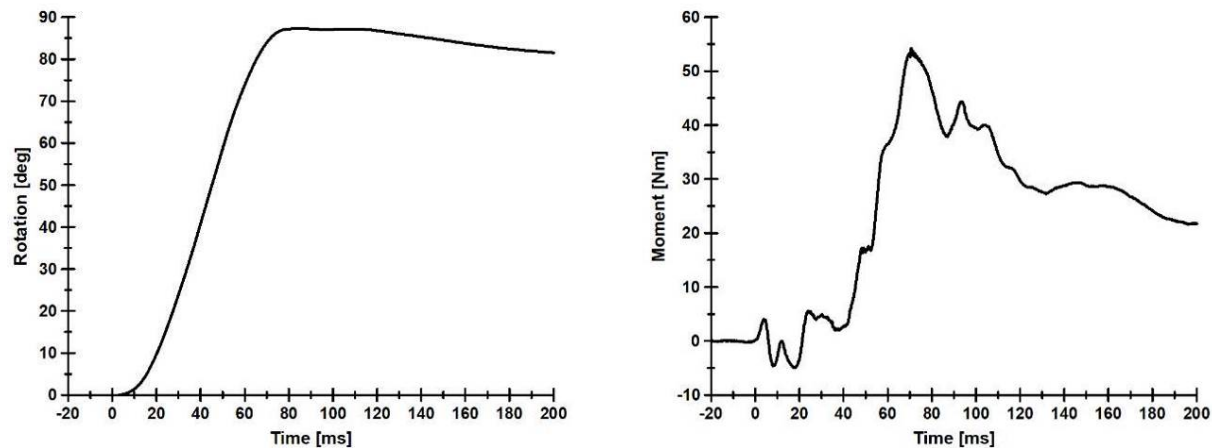


(f) Lower neck moment about the z axis

Fig. 10. Head rotation and lower neck moments (oblique impact).

Neck Twist

For the twist condition, the sled disk rotation and lower neck moment about the z axis are shown in Fig 11(a) and 11(b), respectively. Lower neck moment with respect to the rotation is shown in Fig. A7.



(a) Sled disk rotation about the z axis

(b) Lower neck moment about the z axis

Fig. 11 Rotation and moment in twist condition

IV. DISCUSSION

This study provides biomechanical responses of the head and neck in frontal, side, oblique and twist directions. A novel test set-up using a custom-sized ring and novel instrumentation techniques allowed the measurement of lower neck loads that included muscle tensions so that the inverse dynamic technique using a massless link assumption was not necessary. Although current ATDs are capable of measuring lower neck loads using load cells, ATD lower neck loads are not generally considered in biofidelity evaluation since very little biomechanical data have been published to evaluate ATD biofidelity. The present study provides comprehensive head and neck kinematics and kinetics, including C3 and C6 rotational kinematics, for biofidelity assessment of ATDs and finite element models. Biomechanical data from the current study also can be used to better understand the importance of lower neck loads when considering musculature so a more biofidelic neck and upper thoracic spine can be designed and added in the current or future ATDs.

Frontal Impacts

Five PMHSs were tested in the frontal impact scenario. Frontal impact testing was conducted first on each PMHS. Biomechanical targets, i.e., corridors, were created using the five PMHSs. Due to page limits, only head acceleration and rotation, C3 and C6 rotation, and lower neck Fx, Fz and My are presented in this study (Fig. 3 – 5 and Fig. A2). The head kinematics and lower neck loads obtained from the current study were comparable to data from a previous frontal impact PMHS study [7]. Average head accelerations at the CG measured in this study were 18.4 g in the x direction and 8.6 g in the z direction, while those measured from study [7] were 14.9 g and 20.0 g in the x and z directions, respectively. The shear forces measured from the current study ranged from -868.4 to -450.0 N while those from study [7] ranged from -784.9 to -374.5 N as shown in Table II. The axial forces from the current study ranged from 441.4 to 601.1 N, while study [7] reported axial forces from 462.2 to 870.8 N (Table II). It should be noted that study [7] used a full body PMHS sled set-up, while the current study used a mini-sled with the head-neck complex including T2-T3. These shear and axial forces were comparable to those in study [7], however, caution should be used for the comparison. A previous study found that the lower neck loads from employing the massless link method can be underestimated by over 20% [28]. The comparable outcomes found in the current study could be due to underestimation of the loads from study [7].

It should be noted that the lower neck moments about the y axis in this study were higher than those calculated in study [7], which used a massless link inverse dynamics approach. The lower neck moment ranged from 89.3 to 139.7 Nm in the current study, while those calculated using massless link theory were 68.1 to 125.1 Nm (Table II). Lower neck moments measured from the volunteer study in Naval Biodynamics Laboratory

(NBDL) frontal impacts varied from around 59 to 114 Nm [34], which is comparable to both PMHS studies (current and Pintar studies) even though their sled inputs (43.2 and 62.0 km/h) were higher. It is likely that the lower neck loads were higher in the current study due to incorporation of muscle tension forces into the lower neck moment calculation. In addition, the direct measurement method using cervical kinematics in this study should provide better accuracy of the lower neck loads than the massless link method used in both PMHS and volunteer studies [7][34]. It is also likely due to a limitation of the mini-sled system, which only translated in the x direction with no translation in the z direction and no rotation in the y axis, causing the lower neck moment to be higher than that measured in full body PMHS tests.

C3 and C6 rotations about the y axis ranged from -64.5 to -45.8 degrees and -39.7 to -19.7 degrees, respectively (Fig. A2). To the best of the authors' knowledge, no studies have reported rotational kinematics at C3 and C6 in frontal impacts. These C3 and C6 rotational kinematics will be valuable biomechanical targets for finite element models.

TABLE II
PMHS RESULTS COMPARISON TO OTHER RESEARCH RESULTS IN FRONTAL IMPACTS

	PMHS1	PMHS2	PMHS3	PMHS4	PMHS5	Study [7]	Study [34]
<i>Lower neck Fx (N)</i>	-512.6	-722.9	-848.4	-450.0	-868.4	-784.9 to -374.5	NA
<i>Lower neck Fz (N)</i>	472.4	601.1	441.4	500.0	561.0	462.2 to 870.8	NA
<i>Lower neck My (Nm)</i>	91.5	92.7	119.1	89.3	139.7	68.1 to 125.1	59 to 114

Side Impacts

Two PMHSs (PMHS1 and PMHS2) were tested in the side impact scenario. Although neck injury and responses have been documented and quantified using PMHSs previously [8][36-39], to the best of the authors' knowledge, only one study quantified lower neck force and moment of PMHSs in side impact conditions [39]. Study [39] conducted side impact sled tests using eight PMHSs with change in velocities of 31.3 to 64.4 km/h. Head and neck responses, including lower neck force and moment, were reported in their study [39]. Unfortunately, they did not report linear acceleration measured at the head but rather angular acceleration. Angular acceleration determined from this study ranged from -1,839.2 to -1,616.6 rad/s² with approximately 42.9 to 52.1 ms duration while study [39] showed -1,700 to -1,500 rad/s² (these values were estimated from the time history plot provided in the study [39]) with around 40 ms duration in their 31.3 km/h test. Based on this consistency in head angular acceleration, the mini-sled tests, with ΔV of 14 km/h (11 g with 70 ms) applied to the upper thoracic spine, was comparable to their 31.3 km/h (13 g with 100 ms) full body PMHS tests. Table III shows a comparison of lower neck force and moment measured from this study to those from study [39]. Lower neck Fx (304.7 – 418.7 N) and Mx (69.6 – 127.5 Nm) were comparable to the results from study [39] (572.5 ± 227.4 N and 68.4 ± 59.6 Nm, respectively), while lower neck Fz (319.2 – 369.3 N) was almost four times smaller than that from study [39] (1253.2 ± 351.5 N). This is probably due to the limitation of the mini-sled system, which only allowed for a single degree of freedom motion at the T3 (translation in the y direction) with no translation in the z direction so neck tension in the mini-sled could be smaller than that from the full body sled testing. It should be noted that lower neck Mx (69.6 – 127.5 Nm) from this study was greater than that from study [39] (68.4 ± 59.6 Nm). A large portion of the applied energy could be transferred to the neck tension in the full body sled testing, while this energy was transferred into lateral neck bending with less neck tension in the mini-sled system. In order to use a well-controlled mini-sled set up with a head-neck complex, the lower neck (or upper thoracic spine) had to be fixed to the sled base, which restricted motion of the lower neck. This restricted motion resulted in more bending moment than the full body side impact sled tests in study [39], since the upper thoracic spine was allowed to translate with respect to the three-point seat belt in that test configuration. PMHS1 had lower head rotation than PMHS2, while PMHS1 had higher moment about the x axis than PMHS2 (Fig. 8). This is likely due to subject variation between two PMHSs, i.e., PMHS2 had a more flexible neck than PMHS1).

C3 and C6 rotations about the x axis ranged from -42.9 to -35.0 degrees and -29.2 to -25.2 degrees, respectively (Fig. A3). PMHS2 had larger C3 and C6 rotations than PMHS1 likely due to the biomechanical variation between PMHSs. To the best of the authors' knowledge, no studies have reported rotational kinematics at C3 and C6 in side impacts.

TABLE III
PMHS RESULTS COMPARISON TO OTHER RESEARCH RESULTS IN SIDE IMPACTS

	PMHS1	PMHS2	31.3 km/h test In study [39] Mean (SD)
<i>Lower neck Fx (N)</i>	418.7	304.7	572.5 (227.4)
<i>Lower neck Fz (N)</i>	369.3	319.2	1253.2 (351.5)
<i>Lower neck My (Nm)</i>	127.5	69.6	68.4 (59.6)

Oblique Impacts

Two PMHSs (PMHS1 and PMHS2) were tested in the oblique scenario with ΔV of 14 km/h. The head linear accelerations, rotations, and lower neck loads are shown in Fig. 9 and 10. It should be noted that accelerations observed in the PMHS1 test (Fig. 9) exhibited noise around 105 milliseconds in x, y, and z directions. This was due to direct contact between a wire tie used for the head harness attached to the release system and the head instrumentation. After the PMHS1 test, the head instrumentation fixture was better protected from this contact. It should be noted that both PMHS1 and 2 show repeatable responses for the head acceleration and lower neck forces shown in Fig. 9 but not for the head rotations and lower neck moments shown in Fig. 10.

A few studies have focused on PMHS and ATD responses in oblique scenarios; however, neck responses were not reported in the studies [29][40-42]. The main responses quantified in those studies were whole body trajectories for each body segment or thoracic responses due to belt interaction in oblique sled testing. One volunteer study discussed lower neck loads using NBDL data [43]. They assumed the head-neck complex as a 2-pivot linkage mechanism with head torsion [43]. They used a 45 degree impact direction (same as the current study) with ΔV ranging from 10.9 to 15.0 km/h, which is also comparable to the current study (14 km/h). Table IV shows a quantitative comparison of responses from the current study to study [43]. Head rotation about the y axis (-54.6 to -40.4 deg) measured in the current study is comparable to that reported in study [43] (-57.9 $\pm$ 13.1 deg). They also reported the head torsional rotation (-12.1 $\pm$ 6.8 deg), which was comparable to the head torsional rotation measured in PMHS2 (-17.9 deg) but larger than that from PMHS1 (-3.1 deg). Even though the neck linkage rotation they measured was not exactly the same as the C3 rotation measured in the current study, their neck linkage rotation was similar to the C3 rotation from one of the PMHS. The neck linkage rotation about the y axis (-70.1 $\pm$ 14.4 deg) was comparable to PMHS2 C3 rotation (-68.5 deg) but larger than PMHS1 C3 rotation (-52.9 deg), as shown in Table IV. Lower neck moments about the y axis (71 – 73 Nm) were lower than those from the study [43] (117 – 160 Nm) as shown in Table IV. For the volunteer studies like study [43], they had to estimate head inertial properties, such as mass, centre of gravity and mass moment of inertia, by using regression models from the literature. This discrepancy could be due to inevitable errors from the estimation of inertial properties of the head from the Wismans study (i.e., volunteer study). Inertial properties of the head have been characterised previously using PMHS and human volunteers [48-54], and it has been demonstrated how error in inertial properties can produce inaccurate upper neck loads. Study [9] found that 3 mm CG location differences can affect upper neck loads by as much as 17%, and 5% of error in MMI can result in 17% error in upper neck moments. C3 and C6 rotation about both the x and z axes can be found in Appendix A (Fig. A4).

TABLE IV
PMHS RESULTS COMPARISON TO OTHER RESEARCH RESULTS IN OBLIQUE IMPACTS

	PMHS1	PMHS2	Study [43] Mean (SD)
<i>Head rotation (deg) about the y axis</i>	-54.6	-40.4	-57.9 (13.1)
<i>Head rotation (deg) about the z axis</i>	-3.1	-17.9	-12.1 (6.8)
<i>Neck linkage rotation (deg) about the y axis</i>	N/A	N/A	-70.1 (14.4)
<i>C3 rotation (deg) about the y axis</i>	-52.9	-68.5	N/A
<i>Lower neck My (Nm)</i>	71.7	73.0	117 to 160 Nm*

* Lower neck My from study [43] was estimated from the plot provided in their study.

Twist Impacts

Atlantoaxial dislocation and unilateral facet dislocations can occur with excessive neck torsion. In order to

understand neck torsional responses and injury, studies using PMHS have been performed [44-46]. In the current study, one PMHS (PMHS3) was tested in the neck twist condition. It should be noted that the current experimental set up allows for the measurement of lower neck torsional moment including passive muscle resistance, which no previous studies have reported. Also, an input rotational rate of 1,800 deg/s used in this study is higher than previous studies in which isolated cervical spines were used for torsion experiments. A previous study used a maximum input rotation rate of 500 deg/s [46]. Myers's study used the ligamentous cervical spine with no muscles, while the current study used intact head and neck with muscles. Myers's study used the dens as the upper center of rotation and 1/5 length of the T1 as the lower center of rotation, while the current study used the mastoid process as the upper center of rotation and the anterior edge of the T1 as the lower center of rotation. The maximum failure torque found in Myers's study was approximately 32 Nm, which was estimated from Figure 9 in Myer's study [46], while the maximum torque value obtained from the current study was 54.2 Nm (Fig 11 and A7). This larger torque measured from the current study than that from Myers's study is likely due to the passive muscle influence on the torsional resistance of the neck and also due to different measuring locations (occiput in Myers's study and T1 in the current study). Since the current study has only one sample, more data are needed to evaluate the passive neck muscle effect on the neck's torsional response and injury potential.

Limitations

For the side, oblique, and twist tests, biomechanical targets could not be created due to small sample size of one or two PMHS. Some PMHS were also tested multiple times (PMHS 1, 2 and 3) to produce head-neck responses in various impact modes. Palpation and x-ray were performed after each test to check for cervical spine injury. However, soft tissue injury cannot be detected by this injury assessment method; therefore, caution should be used when reviewing responses from the second and third tests (e.g., side, oblique and twist tests) from a given PMHS. However, since head and neck responses from PMHS in these various impact directions are limited in the literature, this study should help to better understand biomechanical responses of the head and neck, especially lower neck loads and cervical kinematics. One major challenge of the experimental setup in this study was how to maintain anatomical similarity of the neck musculature. The initial neck muscle tension was set up while the head was held by the head harness and release system. Since the head release system was activated right before the time of event, the initial neck tension could be reduced. This reduction could also affect kinematics of the head and neck. This limitation can be improved by using spring muscle replicators used in previous studies [11,12]. It was assumed that head and neck responses including neck musculature would provide more realistic boundary conditions than using an isolated cervical spine. However, uniaxial load cells used to measure neck muscle tension were installed in the global z directions. Future tests should consider the use of multi-axis load cells so that forces in x and y directions can be measured to calculate resultant forces for the neck muscle tension. Since PMHS were used, lack of muscle activation should be considered when the results from the study are used in ATD or finite element modeling studies. A nominal mini-sled input was used for the frontal, side and oblique impacts. This nominal mini-sled input was based on the T1 acceleration in the x direction from the full body PMHS tests in frontal impacts [7]. For side and oblique impacts, the head and neck responses were compared to those from the literature to justify the mini-sled input, which is a limitation of this study. The mini-sled used in this study allowed only one degree of freedom (e.g., single axis acceleration) so that motions and rotations about off-axes at the lower neck could not be applied to the mini-sled system as inputs. However, since the mini-sled setup was well controlled, researchers should be able to replicate the experiment from this study so that a physical or virtual computational model can be evaluated or validated against the results from this study.

V. CONCLUSIONS

Biomechanical responses of the head and neck were measured in various impact conditions, e.g., frontal, side, oblique, and twist. Biomechanical targets in the frontal impact and responses in side, oblique, and twist modes were created and provided for future biofidelity evaluation for ATDs and computational human body models.

VI. ACKNOWLEDGEMENT

Authors would like to thank Allison Yard, Arriana Willis, David Stark, Michelle Murach, Sam Goldman and all

members in the Injury Biomechanics Research Center at the Ohio State University, USA, and Brian Suntay from Transportation Research Center, Inc., USA, for their considerable support during testing days.

VII. REFERENCES

- [1] Quinlan K.P., Annett J.L., Myers B., Ryan G., Hill H. (2004) Neck strains and sprains among motor vehicle occupants—United States, 2000. *Accident Analysis & Prevention*, **36**(1): pp. 21–7.
- [2] Kullgren, A., Krafft, M., Ydenius, A., Lie, A., Tingvall, C. Developments in car safety with respect to disability: injury distributions for car occupants in cars from the 80's and 90's. *Proceedings of International IRCOBI Conference*, 2002, Munich, Germany.
- [3] Kang, Y. S., Bolte IV, J. H. *et al.* (2012) Biomechanical responses of PMHS in moderate-speed rear impacts and development of response targets for evaluating the internal and external biofidelity of ATDs. *Stapp Car Crash Journal*, **56**: p.105.
- [4] Kang, Y. S., Moorhouse, K., Herriott, R., Bolte IV, J. H. (2013) Comparison of cervical vertebrae rotations for PMHS and BioRID II in rear impacts. *Traffic Injury Prevention*, **14**(sup1): pp. S136–47.
- [5] McIntosh, A. S., Kallieris, D., Frechede, B. (2007) Neck injury tolerance under inertial loads in side impacts. *Accident Analysis & Prevention*, **39**(2): pp.326–33.
- [6] Goldberg, W., Mueller, C. *et al.* NEXUS Group. (2001) Distribution and patterns of blunt traumatic cervical spine injury. *Annals of emergency medicine*, **31**;38(1): pp.17–21.
- [7] Pintar, F. A., Yoganandan, N., Maiman, D. J. (2010) Lower cervical spine loading in frontal sled tests using inverse dynamics: potential applications for lower neck injury criteria. *Stapp Car Crash Journal*, **54**: p.133.
- [8] Kallieris, D., Mattern, R., Miltner, E., Schmidt, G., Stein, K. (1991) Considerations for a neck injury criterion. *SAE Technical paper*.
- [9] Pintar, F. A., Yoganandan, N., Baisden, J. (2005) Characterizing occipital condyle loads under high-speed head rotation. *Stapp Car Crash Journal*, **49**: p.33.
- [10] Pintar, F. A., Voo, L. M., Yoganandan, N. A., Cho, T. H., Maiman, D. J. Mechanisms of hyperflexion cervical spine injury. *Proceedings of IRCOBI Conference*, 1998, Gothenburg, Sweden.
- [11] Panjabi, M. M., Pearson, A. M. *et al.* (2004) Cervical spine ligament injury during simulated frontal impact. *Spine*. **29**(21): pp.2395–403.
- [12] Pearson, A. M., Panjabi, M. M. *et al.* (2005) Frontal impact causes ligamentous cervical spine injury. *Spine*, **30**(16): pp.1852–8.
- [13] Sundararajan, S., Prasad, P. *et al.* (2004) Effect of head-neck position on cervical facet stretch of post mortem human subjects during low speed rear end impacts. *Stapp Car Crash J.*, **48**: pp.331–72.
- [14] Yoganandan, N., Haffner, M. *et al.* (1989). Epidemiology and injury biomechanics of motor vehicle related trauma to the human spine (No. 892438). SAE Technical Paper.
- [15] Yoganandan, N., Pintar, F. A. *et al.* (2000) Biomechanics of human occupants in simulated rear crashes: documentation of neck injuries and comparison of injury criteria. *Stapp car crash journal*, **44**: pp.189–204.
- [16] NHTSA. (2000) FMVSS 208 Occupant Crash Protection. Docket of Federal Regulations 49, Part 571.208, US Government Printing Office, Washington, D.C.
- [17] Mertz, H.J. and Patrick, L.M., (1967) Investigation of the kinematics and kinetics of whiplash (No. 670919). SAE Technical Paper.
- [18] Mertz, H.J. and Patrick, L.M., (1971) Strength and response of the human neck (No. 710855). SAE Technical Paper.
- [19] Ewing, C.L. and Thomas, D.J., (1973) Torque versus angular displacement response of human head to -Gx impact acceleration (No. 730976). SAE Technical Paper.
- [20] Sundararajan, S., Prasad, P. *et al.* (2004) Effect of head-neck position on cervical facet stretch of post mortem human subjects during low speed rear end impacts (No. 2004-22-0015). SAE Technical Paper.
- [21] Howard, R. P., Bowles, A. P., Guzman, H. M., & Krenrich, S. W. (1998). Head, neck, and mandible dynamics generated by whiplash'. *Accident Analysis & Prevention*, **30**(4), 525-534.
- [22] Ono, K., Kaneoka, K., & Wittek, A. (1997). Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics During Low Speed Rear Impacts, No. 973340. In Proc. 41st Stapp Car Crash Conference (P-315) (pp. 339-356).
- [23] Lopez-Valdes, F.J., Lau, A. *et al.* (2010) January. Analysis of spinal motion and loads during frontal impacts. Comparison between PMHS and ATD. In *Annals of Advances in Automotive Medicine/Annual Scientific Conference* (Vol. 54, p. 61). Association for the Advancement of Automotive Medicine.

- [24] Vijayakumar, V., Scher, I. *et al.* (2006) Head kinematics and upper neck loading during simulated low-speed rear-end collisions: a comparison with vigorous activities of daily living (No. 2006-01-0247). SAE Technical Paper.
- [25] Seacrist, T., Arbogast, K.B. *et al.* (2012) Kinetics of the cervical spine in pediatric and adult volunteers during low speed frontal impacts. *Journal of biomechanics*, **45**(1), pp.99-106.
- [26] Yoganandan, N., Maiman, D.J., Guan, Y. and Pintar, F., 2009. Importance of physical properties of the human head on head-neck injury metrics. *Traffic injury prevention*, **10**(5), pp.488-496.
- [27] Wismans, J.S.H.M., Van Oorschot, H. and Woltring, H.J., 1986. Omni-directional human head-neck response (No. 861893). SAE Technical Paper.
- [28] Kang, Y.S., Stammen, J., Moorhouse, K., Herriott, R. and Bolte IV, J.H., 2016. PMHS Lower Neck Load Calculation using Inverse Dynamics with Cervical Spine Kinematics and Neck Mass Properties. International Research Council on Biomechanics of Injury.
- [29] Acosta, S.M., Ash, J.H., Lessley, D.J., Shaw, C.G., Heltzel, S.B. and Crandall, J.R. Comparison of Whole Body Response in Oblique and Full Frontal Sled Tests. *Proceedings of IRCOBI Conference*, 2016, Malaga, Spain.
- [30] Kang, Y. S., Moorhouse, K., Bolte IV, J. H. Instrumentation Technique for Measuring Six Degrees of Freedom Head Kinematics in Impact Conditions using Six-Accelerometers and Three-Angular Rate Sensors (6aw Configuration) on a Lightweight Tetrahedron Fixture. *Proceedings of 24th International Technical Conference on the Enhanced Safety of Vehicles (ESV)*, 2015 (No. 15-0288), Gothenburg, Sweden.
- [31] Yoganandan, N., Pintar, F. A. *et al.* Upper and lower neck loads in belted human surrogates in frontal impacts. *Proceedings of Annals of Advances in Automotive Medicine/Annual Scientific Conference*, 2012 (Vol. **56**, p.125), Seattle, Washington.
- [32] Philippens, M.M.G.M., Wismans, J., Forbes, P.A., Yoganandan, N., Pintar, F.A. and Soltis, S.J., (2009) ES2 neck injury assessment reference values for lateral loading in side facing seats. *Stapp car crash journal*, **53**, p.421.
- [33] Wismans, J., Philippens, M., Van Oorschot, E., Kallieris, D. and Mattern, R., Comparison of human volunteer and cadaver head-neck response in frontal flexion *Proceedings of STAPP Car Crash Conference*, 1987, New Orleans, Louisiana.
- [34] Wismans, J., Spenny, C. H. Head-neck response in frontal flexion. *Proceedings of STAPP Car Crash Conference*, 1984 (Vol. **28**, pp. 161-171), Chicago, Illinois.
- [35] Yoganandan, N.A., Pintar, F.A. *et al.* Sled tests using the THOR-NT device and post mortem human surrogates in frontal impacts. *Proceedings of IRCOBI Conference*, 2012, Dublin, Ireland.
- [36] Bendjellal, F., Tarriere, C., Gillet, D., Mack, P. and Guillon, F., 1987. Head and neck responses under high G-level lateral deceleration (No. 872196). SAE Technical Paper.
- [37] Kallieris, D., Schmidt, G. and Mattern, R. Vertebral column injuries in 90-degrees collisions. A study with post-mortem human objects. In *Proceedings of the International Research Council on the Biomechanics of Injury conference* (Vol. 15, pp. 189-202). *Proceedings of IRCOBI Conference*, 1987 Birmingham, UK.
- [38] Kallieris, D. and Schmidt, G., 1990. Neck response and injury assessment using cadavers and the US-SID for far-side lateral impacts of rear seat occupants with inboard-anchored shoulder belts (No. 902313). SAE Technical Paper.
- [39] Yoganandan, N. A., Pintar, F. A., Miaman, D., Philippens, M., & Wismans, J. Head kinematics, neck loads, and injuries in side impact sled tests. *Proceedings of IRCOBI Conference*, 2008, Bern, Switzerland.
- [40] Parent, D., Craig, M., & Moorhouse, K. (2017). Biofidelity Evaluation of the THOR and Hybrid III 50th Percentile Male Frontal Impact Anthropomorphic Test Devices. *Stapp car crash journal*, **61**, 227-276.
- [41] Törnqvall, F. V., Holmqvist, K. *et al.* (2008). Evaluation of dummy shoulder kinematics in oblique frontal collisions. *Proceedings of IRCOBI Conference*, 2008, Bern, Switzerland.
- [42] Kent, R., Shaw, G., Lessley, D., Crandall, J., Kallieris, D., & Svensson, M. (2003). Comparison of belted Hybrid III, THOR, and cadaver thoracic responses in oblique frontal and full frontal sled tests (No. 2003-01-0160). SAE Technical Paper.
- [43] Wismans, J. S. H. M., Van Oorschot, H., & Woltring, H. J. (1986). Omni-directional human head-neck response (No. 861893). SAE Technical Paper.
- [44] Myers, B. S., McELHANEY, J. H., Doherty, B. J., Paver, J. G., & Gray, L. (1991). The role of torsion in cervical spine trauma. *Spine*, **16**(8), 870-874.
- [45] Goel, V. K., Winterbottom, J. M. *et al.* (1990). Ligamentous laxity across C0-C1-C2 complex. Axial torque-rotation characteristics until failure. *Spine*, **15**(10), 990-996.

- [46] Myers, B. S., McElhaney, J. H. *et al.* (1989). Responses of the human cervical spine to torsion (No. 892437). SAE Technical Paper.
- [47] Platzer, P., Jaendl, M. *et al.* (2007). Cervical spine injuries in pediatric patients. *Journal of Trauma and Acute Care Surgery*. Feb 1;**62**(2):389-96.
- [48] Self B P, Spittle E K, Kaleps L, and Albery C B. Accuracy and repeatability of the standard automated mass properties measurement system, AL-TR-1992-0137, Armstrong Laboratory, Wright-Patterson AFB, Ohio, 1992.
- [49] Van Ee, C.A. and Nightingale, R.W., 2000. Tensile properties of the human muscular and ligamentous cervical spine (No. 2000-01-SC07). SAE Technical Paper.
- [50] Chancey, V.C., Nightingale, R.W., Van Ee, C.A., Knaub, K.E. and Myers, B.S., 2003. Improved estimation of human neck tensile tolerance: reducing the range of reported tolerance using anthropometrically correct muscles and optimized physiologic initial conditions (No. 2003-22-0008). SAE Technical Paper.

VIII. APPENDIX

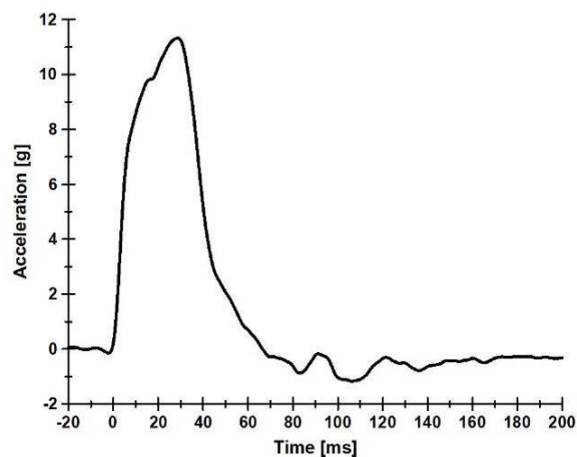
Appendix A

TABLE AI
PMHS HEAD AND NECK ANTHROPOMETRY (unit: cm)

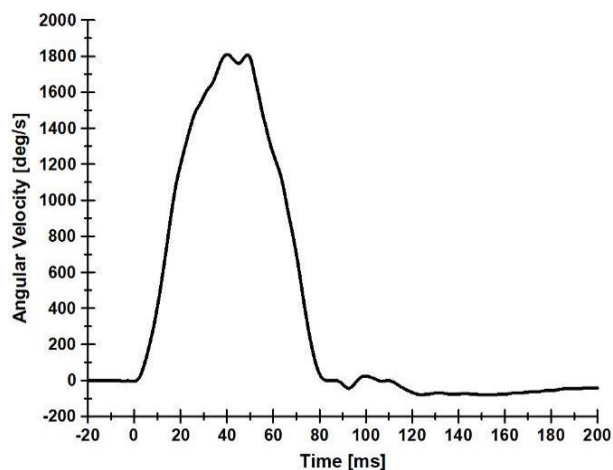
	Head breadth	Head height	Head depth	Head circumference	Neck breadth	Neck depth	Neck circumference
<i>PMHS1</i>	15.5	23.5	19.0	56.4	11.7	10.6	36.5
<i>PMHS2</i>	14.1	22.0	19.0	58.0	12.7	9.2	35.5
<i>PMHS3</i>	15.3	23.0	20.3	61.0	12.7	12.8	43.0
<i>PMHS4</i>	13.5	20.9	18.8	57.0	11.5	8.8	35.5
<i>PMHS5</i>	13.9	23.6	19.0	57.6	11.4	13.5	41.5
<i>Mean</i>	14.5	22.6	19.2	58.0	12.0	11.0	38.4
<i>SD</i>	0.9	1.1	0.6	1.8	0.6	2.1	3.6

TABLE AII
HEAD AND NECK WEIGHT AND MASS MOMENTS OF INERTIA
UNIT: KG AND KG·CM²
HD: HEAD, WN: WHOLE NECK

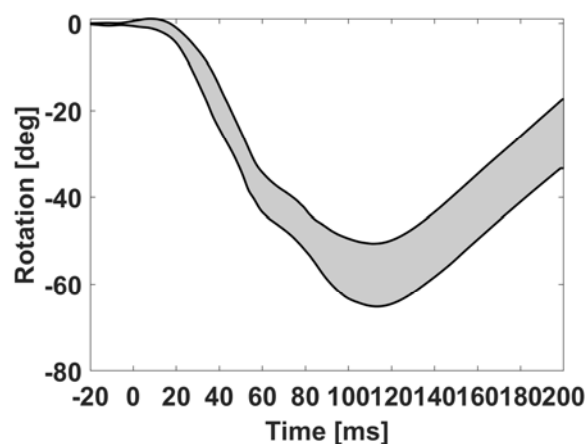
	HD Wt	HD Ixx	HD Iyy	HD Izz	WN Wt	WN Ixx	WN Iyy	WN Izz
<i>PMHS1</i>	3.8	184.8	200.6	143.9	1.4	38.6	33.5	22.4
<i>PMHS2</i>	3.6	129.0	184.3	171.6	1.5	41.1	41.7	27.8
<i>PMHS3</i>	4.3	164.3	234.4	208.4	1.6	46.8	39.8	30.6
<i>PMHS4</i>	3.6	104.3	182.8	158.3	1.4	34.8	31.2	24.4
<i>PMHS5</i>	3.7	116.1	183.6	193.7	1.6	42.0	29.2	35.3
<i>Mean</i>	3.8	139.7	197.1	175.2	1.5	40.7	35.1	28.1
<i>SD</i>	0.3	30.2	19.8	23.3	0.1	4.0	4.9	4.6



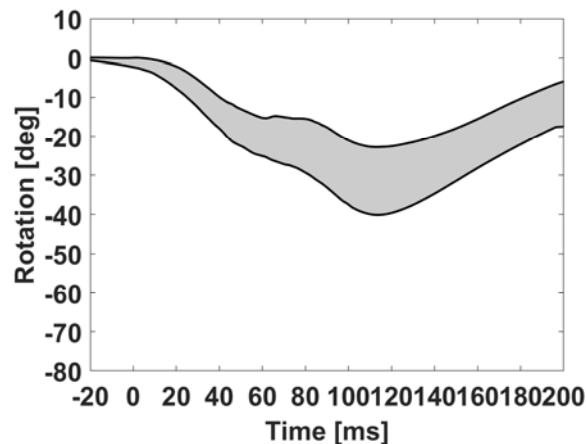
(a) Acceleration for frontal, side and oblique impacts
Fig. A1. Mini-sled inputs.



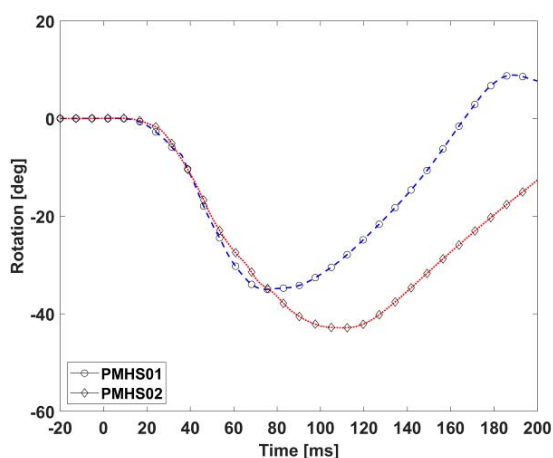
(b) Angular velocity for twist impact



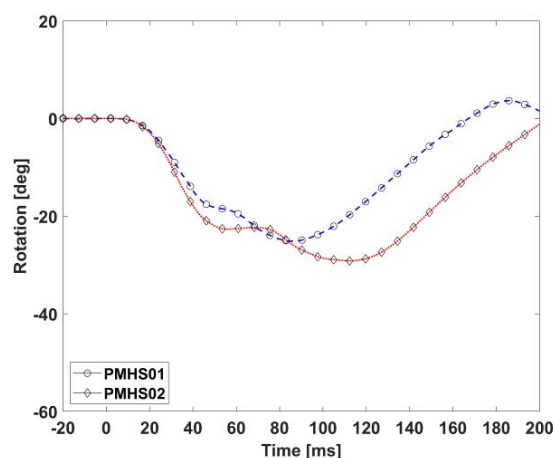
(a) C3 rotation about the y axis
Fig. A2. Cervical rotations in the frontal impact.



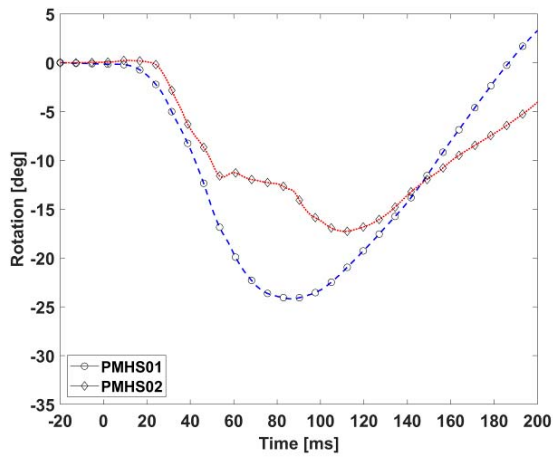
(b) C6 rotation about the y axis



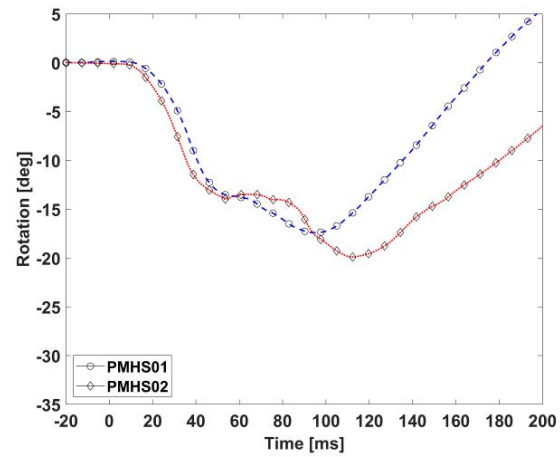
(a) C3 rotation about the x axis
Fig. A3. Cervical rotations in side impacts.



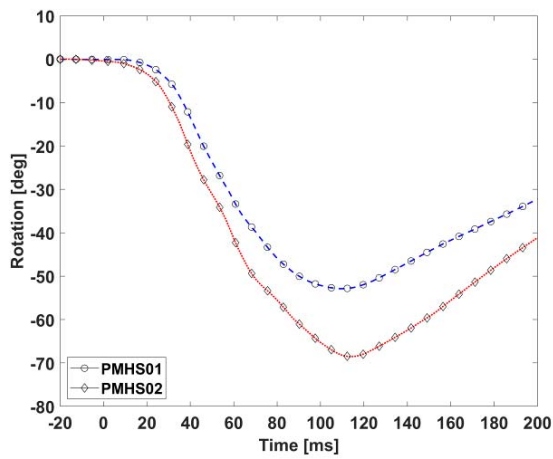
(b) C6 rotation about the x axis



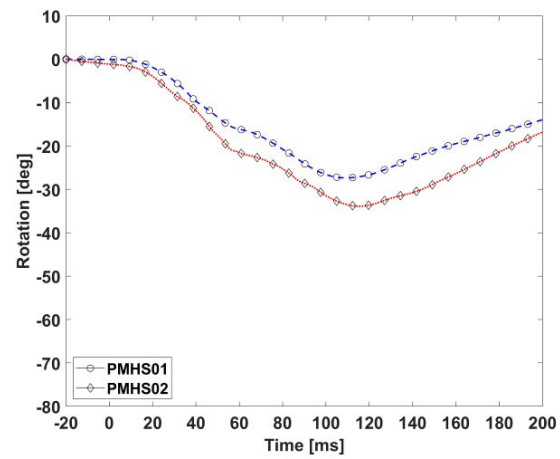
(a) C3 rotation about the x axis



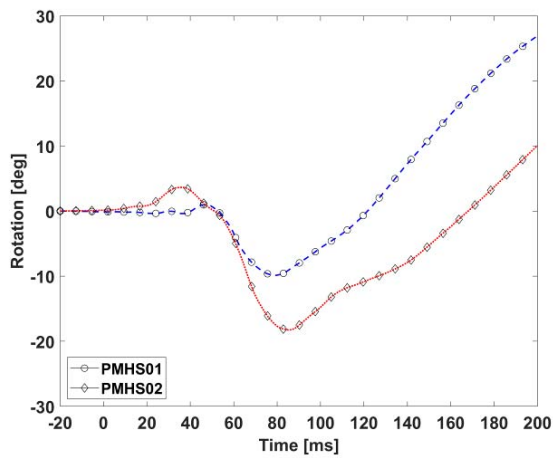
(b) C6 rotation about the x axis



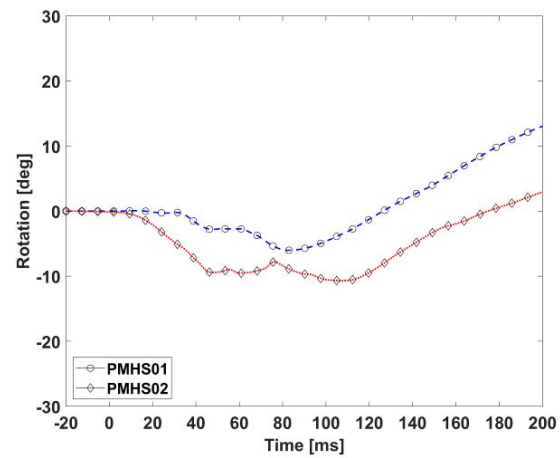
(c) C3 rotation about the y axis



(d) C6 rotation about the y axis

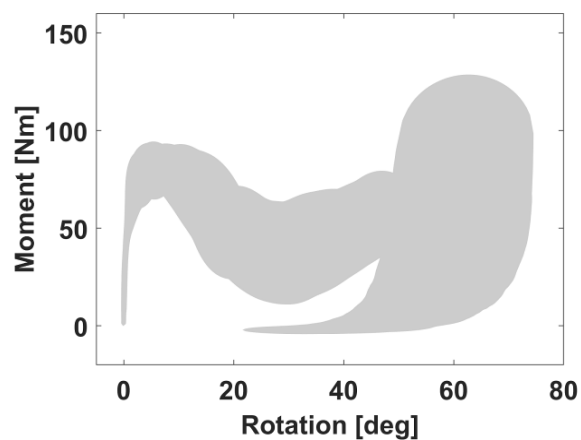


(e) C3 rotation about the z axis

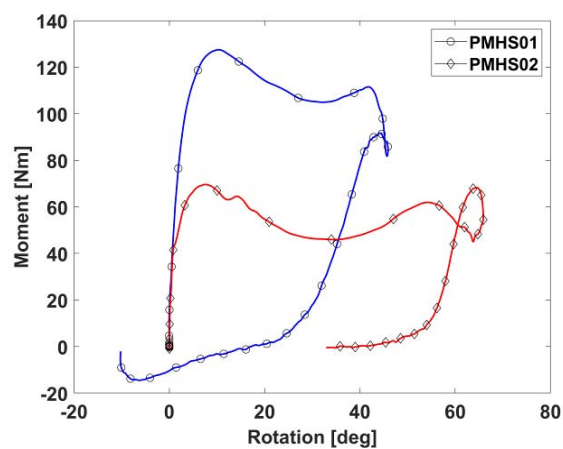


(f) C6 rotation about the z axis

Fig. A4. Cervical rotations in oblique impacts.

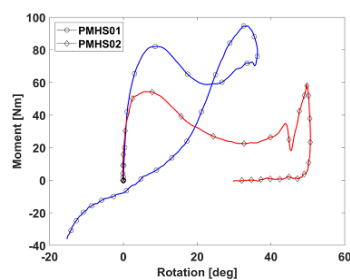


(a) Frontal impact, y axis

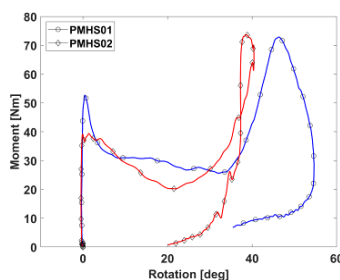


(b) Side impact, x axis

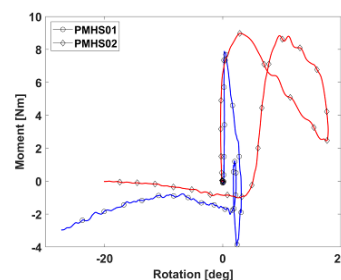
Fig. A5. Lower neck moment vs. head rotation.



(a) x axis



(b) y axis



(c) z axis

Fig. A6. Lower neck moment vs. head rotation in the oblique impact.

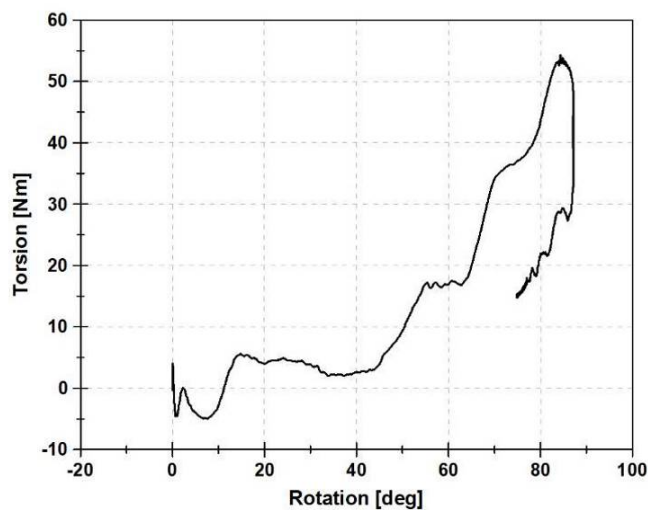
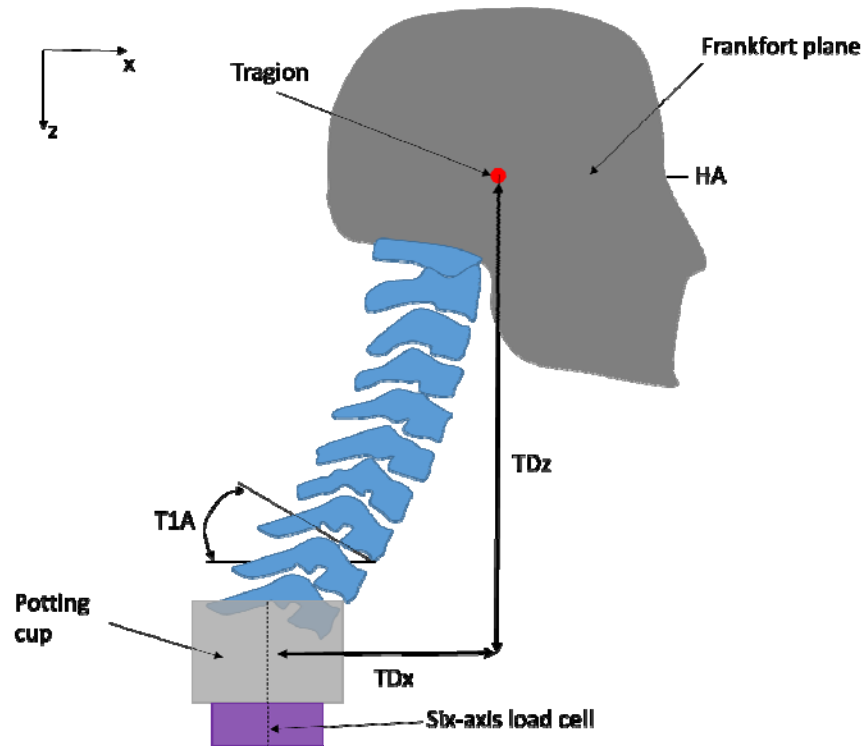


Fig. A7. Lower neck torsion vs. rotation.

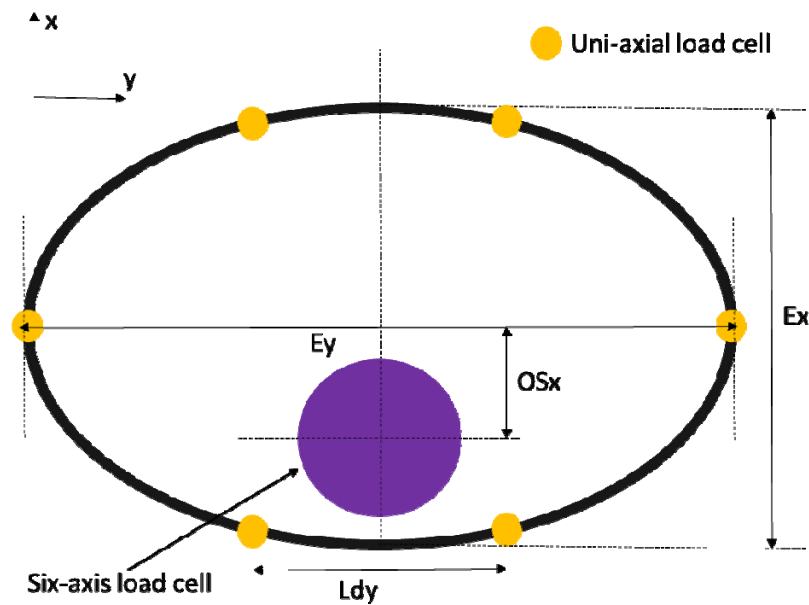
Appendix B



(a) Side view

Head angle (HA): 0 ± 5 deg; T1 angle (T1A): 24 ± 5 deg;

TDx: 8.1 ± 3.4 cm; TDz: 22.9 ± 2.1 cm;



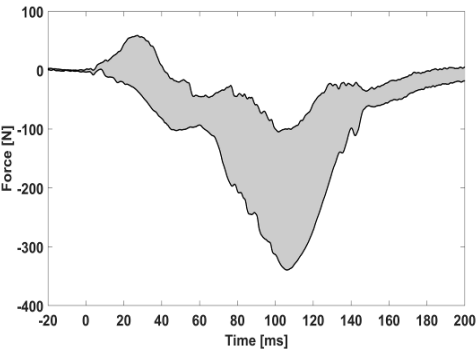
(b) Top view

Ex: 17.2 cm ; Ey: 31.1 cm; OSx: 2.8 cm; Ldy: 10.2 cm

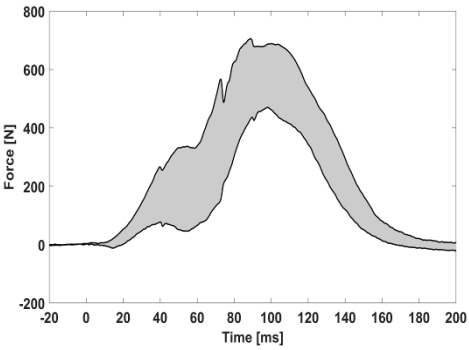
Fig. B1. Information for human body model - head position with respect to center of potting cup and load cell locations. TDx and TDz were determined from FARO measurements for all five PMHS.

TABLE BI
INFORMATION FOR MUSCLE ATTACHMENT TO RING

Anterior neck muscles that were attached to the ring	Posterior neck and back muscles that were attached to the ring
• Sternocleidomastoid muscles • Scalene muscles • Sternohyoid and sternothyroid muscles • Platysma muscle	• Trapezius muscle • Rhomboid major and minor muscles • Serratus posterior superior muscle • Splenius cervicis and capitis muscles • Semispinalis cervicis and capitis muscles • Erector spinae muscles • Levator scapulae

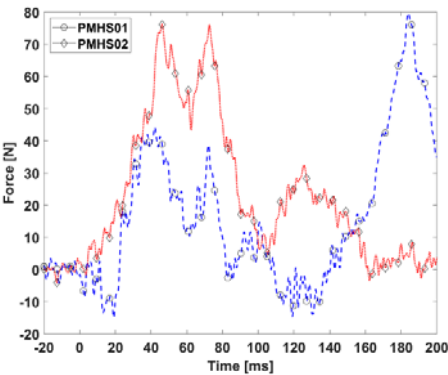


(a) Anterior load cells

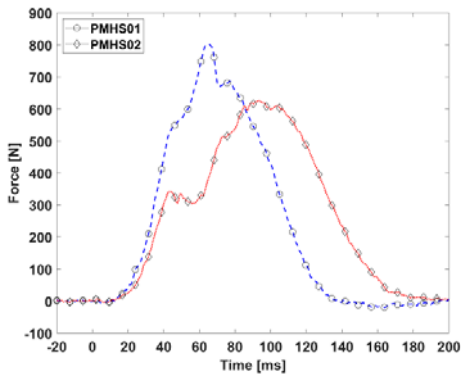


(b) Posterior load cells

Fig. B2. Neck muscle load cells in the frontal impact (+: tension and -: compression)

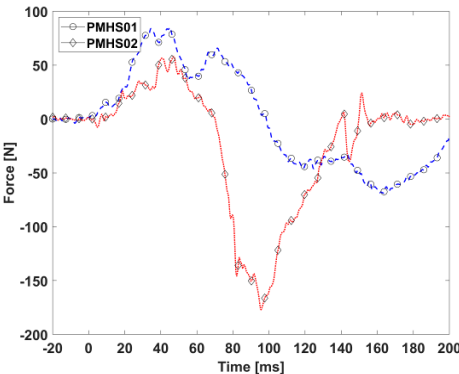


(a) Left side load cells

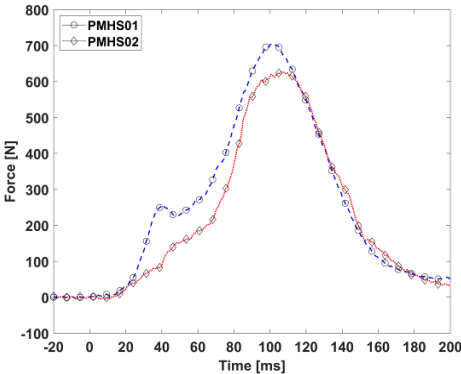


(b) Right side load cells

Fig. B3. Neck muscle load cells in the side impact (+: tension and -: compression)



(a) Anterior load cells



(b) Posterior load cells

Fig. B4. Neck muscle load cells in the oblique impact (+: tension and -: compression)