Evaluating Head and Neck Kinematics and Injury Outcomes During Hyperextension in a Low-Speed Rear Impact Scenario

Reed Finfrock [1], William Millis [2], Angelo Marcallini [1], Colton Thomas [2], Alex Bendig [3], Kevin Moorhouse [2], Yun-Seok Kang [1]

[1] The Ohio State University; [2] National Highway Traffic Safety Administration/VRTC., [3] Transportation Research Center, Inc.

ABSTRACT

Whiplash-associated disorders (WAD) remain common in rear impacts, leading to high societal costs and prolonged recovery. FMVSS 202a includes an optional dynamic rear-impact test that limits head-to-T1 relative rotation (HTTRR) to 12° of extension of the Hybrid III. However, previous BioRID-II studies showed that despite its enhanced biofidelity, the dummy's neck did not exhibit extension even when conditions were adjusted so extension was induced. This lack of extension on the BioRID II neck limited its utility for evaluating segment-specific extension injuries. To better understand extension-based injury criteria, postmortem human subject (PMHS) testing is needed to examine intervertebral kinematics linked to sub-failure injury outcomes in low-speed rear impacts. This study evaluates cervical spine kinematics and injury mechanisms using intervertebral motion and injury findings using six 50th-percentile male PMHS. Instrumentation included 6aw at the head and motion blocks at C2–T1. PMHS were positioned in a rigid seat without a head restraint to allow full neck extension. A pendulum-induced impact produced an average ΔV of 9 km/h. HTTRR ranged from 45° to 66°. Neck retraction and S-shaped cervical spine curvature preceded peak extension and rebound into flexion. Intervertebral extension ranged from ~5° to ~10°, with C4–C5 most frequently exhibiting the largest values. Segments that exhibited the greatest combined intervertebral motion frequently corresponded with soft tissue damage identified during post-test dissection. In total, segmental laxity was identified at 9 of 36 levels across 6 subjects. These results emphasize the importance of measuring intervertebral rotation to identify sub-failure cervical spine injury. Segment-level motion may serve as a more sensitive injury predictor than gross kinematics alone. Future studies can use these data to improve the biofidelity of anthropomorphic test devices and human body models for low-speed rear impacts.

INTRODUCTION

Whiplash-associated disorders (WAD) remain common in rear impacts, leading to high societal costs and prolonged recovery [1], [2]. Of these, cervical spine injuries are the most prevalent source of chronic and persistent pain following motor vehicle accidents [2]. Challenges are often encountered when diagnosing cervical spine injuries because neck pain is frequently reported in the absence of radiological findings, as symptoms can manifest even in the absence of fractures or visible structural damage [3]. Current diagnostic standards consider WAD grade classifications in evaluating the extent of the injury [4]. WAD classifications range from WAD1 to WAD4, of which only grades of WAD4 exhibit radiological evidence of fractures or

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dislocations. To reduce the risk of WAD in rear impacts, neck injury thresholds should be sensitive enough to detect soft tissue injuries. Extensive studies have evaluated the potential mechanisms of injury and pain propagation as well as developed criteria for predicting the occurrence of these injuries [5-14].

The Federal Motor Vehicle Safety Standard (FMVSS) 202a aims to mitigate WAD by permitting an optional dynamic rear-impact test in the United States. As part of this test, head-to-T1 relative rotation (HTTRR) measured with the Hybrid III dummy is limited to 12° of neck extension. This threshold was originally established through a study that evaluated HTTRR in production seats and linked those measurements to real-world injury data to assess seat safety performance [14], [18]. EuroNCAP and IIHS include additional metrics such as Neck Injury Criterion (NIC), normalized neck injury criterion (Nkm), upper and lower neck forces and moments, T1 acceleration, and timing of head-to-head restraint contact [15-17]. The criteria described above don't address soft tissue injuries, which are prevalent in the field despite the utilization of these metrics [14].

Previous studies examined BioRID-II's ability to replicate cervical spine motion and found that, despite its detailed neck components, the ATD's neck did not exhibit extension in a production seat under similar loading conditions, even with a large backset, which is a horizontal distance between the back of the head and the front of the head restraint [19]. This was attributed to the initial neck retraction phase, where the head rotates minimally due to the difference in inertia between the head and torso before contacting the head restraint. These efforts demonstrated that the Hybrid III's extension limit could not be applied to the BioRID-II when evaluating production seat and head restraint performance [19]. Subsequent testing successfully induced gross neck extension of the BioRID-II by introducing excessive backsets, but extension occurred in only some intervertebral segments, preventing the evaluation of intervertebral extension limits [19]. To establish the BioRID-II as a biofidelic surrogate, it is essential to evaluate its mechanical performance in areas where it more accurately replicates cervical vertebra. This comparison would help identify parameters with improved utility relative to those used for the Hybrid III. Furthermore, leveraging the BioRID-II's detailed anatomical structure could enhance injury prediction by enabling vertebral-level analysis of neck extension and its relationship to sub-failure soft tissue injuries. Therefore, neck extension responses and injury criteria for the BioRID-II should be investigated through targeted PMHS testing to define extension-based metrics that align with injury outcomes in low-speed rear impacts.

This study aimed to capture intervertebral kinematics during full-body PMHS testing to evaluate potential injury predictors beyond head-to-T1 rotation (HTTRR). By analyzing motion at the vertebral level, the study addresses gaps in current injury criteria and enables more precise evaluation of cervical motion patterns. Direct measurement of intervertebral kinematics allows for improved assessment of soft tissue injury risk, providing a foundation for developing enhanced thresholds that reflect sub-failure injuries during neck extension.

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METHODS

Sled Set-up

The sled system included a rigid, adjustable seat without a head restraint to allow full neck extension. The height of the telescoping seatback (Figure 1A) was adjusted to align the top of the seatback with the PMHS T8 vertebra and secured in place. The contact surface was rounded to avoid load concentration at edges. The seatback and seat pan were fixed at 24° (Figure 1B) from vertical and 19° from horizontal (Figure 1C), respectively.

Mounted on a raised rail system, the mini-sled traveled forward and was halted by a spring-damper stopper (Figure 1D). PMHS were positioned by following a seating procedure used in a previous study [19], where modifications were required due to the elevated foot positioning as well as restrictions related to having a rigid seat. The subject head was held in place using support cables which were cut upon contact with the sled (Figure 1E). A vertical support pole held a D-ring setup (Figure 1F) for restraint tensioning (4 lbs lap, 6 lbs shoulder). Figure 1 shows the rigid seat, sled, and initial seated PMHS posture achieved for an example test.

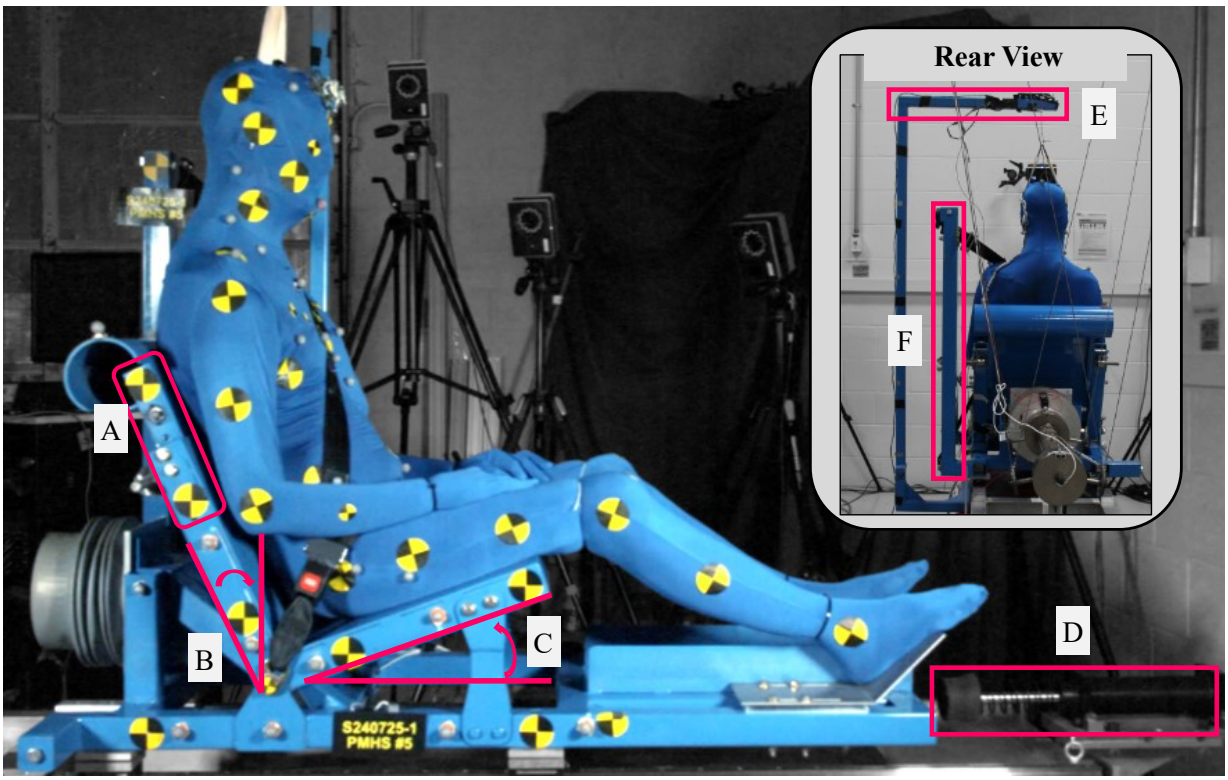
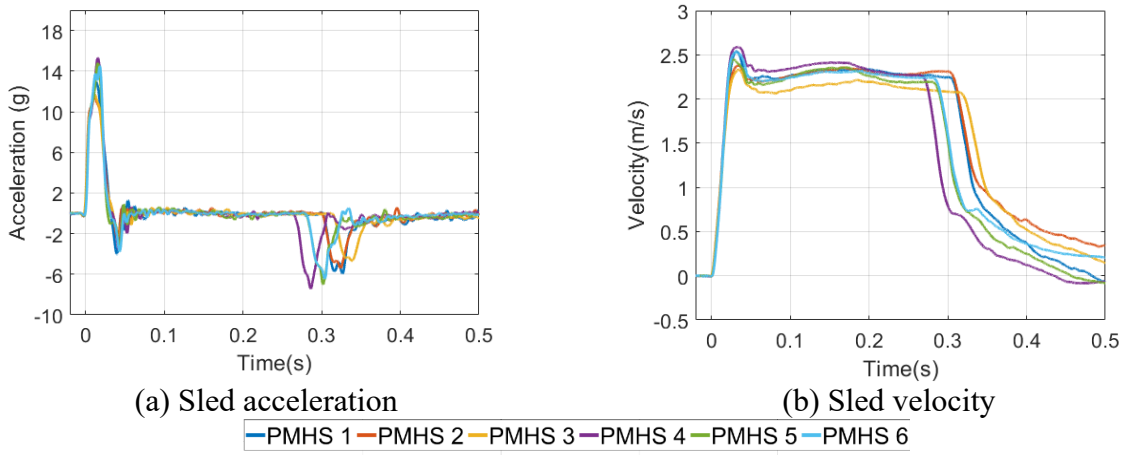


Figure 1: Seat configuration and seated posture.

Sled pulses were generated using a pendulum (37.6 kg) released from a controlled height, producing impact speeds of 7 m/s for the first three PMHS and 8 m/s for the second set. Upon pendulum impact, the stationary sled translated along the rails with peak accelerations ranging from 12.3–14.9 g with an average ΔV of 2.46–2.50 m/s (8.8–9.0 km/h). These low-speed pulses

aimed to produce sufficient neck extension without exceeding BioRID II rotational limits, preserving comparability for future studies.



(a) Sled acceleration (b) Sled velocity

Figure 2: Sled velocity and acceleration profiles.

PMHS Information

Six mid-size male PMHS were selected for use in this study to enable comparison against the BioRID II dummy under matched input conditions. PMHS height and weight were within the 50th percentile male range to support anthropometric consistency. PMHS were screened using dual-energy X-ray absorptiometry (DXA), and those exhibiting abnormal cervical spine conditions—such as pronounced cervical osteophytes or degenerative disc disease—were excluded. General PMHS characteristics are provided in Table 1, while head and neck specific anthropometry is summarized in Table 2.

Table 1: PMHS information

Subject	Age (yrs.)	Height (cm)	Weight (kg)	BMI (m/kg^2)	aBMD T-Score L2-L4
PMHS1	61	176.6	60.1	19.3	-1.2
PMHS2	81	171.8	67.0	22.7	0.0
PMHS3	54	179.0	75.8	23.7	-2.4
PMHS4	87	177.7	63.7	20.2	-1.9
PMHS5	63	182.7	75.3	22.6	0.9
PMHS6	71	192.0	76.7	20.8	1.0
Mean (SD)	69.5 (12.6)	180.0 (6.9)	69.8 (7.1)	21.5 (1.7)	0.9 (0.6)
50th Male[20]	-	175.3	77.30	25.2	-

Table 2: PMHS 1-6 neck specific anthropometry (cm)

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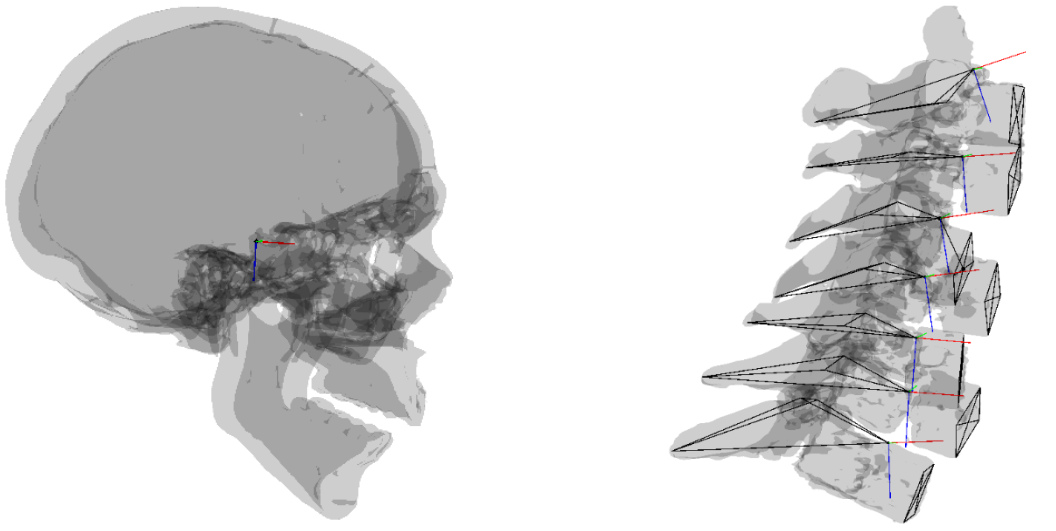
Subject	Head Height	Head Length	Head Breadth	Head Circ.	Neck Breadth	Neck Depth	Neck Circ.
PMHS1	23.6	19.6	13.5	57.2	9.6	9.6	34.0
PMHS2	22.4	18.6	14.8	58.0	9.7	10.3	39.0
PMHS3	21.7	18.2	12.9	55.1	10.1	12.3	40.6
PMHS4	23.1	19.2	14.3	57.2	9.2	11.2	36.7
PMHS5	23.9	19.6	16.0	58.5	10.7	17.7	38.0
PMHS6	22.7	20.2	13.5	58.0	8.5	14.9	38.4
Mean (SD)	22.9 (0.8)	19.2 (0.7)	13.9 (0.7)	57.2 (1.2)	9.6 (0.8)	12.7 (3.1)	37.8 (2.3)
50th Male [20]	23.1	19.7	15.8	57.1	11.4	11.5	38.3

PMHS Instrumentation

Head kinematics were collected utilizing the 6aω tetrahedron assembly [21] and transformed to the head center of gravity. Acceleration and angular rate data were collected, with the primary acceleration occurring along the x-axis. Head angular rate velocities about the y-axis were integrated to calculate rotations for evaluation of HTTRR.

PMHS cervical spine (C2–C7) and T1 were instrumented with six-degree-of-freedom (6DOF) motion blocks placed on each vertebral body [22]. The orientation of the blocks relative to the vertebral body was defined using computed tomography (CT) scans for transformation of the data from the blocks to local kinematics of the vertebra.



(a) Head coordinate system (b) Cervical coordinate systems

Figure 3: Local coordinate system definitions: red (x), green (y), blue (z).

Data Collection and Analysis

Data were collected using a SLICE PRO data acquisition system (Diversified Technical Systems [DTS], Seal Beach, CA) at a sampling rate of 20,000 Hz. The data were then trimmed to focus on responses from the main event. All signals were filtered in accordance with SAE J211 standards.

Pre-test and post-test X-rays, as well as CT scans, were captured to aid in evaluation of PMHS conditions and instrumentation locations.

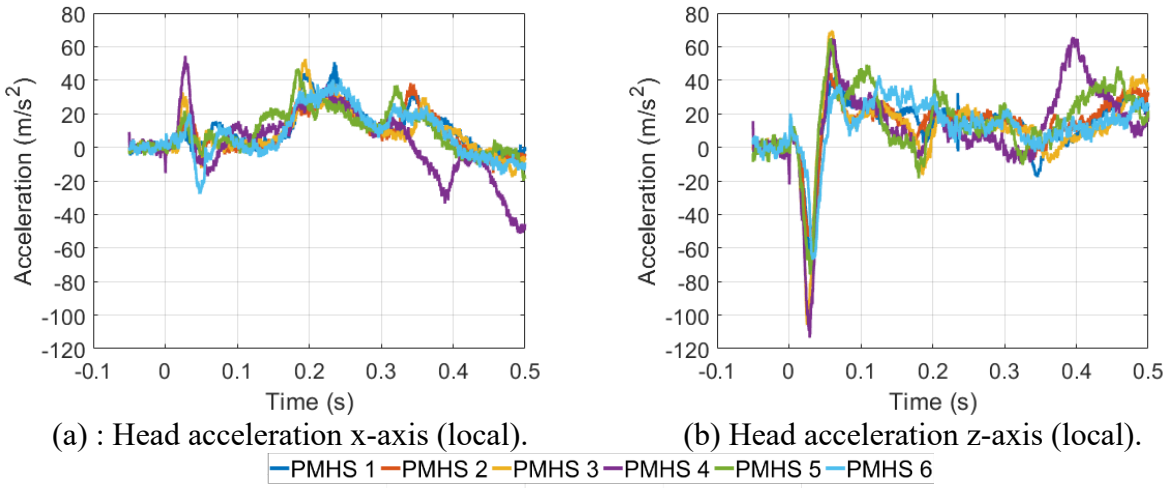
Detailed post-test dissections were conducted to assess injuries from all body regions, while the primary focus was isolating the cervical spine. Manual flexion and extension motions were applied to the isolated ligamentous cervical spine at each intervertebral level to evaluate soft-tissue sub-failure injuries and intervertebral instability. Intervertebral laxity was defined as an abnormal increase in mobility, frequently accompanied by visible tissue differences.

Injury outcomes were compared to segment-specific kinematics obtained during testing to evaluate the predictive capability of vertebral kinematics for soft tissue injuries. To preliminarily assess which parameter best predicted injury presence, binary logistic regression was conducted using Minitab. The analysis included calculations of Goodman-Kruskal Gamma (GKG) scores, the area under the receiver operating characteristic curve (AUROC), and associated p-values.

RESULTS

Head Kinematics

Figure 4 shows the x- and z-axis acceleration for all PMHS.



(a) : Head acceleration x-axis (local). (b) Head acceleration z-axis (local).

Figure 4: Head acceleration in x- and z-axes.

Head rotation occurred primarily about the y-axis, which are presented in Figure 5.

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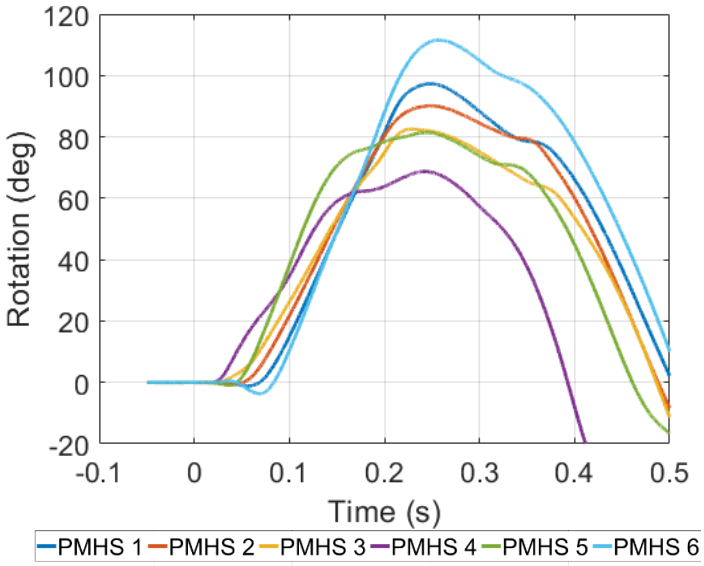
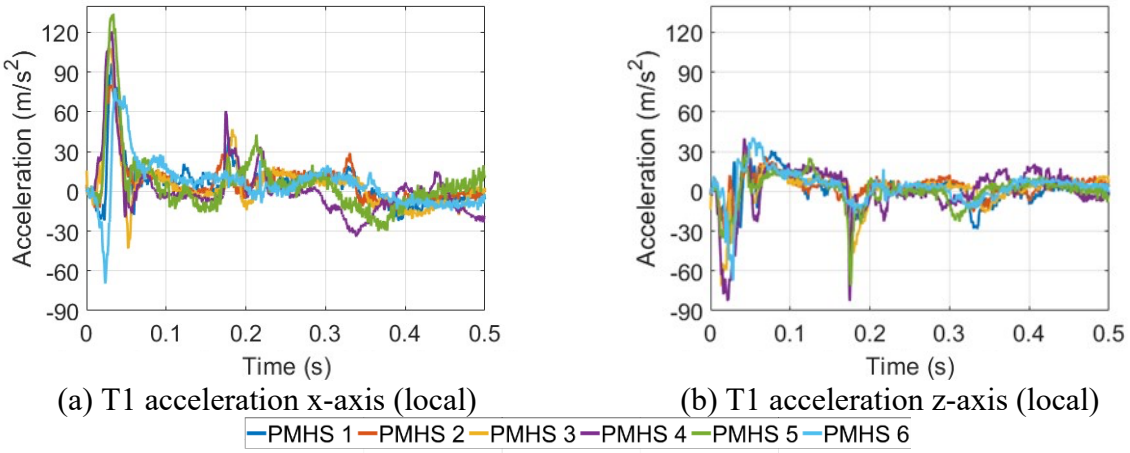


Figure 5**:** Head rotation about the y-axis.

Vertebral Kinematics

Figure 6 illustrates the primary x- and z-axis accelerations acting at T1 throughout the event.



(a) T1 acceleration x-axis (local) (b) T1 acceleration z-axis (local)

Figure 6: T1 accelerations in primary x- and z-axes.

To evaluate propagation of the impulse through the cervical spine and then into the head, vertebral rotation about the y-axis is displayed for each subject in Figure 7.

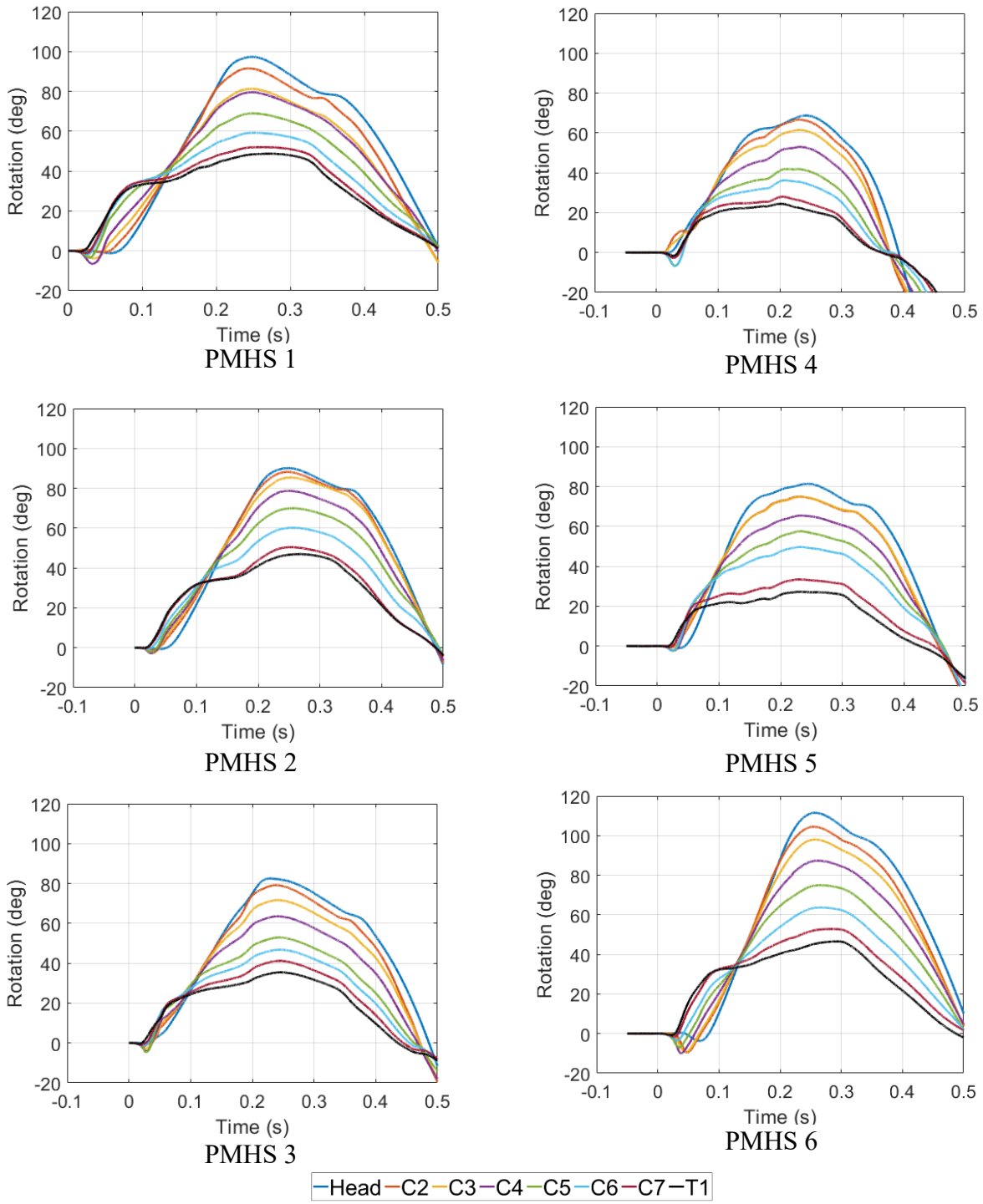


Figure 7: Vertebral rotations about the y-axis.

Head to T1 Relative Rotation

To evaluate the overall motion of the neck complex during extension-based testing, Figure 8 shows the HTTRR while Table 3 shows the corresponding peak values for extension as well as the total range from the flexion peak to the extension peak.

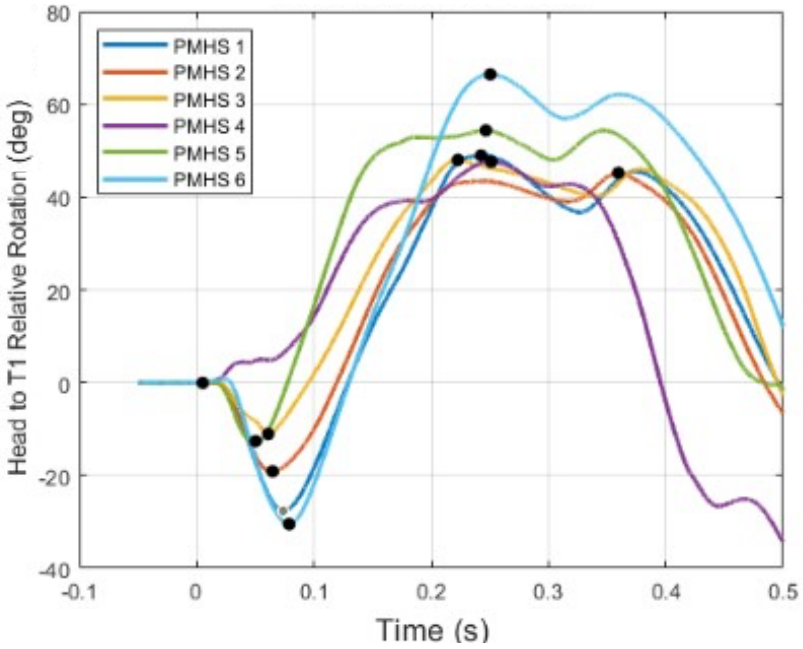


Figure 8: Head to T1 relative rotations.

Table 3: Peak values for HTTRR and timing

Subject	Peak Extension (°)	Peak Extension Times (s)	Peak Flexion (°)	Peak Flexion Times (s)	Total Range (°)
PMHS 1	49.0	0.243	-27.6	0.074	76.6
PMHS 2	45.2	0.360	-19.1	0.065	64.3
PMHS 3	48.0	0.223	-11.1	0.061	59.1
PMHS 4	47.6	0.251	-0.10	0.005	47.7
PMHS 5	54.4	0.247	-12.6	0.050	67.0
PMHS 6	66.4	0.250	-30.5	0.079	96.9
Mean (SD)	51.8 (7.8)	0.262 (0.049)	-16.8 (11.3)	0.056 (0.027)	68.6 (16.8)

Intervertebral Kinematics

To evaluate injury with respect to relative motion of the vertebrae, intervertebral rotation data is presented in Figure 9. These rotations were calculated by subtracting the inferior vertebra's

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rotation from the superior vertebra's (e.g., C2–C3). Positive values indicate additional rotation at superior levels, signifying intervertebral extension, while negative values denote intervertebral flexion. In the figures below, the circles represent the peak rotations at each level, with black open circles representing peaks where no injury observed and red filled circles represent peaks associated with sub-failure injuries observed during post-test dissection.

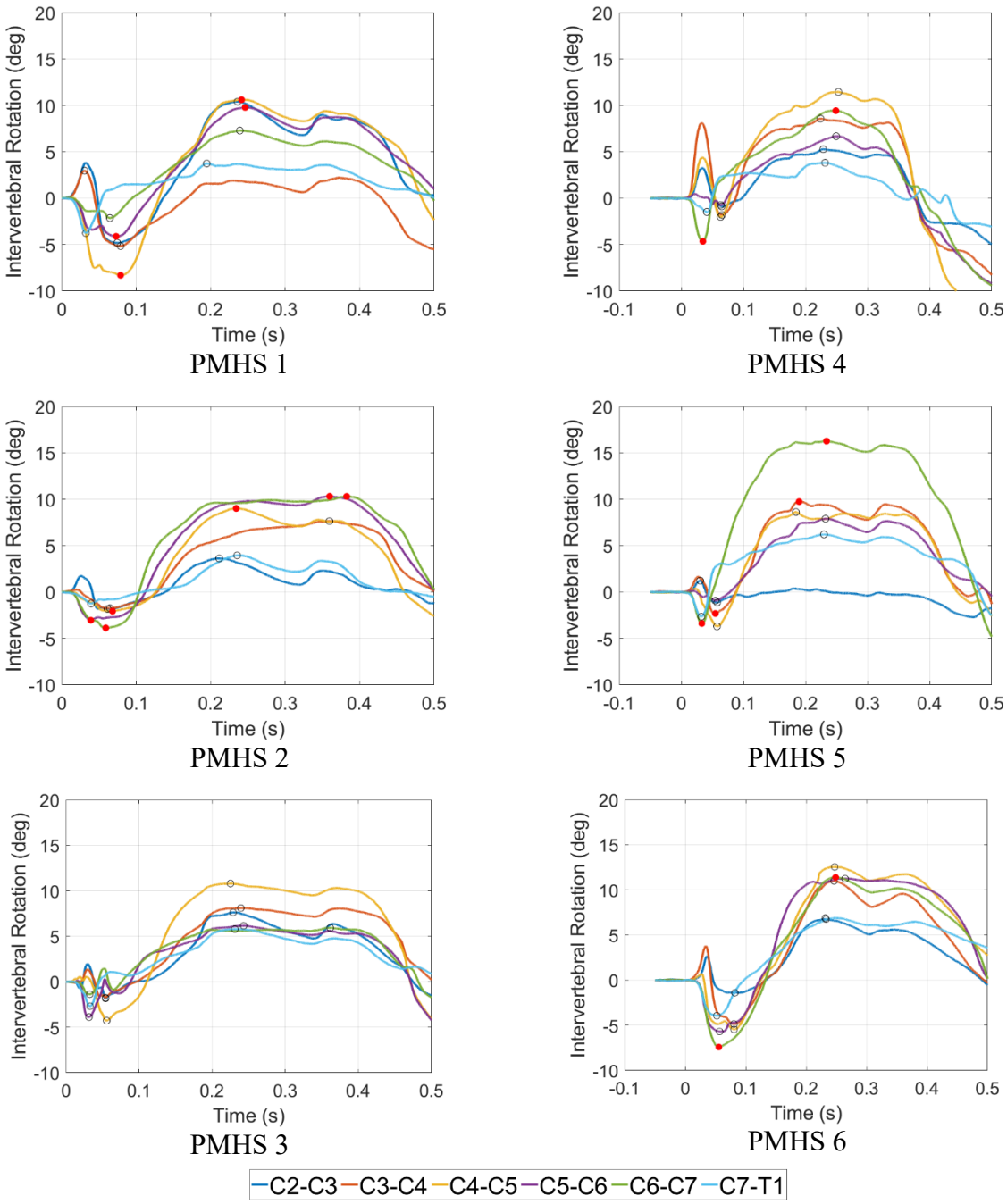

Figure 9: Intervertebral rotations.

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Table 4 shows the peak values observed for intervertebral extension. Table 5 shows the peak values observed for intervertebral flexion. Table 6 considers the range from the intervertebral flexion peak to the extension peak.

Table 4: Peak values for intervertebral extension

	C2-C3 (°)	C3-C4 (°)	C4-C5 (°)	C5-C6 (°)	C6-C7 (°)	C7-T1 (°)
PMHS1	10.4	3.0	10.6	9.8	7.3	3.7
PMHS2	3.6	7.6	9.0	10.3	10.3	4.0
PMHS3	7.6	8.1	10.8	6.2	5.9	5.8
PMHS4	5.3	8.6	11.5	6.7	9.4	3.8
PMHS5	1.2	9.8	8.6	7.9	16.3	6.2
PMHS6	6.8	11.0	12.6	11.3	11.4	6.9
Mean (SD)	5.8 (3.2)	8.0 (2.7)	10.5 (1.5)	8.7 (2.1)	10.1 (3.6)	5.1 (1.4)

Table 5: Peak values for intervertebral flexion

	C2-C3 (°)	C3-C4 (°)	C4-C5 (°)	C5-C6 (°)	C6-C7 (°)	C7-T1 (°)
PMHS1	-4.8	-5.2	-8.3	-4.1	-2.1	-3.8
PMHS2	-1.8	-1.7	-2.1	-3.0	-3.9	-1.2
PMHS3	-1.8	-1.8	-4.3	-3.9	-1.4	-2.7
PMHS4	-0.9	-1.8	-2.0	-0.7	-4.6	-1.5
PMHS5	-1.1	-2.3	-3.7	-0.9	-3.4	-2.6
PMHS6	-1.4	-4.9	-5.4	-5.7	-7.4	-3.9
Mean (SD)	-2.0 (1.4)	-2.9 (1.6)	-4.3 (2.4)	-3.1 (1.9)	-3.8 (2.1)	-2.6 (1.1)

Table 6: Total peak flexion to peak extension intervertebral rotation range

	C2-C3 (°)	C3-C4 (°)	C4-C5 (°)	C5-C6 (°)	C6-C7 (°)	C7-T1 (°)
PMHS1	15.2	8.1	18.9	13.9	9.4	7.5
PMHS2	5.5	9.4	11.1	13.4	14.2	5.2
PMHS3	9.4	9.9	15.1	10.1	7.3	8.5
PMHS4	6.2	10.4	13.5	7.4	14.1	5.3
PMHS5	2.4	12.1	12.3	8.8	19.7	8.9
PMHS6	8.2	15.9	18.0	16.9	18.8	10.8
Mean (SD)	7.8 (4.3)	11.0 (2.7)	14.8 (3.1)	11.8 (3.6)	13.9 (4.9)	7.7 (2.2)

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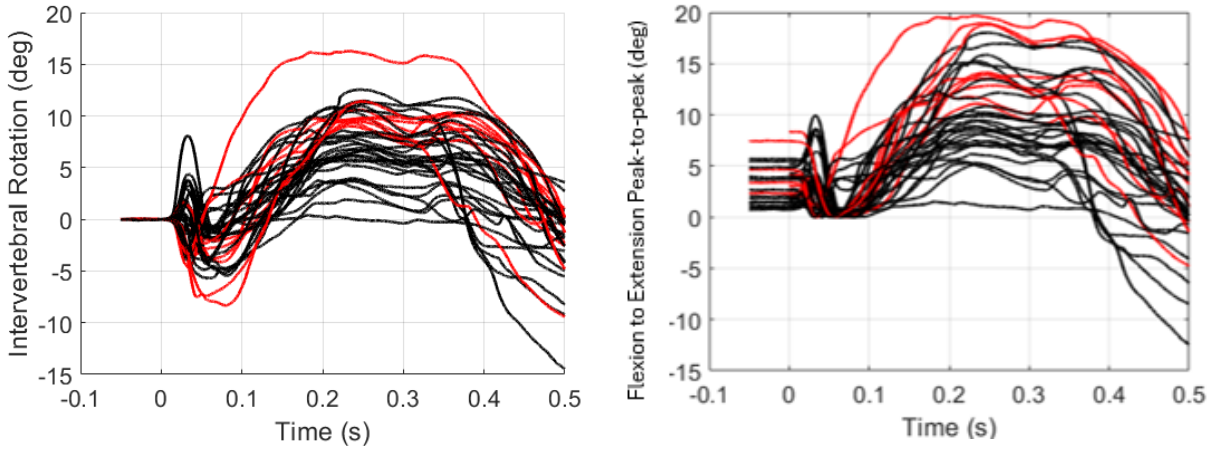
Injury Criteria

Using tetrahedron data transformed to the occipital condyles, forces and moments were calculated, which were then used to compute the Nkm injury criteria. Table 7 shows the peak values and the times at which extrema occur for both NIC and Nkm injury criteria.

Table 7: NIC and Nkm extrema and timing

Subject	NIC Max	NIC Max Times (s)	Nkm Max	Nkm Max Times (s)
PMHS 1	18.15	0.032	0.468	0.235
PMHS 2	12.63	0.030	0.355	0.216
PMHS 3	17.49	0.031	0.368	0.222
PMHS 4	15.37	0.033	0.304	0.028
PMHS 5	26.20	0.034	0.380	0.182
PMHS 6	19.55	0.047	0.453	0.233
Mean (SD)	18.30 (4.59)	0.035 (0.006)	0.388 (0.062)	0.186 (0.080)

To evaluate the impact of intervertebral rotation on injury outcomes, Figure 10 displays the rotation magnitudes from each test plotted together. Figure 10 (a) presents the intervertebral rotation magnitudes. In contrast, Figure 10 (b) shows the full range of motion by shifting each signal so that its flexion (negative) peak aligns with zero. This method emphasizes the total excursion of each segment, combining both flexion and extension.



(a) Intervertebral extension

(b) Intervertebral peak flexion to peak extension range

Figure 10: Intervertebral rotation and intervertebral peak-to-peak range.

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Table 8 shows the statistical results evaluating the quality of predictive capability for each metric.

Table 8: Statistical evaluation considering binary regression of predictive variables

Predictive Variables	AUROC	GKG	P-Value
Flexion to Extension Peak-to-Peak	0.86	0.71	0.006
Intervertebral Extension	0.82	0.65	0.014

Injury

Dissection findings revealed signs of laxity in 9 of the 36 evaluated intervertebral levels. PMHS 4 sustained a partial tear of the anterior longitudinal ligament (ALL). Shown in Table 8 is a summary of all the observed injuries.

Table 8: Injury outcomes for each PMHS

	PMHS1	PMHS2	PMHS3	PMHS4	PMHS5	PMHS6
C1-C2	-	-	-	-	-	-
C2-C3	-	-	-	-	-	-
C3-C4	-	-	-	-	Laxity	-
C4-C5	Laxity	Laxity	-	-	-	-
C5-C6	Laxity	Laxity	-	-	-	-
C6-C7	-	Laxity	-	Partial ALL Tear	Laxity	Laxity
C7-T1	-	-	-	-	-	-

DISCUSSION

Head and T1 Accelerations

To evaluate neck kinematics, it is useful to evaluate accelerations that are transmitted through the boundaries of the neck.

At impact, initial head acceleration exhibited a localized positive peak, followed by a secondary negative peak as the head swung rearward relative to the torso. The primary peak occurred along the z-axis, indicating upward head acceleration as the torso was rapidly accelerated forward. These head acceleration peaks, shown in Figure 4b, coincided with peak anterior (x-axis) acceleration at T1, as illustrated in Figure 6a. Prior to T1 x-axis forward acceleration there is a negative peak, likely dependent on initial T1 orientation. This response has been described in previous studies because of spinal straightening, where T1 and other thoracic vertebrae are rapidly thrust out of its natural kyphotic curvature into a more linear alignment [23]. During this early phase, before the acceleration reaches the head, the neck experiences

compressive loading which explains the initial vertical acceleration observed prior to the dominant anterior displacement caused by the sled motion.

Head Rotation

Head rotation, when considered alongside T1 motion, helps quantify the distribution of motion across the neck and estimate HTTRR. As shown in Figure 5, head rotation occurred primarily about the y-axis, with total rotation ranging from 70° to 110°, coinciding with peak vertebral extension. These values exceed those reported in volunteer studies by Sato et al. (2014) with peak accelerations around 42 m/s² (4.3 g) and ΔV of 1.61 m/s, where peak head rotations ranged from 40–60°, likely due to the decreased pulse and potentially impacted by muscle activation or anticipation [24]. Because T1 rotation is largely governed by kyphotic curvature and seat interaction, head rotation magnitude is a key driver of HTTRR and is indicative of the overall severity of neck loading.

Vertebral Angular Kinematics

In Figure 7, peak rearward rotation of T1 ranged from 20°–50°, and then a clear sequential distribution can be observed from T1 to the head, indicating a progressive motion pattern throughout the cervical spine. Certain vertebral levels may undergo relatively greater motion, influenced by subject-specific anatomy or segmental stiffness, or possible soft tissue damage.

Head to T1 Relative Rotations

Historically, HTTRR has been considered a potential risk factor for cervical spine injury, under the assumption that greater relative head to T1 motion corresponds to increased injury likelihood. The present findings support the idea that HTTRR reflects the cumulative rotation between the head and T1, whereas individual vertebral rotations shown in Figure 7 represent discretized contributions to that total, generally scaling in proportion to HTTRR.[14], [15]

Peak HTTRR values, as shown in Figure 8 and Table 3, coincided with peak cervical extension (Figure 7), ranging from 45–66°. These magnitudes align with previous studies [24]. For instance, Sato et al. (2014) reported HTTRR with peaks of approximately 40° in volunteer tests [24]. Similarly, Kang et. al. reported positive HTTRR values from 35–45° but may have been limited by contact with the head restraint and interaction with the production seat [19]. This study, however, reported similar timing for the transition from negative to positive HTTRR, suggesting that the initial negative HTTRR values and initial segmental flexion are consistent in the well-controlled loading condition with no head restraint in this study as well as in production seats at a higher severity [19].

The injury outcomes observed in this study suggest that the injury threshold may lie between 40–66° of HTTRR [19], with consideration being given to the loading rate in which the neck is forced into extension.

This study observed a total HTTRR of 48–100° occurring over 250ms during injurious tests. In comparison, Kang et al. [19] reported high HTTRR magnitudes during moderate-speed testing, with negative peaks approaching –30° and positive peaks of 35–45°, resulting in a total range of 65–75° over a shorter duration of 150ms. In contrast, volunteer studies under low-speed conditions have shown slower rotational rates with peaks at the lower end of current testing at approximately 50° over 250ms [24]. These conditions are associated with more uniform intervertebral rotation and help explain a lack of injury. Taken together, these findings suggest that injury risk may not depend solely on the magnitude of HTTRR but also rate of rotation. Furthermore, even smaller HTTRR values could lead to segmental-level injuries if the motion is non-uniform and concentrated stress is applied to specific cervical segments.

Five of the six tests exhibited neck retraction as initial forward HTTRR (Figure 8), displaying rearward torso rotation paired with near-zero or slight forward head rotation, consistent with previous study [23]. During the onset of loading, five of the six PMHS shifted into an S-shaped neck curvature as described in previous literature [23], followed by peak extension and a rebound into flexion, while PMHS 4 bypassed the initial forward HTTRR phase as the neck proceeded directly into extension. This may be a result of anatomical variation, as the neck has pre-existing rearward curvature combined with a higher chin orientation in seated posture, which may have altered inertial interactions between the head and neck.

One of the primary research objectives was to further evaluate HTTRR and its relationship with injury likelihood [14], [15]. Only one subject, PMHS 3, did not exhibit indicators of sub-failure injury. Although PMHS 3 demonstrated lower HTTRR values (48° peak, 57° total rotation), these were not the lowest values observed.

While a definitive HTTRR injury threshold remains unclear, the relationship between HTTRR and intervertebral rotation offers additional insight into the relationship between discrete cervical kinematics and injury outcomes.

Intervertebral Kinematics

Higher HTTRR represents a greater overall difference between head and torso motion, and this cumulative value is comprised of the individual rotations at each intervertebral level. Rather than a single segment contributing disproportionately, the rotation magnitudes were generally shared across the cervical spine. The intervertebral rotations scale with the total HTTRR, acting as discrete components that sum to the overall head-to-torso motion. This trend is demonstrated in Figure 7 where greater differences between head and T1 rotation require larger gaps between subsequent levels to bridge the difference. This reinforces the idea that elevated HTTRR reflects proportionally greater demands placed on each cervical segment.

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Across segments, peak intervertebral rotations generally ranged from approximately 5° to 10°, with the C4–C5 segment most frequently exhibiting the highest peak values. This trend is consistent with findings from a similar volunteer study utilizing a rigid seatback and no head restraint [24]. The presence of injury in the current study suggests a difference in loading or loading rate between the two scenarios, in which the low-speed volunteer testing lacked the greater degree of initial intervertebral flexion present in the current study and in the moderate speed testing [19]. Additional differences may potentially be influenced by the absence of active musculature in post-mortem specimens.

With six intervertebral levels contributing to the total HTTRR, Figure 8 and Table 3 compared to Figure 9 and Table 4, it was typically observed that individual segmental rotation peaks mirrored the timing of peak HTTRR. However, in some cases, specific segments displayed disproportionately higher motion, deviating from the otherwise distributed pattern. To investigate this further, intervertebral kinematics were examined at each cervical level and compared to observed injuries, potentially refining injury risk assessments beyond the global HTTRR metric. Results indicated that injured levels often coincided with segments displaying higher intervertebral extension peaks. However, certain injuries deviated from this pattern, exhibiting moderate extension peaks but pronounced motion during the initial flexion phase. This observation prompted a new metric: intervertebral peak-to-peak range (Table 6), was found to better correspond with the presence of injury (Table 8) as compared to intervertebral extension peaks. While flexion to extension peak-to-peak magnitudes appear to align well with sub-failure injury presence, it remains uncertain whether elevated peak-to-peak range is a causal contributor to injury by generating excessive intervertebral motion, or simply a kinematic expression of an injury that has already occurred. Further research will aim to validate these findings across a broader range of loading severities and more clearly defined injury outcomes.

Injury Criteria

To evaluate the potential injury severity and contextualize the loading environment, several established injury criteria were considered. Peak values for NIC were calculated using test data, with multiple PMHS exceeding the threshold of 15 m^2/s^2, as seen in Table 6 [9]. Additionally, the Nkm calculations, dominated by the Nea loading or extension and rearward (anterior) shear, support the likelihood of injury as Nkm values were above the proposed threshold of .33 in all but one of the tests [25]. These findings support the relevance of the test conditions for evaluating sub-failure and threshold-level cervical spine injury.

Limitations

This study is limited by the small PMHS sample size (n=6), though intervertebral segments were treated as independent observations (totaling 36), potentially introducing clustering bias. The lack of active muscle response and inability to assess pain perception are inherent limitations of

cadaveric testing, impacting the replication of real-world conditions. Injury assessment also relies on subjective visual indicators, especially in full-body testing where detailed cervical evaluation is only possible post-dissection. Moreover, pain can exist without visible damage. Future studies should use image registration techniques for more objective injury detection. Lastly, while a recreated crash pulse was used instead of a real-world collision, it enabled controlled data collection relevant to impact loading.

CONCLUSIONS

This study evaluates PMHS head and neck kinematics by examining gross and relative rotations and their propagation from the torso to the head during extension in a low-speed rear impact condition. The well-controlled experimental setup allowed full neck extension in the low-speed rear impact. The data suggests that intervertebral rotations could potentially be used to predict injury outcomes. Future research on intervertebral kinematics may refine neck injury criteria and enhance the biofidelity of anthropomorphic test devices and human body models for low-speed rear impacts.

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