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UNITED STATES ARMY AEROMEDICAL RESEARCH LABORATORY

Injury Assessment Reference Values for the Hybrid III 95th Percentile Male Pedestrian Anthropomorphic Test Device Lumbar Spine Under Vertical Loading

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14. ABSTRACT

The U.S. Army Aeromedical Research Laboratory (USAARL) developed injury assessment reference curves (IARCs) for the Hybrid III 95th percentile male pedestrian (HIII-95M-PED) anthropomorphic testing device (ATD) representing larger occupants. The HIII-95M-PED was used instead of the standard HIII-95M due to availability. The major differences between the two ATDs is that the pelvis of the pedestrian ATD has a sit-to-stand hip range of motion and a straight lumbar spine as opposed to the standard ATD that has a seated molded pelvis with limited hip range of motion and a curved lumbar spine. Male post-mortem human subject (PMHS) injury data were leveraged for this study from prior work by Lafferty et al. (2020). The HIII-95M-PED lumbar axial load cell performance IARV was determined for a 10% risk of thoracolumbar spinal injury (AIS 2+).

15. SUBJECT TERMS

lumbar spine, vertical accelerative testing, lumbar injury, vertical loading, injury assessment reference values, IARV, Male, 95th percentile male, Hybrid III, HIII, pedestrian pelvis, thoracolumbar spinal injury, Abbreviated Injury Scale (AIS)

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Executive Summary

U.S. Army aviators are at risk of exposure to high vertical accelerative loadings during helicopter mishaps that can result in severe or fatal injuries. Military specification MIL-S-58095A (AV), created in 1971 and updated in 1986, outlines the requirements for crash-resistant, non-ejection aircrew seats but only stipulates seat pan acceleration test requirements (Department of Defense [DoD], 1986). The specification does not specify injury assessment reference values to assess injury with the test surrogate (DoD, 1986). Furthermore, this specification was canceled in 1996 and rendered inactive without suitable replacement documents (DoD, 1996a). However, the military standard on aircraft crash resistance, MIL-STD-1290A (DoD, 1988), was canceled in 1995, reinstated in 2006, and validated for acquisition purposes in 2019 (DoD, 1995; DoD, 2006; DoD, 2019). It should be noted that MIL-STD-1290A cites MIL-S-58095A (AV) and MIL-S-85510 (AS) (both canceled) for seat performance requirements (DoD, 1981; DoD, 1996a; DoD, 1996b). Based on modern anthropometry (Gordon et al., 2014), there is a need within the military crashworthiness community to expand occupant protection requirements to address the safety of small female and large male aviators.

The U.S. Army Aeromedical Research Laboratory developed injury assessment reference curves (IARCs) for the Hybrid III 95th percentile male pedestrian (HIII-95M-PED) anthropomorphic test device (ATD) representing larger occupants. The HIII-95M-PED was used instead of the standard HIII-95M due to availability. The major differences between the two ATDs are the pelvis and the lumbar spine. The pelvis of the HIII-95M-PED has a sit-to-stand hip range of motion and the lumbar spine is straight, as opposed to the standard HIII-95M that has a molded seated pelvis with a limited hip range of motion and a curved lumbar spine.

Male post-mortem human subject (PMHS) injury data were leveraged for this study from prior work by Lafferty et al. (2020). Lafferty et al. (2020) determined that direct force measurements in an instrumented ATD were a better indicator of injury risk for future military rotary-wing aircraft crashworthy seat development efforts than seat-based acceleration measurements.

This study developed an IARC for the HIII-95M-PED by leveraging 50th percentile male PMHS IRCs from prior research (Lafferty et al., 2020). Matched-pair testing was conducted using the previous PMHS test data and the HIII-95M-PED data to develop an IARC for the HIII-95M-PED ATD. The axial compressive force threshold of 1167 pounds was recommended as the IARV for the lumbar load cell of the HIII-95M-PED to control for a 10% risk of an AIS 2+ injury during dynamic vertical loading.

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Acknowledgments

The Injury Biomechanics and Protection Group would like to thank Nathan Flath, Joseph Parish, Joseph Willett, and Shawn Booms for their assistance with test setup, execution, and data collection.

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Introduction

U.S. Army aviators are at risk of exposure to high vertical accelerative loadings during helicopter crashes. These crashes can result in severe or fatal injuries. Energy attenuating seats were developed and fielded in modern U.S. Army rotary-wing aircraft to provide increased crash protection to the U.S. Army aviator. However, the performance requirements of military seat specification MIL-S-58095A only stipulate seat pan acceleration and not biomechanical response measurements in the test surrogate (Department of Defense [DoD], 1986). Furthermore, this specification was canceled in 1996 and rendered inactive without suitable replacement documents (DoD, 1996a). The military standard on aircraft crash resistance, MIL-STD-1290A (DoD, 1988), was canceled in 1995, reinstated in 2006, and validated for acquisition purposes in 2019 (DoD, 1995; DoD, 2006; DoD, 2019). It should be noted that MIL-STD-1290A cites MIL-S-58095A and MIL-S-85510 (both canceled) for seat performance requirements (DoD, 1981; DoD, 1996a; DoD, 1996b). Biomechanical response measurements (injury assessment reference values [IARVs]) provide a means for the materiel developers and acquisition program personnel to assess seat energy attenuation performance and the resulting injury risk. Relevant biomechanical response measures must be incorporated into the seats' acquisition standard to provide a more appropriate and realistic standard to guide future seat designers and contractual acceptance performance requirements.

In 2020, the U.S. Army Aeromedical Research Laboratory (USAARL) assessed thoracolumbar spinal injury risk during exposure to the 23 G (where G is the standard acceleration due to earth's gravity) acceleration requirement stipulated in MIL-S-58095A (Lafferty et al., 2020). Injury risk curves (IRCs) were developed from male post-mortem human subject (PMHS) data and matched-pair testing was conducted with the standard automotive Hybrid III 50th percentile male (HIII-50M) and the Federal Aviation Administration (FAA) Hybrid III 50th percentile male (FAA-HIII 50M) anthropomorphic test devices (ATDs). The FAA-HIII 50M, used in FAA aircraft seat certification tests, is nearly identical to the standard automotive HIII-50M, except the curved lumbar spine of an HIII-50M was replaced with a straight lumbar spine in the FAA-HIII 50M (Figure 1.). Injury assessment reference curves (IARCs) were developed during the 2020 study for the lumbar axial compressive forces measured in the HIII-50M and the FAA-HIII ATDs. Then, IARVs were determined from the IARCs at discrete levels of injury risk relating the axial compressive lumbar load of the standard HIII-50M and FAA-HIII to moderate (Abbreviated Injury Scale [AIS] 2+) (Gennarelli et al., 2006) and serious (AIS 3+) PMHS injuries. The axial compressive force thresholds of 1135 and 1223 pounds (lb) were recommended as the IARVs for the lumbar load cell of the HIII-50M and FAA-HIII 50M, respectively, to control for a 10% risk of an AIS 2+ injury during dynamic vertical loading.

In 2021, Lafferty et al. continued the IARV development from the male dataset to include the standard Hybrid III 5th percentile female (HIII-5F) ATD exposed to the dynamic vertical loading to control for a 10% risk of AIS 2+ injury. The lumbar load cell axial compressive performance threshold of 909 lb was recommended for the small female ATD. These recommendations were helpful and critical for improving mid-sized male and small female occupant protection during a survivable vertical impact; however, the work did not expand to create IARVs for large occupants (Lafferty et al., 2021).

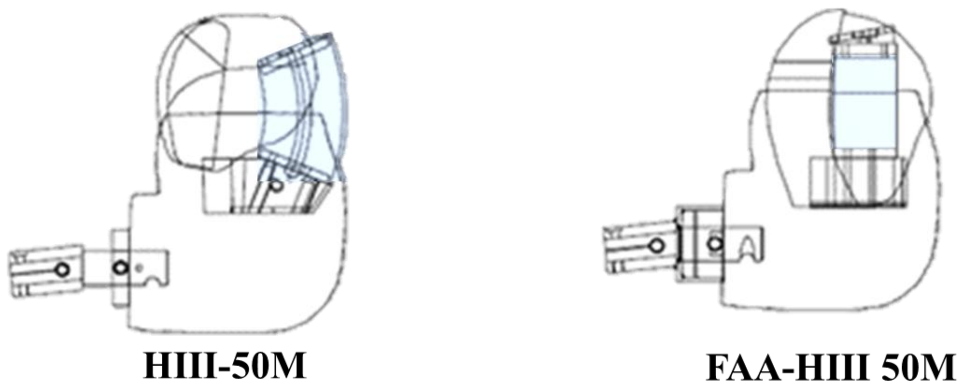

*Figure 1.*The difference in the lumbar spine curvature in the HIII-50M (left) and FAA-HIII 50M (right) ATDs is indicated by the blue shaded region.

For the present study, the Hybrid III 95th percentile male pedestrian (HIII-95M-PED) ATD (The Humanetics Group, Farmington Hills, MI) was used to develop an IARV for the 95th percentile male Soldier due to immediate availability. The HIII-95M-PED has a straight lumbar spine and a sit-to-stand pelvis developed to test interactions between a pedestrian and a vehicle to allow for a more erect seated posture, while the curved lumbar region and molded seated pelvis in the standard HIII-50M and HIII 95th percentile male (95M) maintains these ATDs in a slouched seated posture (Figures 2A and B; Figures 3A and B). Zhang et al. (2013) and Moffatt et al. (2003) used the HIII Pedestrian to study whole-body kinematics in vehicle crashes that included a rollover. The increased hip mobility of the Pedestrian pelvis is thought to offer a more biofidelic response, but studies to evaluate the differences between the Pedestrian and standard pelvis are limited (Zhang et al., 2013; Moffatt et al., 2003).

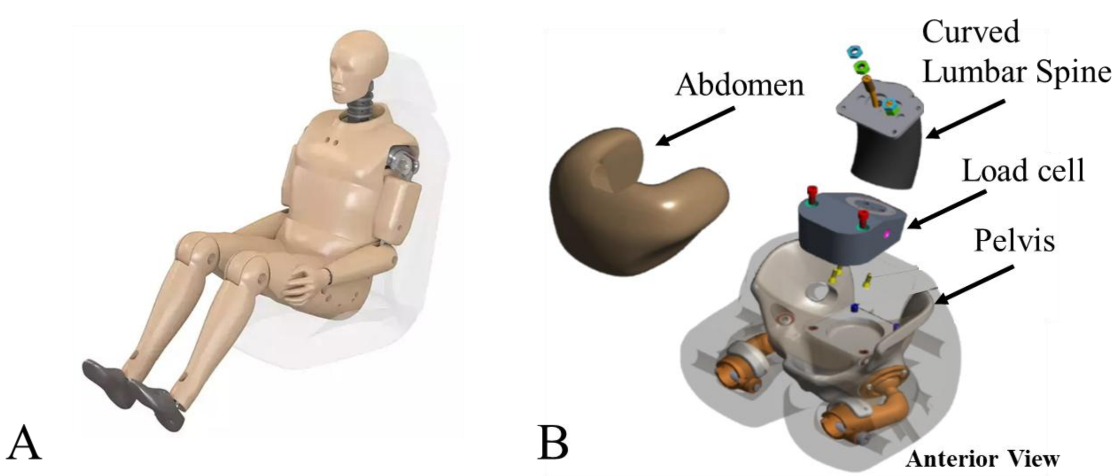

Figure 2. (A) The HIII-95M ATD is shown with the standard molded seated pelvis (reproduced from Humanetics Group, 2024a). (B) The anterior oblique view of the standard HIII-95M ATD lumbar spine and pelvis assembly (reproduced from Humanetics Innovative Solutions, 2017a) is also shown.

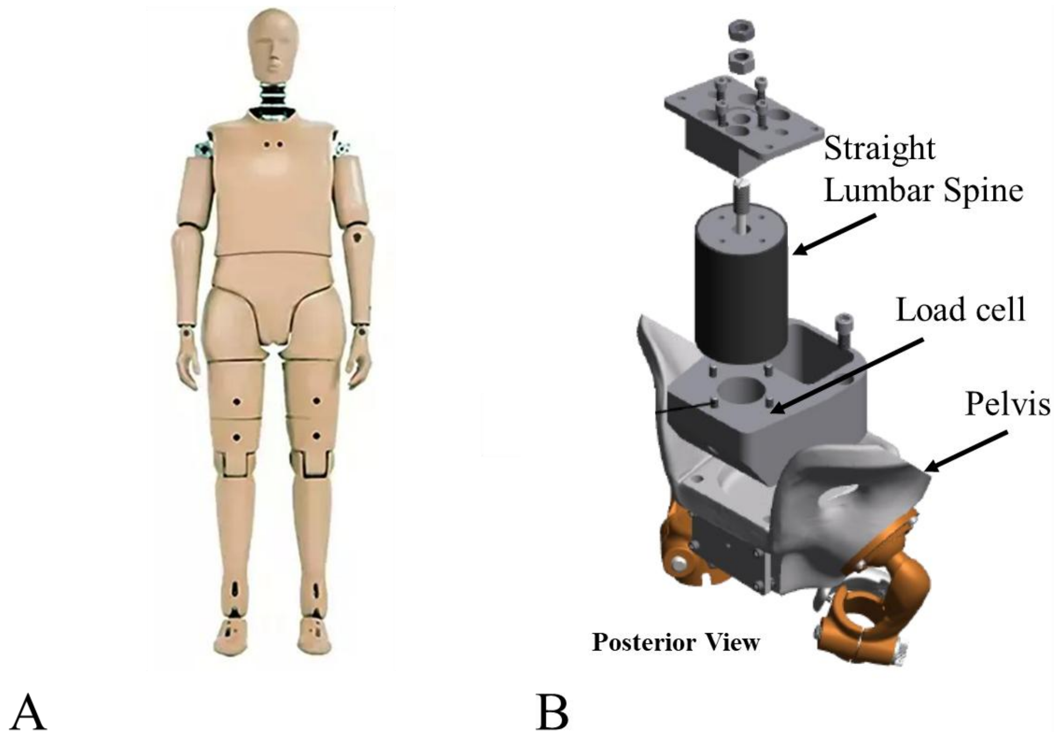

Figure 3. (A) The HIII-95M-PED ATD with the sit-to-stand pelvis (reproduced from Humanetics Group, 2024b) is shown. (B) The posterior oblique view of the HIII-95M Pedestrian lumbar spine and pelvis assembly (reproduced from Humanetics Innovative Solutions, 2017b) is also shown.

There is a need within the military crashworthiness community to expand occupant protection requirements to address the safety of the large male aviator. Research was needed to provide large male occupant protection recommendations to seat designers and program managers. Development of IARVs for the large male population through studies using a large male ATD like the HIII-95M-PED will provide quantitative values as a starting point to improve large male occupant protection during high vertical accelerative loadings.

This study aimed to develop an IARC for the HIII-95M-PED by leveraging 50th percentile male PMHS IRCs from prior research (Lafferty et al., 2020). Matched-pair testing was conducted using the HIII-95M-PED under conditions matching each of the previous PMHS tests to develop an IARC for the HIII-95M-PED ATD. The IARVs at selected injury risk levels were determined from the IARC.

Methods

The USAARL vertical acceleration tower (VAT) was used to apply controlled vertical accelerations to a carriage containing an HIII-95M-PED ATD seated on a rigid 90-90-90 seat (Figure 6). The VAT is equipped with a tower that is 40 feet (ft) tall and a pneumatic HyGe^{TM} actuator (HYGE INC., Kittanning, PA) capable of generating vertical carriage accelerations over 75 G (Figure 4). System input variables (e.g., pressure settings, volumes, metering pin shape, and carriage mass) are used by operators to control the vertical acceleration pulse.

Figure 4. The USAARL HYGE™ (Kittanning, PA) VAT was used to produce the vertical crash pulse for this study.

The rigid seat was attached to the carriage with a horizontal seat pan, vertical seat back, and horizontal foot pan. This geometry reflected a simplified, experimentally controlled scenario for aviator thoracolumbar compression injuries. The ATD was positioned in the seat with the mid-sagittal plane aligned with the center of the seat (Figure 5). During the positioning process, a coordinate measuring machine, FARO Arm Platinum (FARO, Lake Mary, FL), was used to check and document the ATD's posture. The ATD's posture was set to match the back angle of one of the PMHS postures used in the previous study by Lafferty et al. (2020). The back angle was measured as the angle between the vertical seat back and the hip-to-shoulder segment (Figure 6). The ATD was positioned where the back angle was within 2 degrees of the corresponding PMHS test. A rotary buckle five-point restraint system without inertial reels was used to restrain the ATD. The straps were adjusted pre-test to provide 10 to 20 lb of tension in each strap. The arms and legs of the ATD were tethered at the wrists and ankles, respectively, with slack to allow unimpeded reaction during loading, while reducing reactionary extremity flail during carriage deceleration. After each test, the ATD was inspected for degradation or wear in the pelvis due to repeated use.

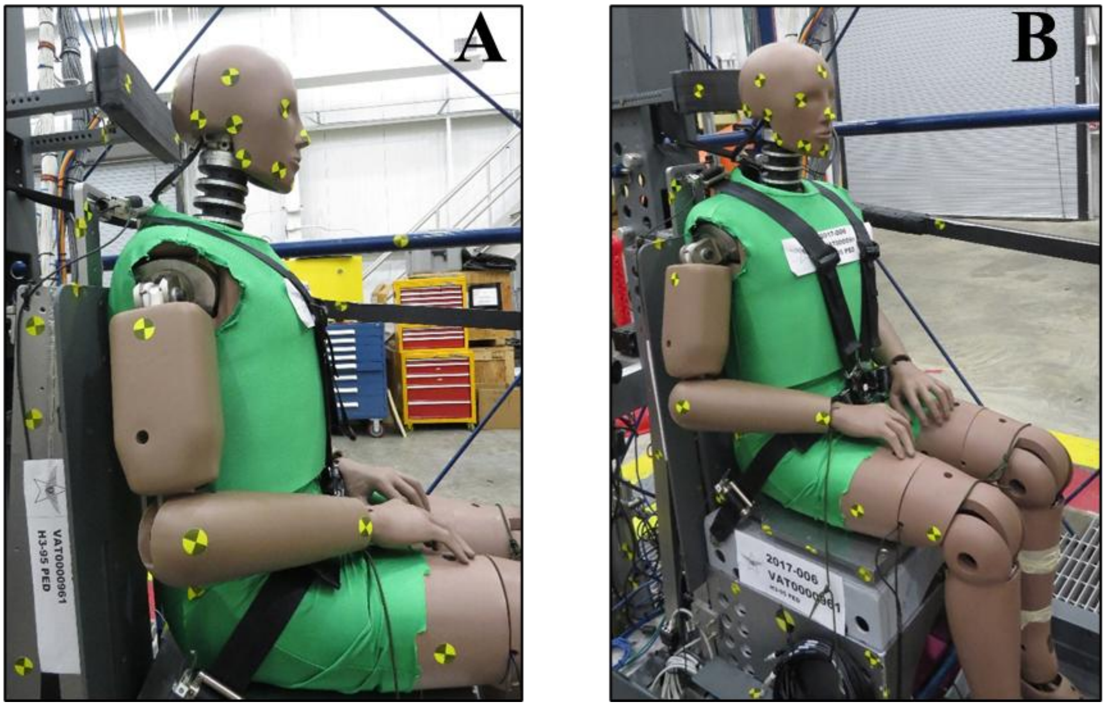

Figure 5. The HIII-95M-PED was positioned on the VAT in the rigid seat; (A) shows the side view and (B) shows the isometric view.

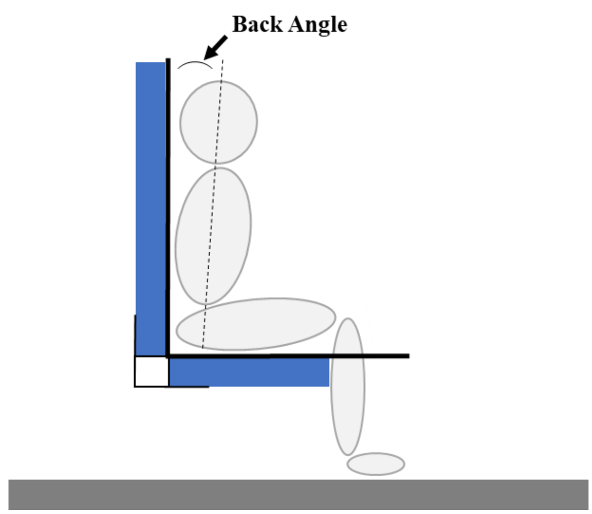

Figure 6. The back angle measurement to control ATD positioning in the 90-90-90 rigid seat is shown.

The VAT acceleration exposures were designed to match the previous PMHS tests. Tests were grouped into four different exposures (Exposure 1, 1b, 2, and 3) (Table 1) (Lafferty et al., 2020). The matches were based on duplication of the following conditions: back angle, peak acceleration, total change in velocity (ΔV) in feet per second (ft/s), and acceleration onset rate in G per second (G/s). The HIII-95M-PED was instrumented with accelerometers in the head, chest, and pelvis; angular rate sensors in the head; and load cells in the pelvis and upper and lower neck. Sensor data were collected at 200,000 samples per second. Each ATD test was documented using multiple high-speed video cameras that collected data at 2000 images per second. The Society of Automotive Engineers (SAE) Recommended Practice J211-1 Part 1 was applied to guide data acquisition and reduction (Society of Automotive Engineers, 2007). Data were time-aligned based on initial carriage motion as captured by the carriage accelerometer, where time-zero was calculated to be when acceleration, filtered at channel frequency class (CFC) 60, first reached 5% of its peak. Metrics calculated from the carriage's acceleration included the carriage's velocity and acceleration onset rate. The onset rate was calculated as the maximum onset slope over a 5 millisecond (ms) duration before obtaining its peak acceleration. The carriage velocity was calculated as the numerical integration of the acceleration-time data.

Table 1. Mean Vertical Exposure Parameters of the Carriage from PMHS Tests Identified From Lafferty et al. (2020) for Replication With the HIII-95M-PED

Exposure	Peak Acceleration (G)	ΔV (feet/second [fps])	Onset Rate (G/s)
1	21.3	41.8	1408
1b	16.5	35.7	1154
2	21.5	41.6	1116
3	16.2	42.0	1003

Thirty HIII-95M-PED vertical acceleration tests were performed to identify a matched-pair for each of the 14 PMHS tests previously conducted (Lafferty et al., 2020). The matching criteria used to identify one unique ATD test for each PMHS test included peak carriage accelerations matched to within ±2%, ΔVs to within ±2%, onset rates within ±10%, and back angle to within ±2° of the PMHS by Lafferty et al. (2020). One matched-pair test used the carriage thrust load cell (LC) after the failure of the carriage accelerometers (Lafferty et al., 2020). Additional exclusion criteria considered when determining the ATD matched-pairs were those ATDs with a "knees up" non-horizontal thigh positioning or "arms up," where they were not initially in contact with the lap.

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The IARC was developed using parametric survival analysis techniques pairing the peak axial compressive HIII-95M-PED lumbar load data with the male PMHS AIS 2+ thoracolumbar injuries from Lafferty et al. (2020) (Appendix). The best distribution for the IARC was selected using the Akaike information criterion (AIC). The AIC estimated the quality of the survival analysis model given three underlying distributions: log-normal, log-logistic, and Weibull, where the lowest AIC score indicated the best distribution. Data were considered uncensored observations. It should be noted that both bone mineral density and age were evaluated as covariates in the development of the IARCs developed by Lafferty et al. (2020), but neither were found to be significant during survival analysis ($p < 0.05$) (Appendix). The quality of the IARC was checked with the normalized confidence injury score (NCIS) and calculated as the ratio of the IARC's confidence interval (CI) to the mean value at discrete probabilities. The NCIS values less than 0.5 were considered "good" (Petitjean et al., 2015). The IARV was selected for a 10% risk of injury from the IARC.

Results

For each of the 14 PMHS tests, a singular HIII-95M-PED ATD test was selected as a matched-pair (Table 2) and removed from the candidate pool. Time traces for the matched test carriage accelerations of the HIII-95M-PED and PMHS for each exposure are shown in Figure 7 through 10. All but one of the ATD tests were matched using the selection criterion. The onset rate of the carriage for the matched-pair ATD Test 980 associated with PMHS Test 9 was 16% lower and outside of the ±10% match criterion (Table 2); however, since all other metrics used to match the ATD Test 980 fell within the desired range, this test was used for analysis.

Another exception occurred when the carriage accelerometers failed during PMHS Test 10. The carriage thrust LC was examined for ATD Test 981 to match the pair as per the procedure conducted by Lafferty et al. (2020). The carriage thrust LC of PMHS 10 and ATD Test 981 were within 1% of each other (Figure 11).

After each test, the ATD pelvis was inspected for wear and damage. No visual degradation or wear to the ATD pelvis was noted upon post-test inspections.

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Table 2. Exposure Parameters for PMHS and Corresponding HIII-95M-PED Test

Exposure	Surrogate	ID	Peak Carriage Acceleration		Carriage ΔV		Peak Carriage Onset Rate		Back Angle	
			G	Percent Difference	fps	% Diff	G/s	% Diff	deg	Difference (deg)
1	PMHS	1	21.4	-1.4%	41.3	1.0%	2009.9	-2.7%	-2.9	0.5
	ATD	967	21.7		40.9		2064.6		-3.4	
	PMHS	2	22	1.4%	42	-0.2%	2146.4	-4.7%	0.6	-0.2
	ATD	979	21.7		42.1		2247.7		0.8	
	PMHS	3	21.8	1.8%	42	-0.5%	2022.6	-3.7%	-2	0.1
	ATD	959	21.4		42.2		2097.2		-2.1	
	PMHS	4	21	-1.4%	40.9	-0.7%	1986.9	-4.4%	-3.6	-1.1
	ATD	968	21.3		41.2		2073.4		-2.5	
	PMHS	5	21.8	0.0%	42.2	1.2%	1997.7	-7.6%	-1.7	-0.7
	ATD	962	21.8		41.7		2150.5		-1	
	PMHS	6	21.8	1.4%	42	-0.2%	1893.6	-8.3%	-0.9	1.3
	ATD	960	21.5		42.1		2050.1		-2.2	
	PMHS	8	21.8	1.8%	42	-1.0%	1958.1	-8.6%	-3.3	-0.6
	ATD	961	21.4		42.4		2127.1		-2.7	
1b	PMHS	7	16.8	0.0%	35.5	-1.4%	1609.4	1.5%	-3.1	-0.3
	ATD	977	16.8		36		1584.8		-2.8	
2	PMHS	9	21.8	1.8%	41.5	-1.0%	1422.2	16.1%	-3.5	0.3
	ATD+	980	21.4		41.9		1192.6		-3.8	
	PMHS	10	*	NA	*	NA	*	NA	-3.3	-0.1
	ATD	981	21.5		42.5		1254.2		-3.2	
3	PMHS	11	16.7	1.8%	42	0.7%	1258.8	6.7%	-2.7	1.4
	ATD	974	16.4		41.7		1174.1		-4.1	
	PMHS	12	16.5	-1.2%	42.1	-0.2%	1250.4	4.3%	0.7	0.6
	ATD	973	16.7		42.2		1196.1		0.1	
	PMHS	13	16.5	0.0%	42.3	-0.2%	1243.6	2.9%	-0.8	-0.4
	ATD	970	16.5		42.4		1207		-0.4	
	PMHS	14	16.5	1.2%	42	-0.2%	1191.1	3.1%	-0.1	0.5
	ATD	969	16.3		42.1		1153.6		-0.6	

Note. PMHS and ATD tests were grouped by matched-pairs denoted by the shaded and unshaded regions.

+Peak carriage onset rate for this test was 16% lower and outside the ±10% match criterion.

*One matched-pair test used the carriage thrust LC after the failure of the carriage accelerometers. The match was made using the carriage thrust. See Figure 9 and 11.

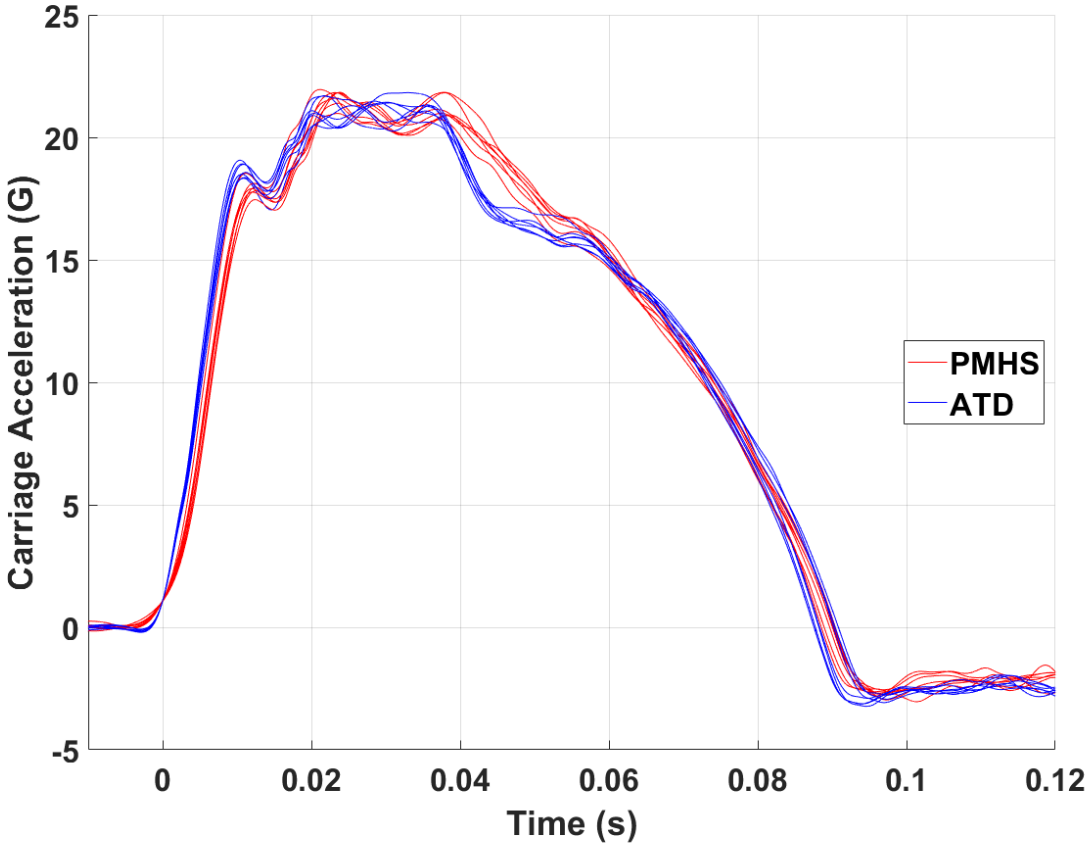

Figure 7. The carriage accelerations of Exposure 1 are shown for the PMHS and ATD, where seven tests are shown for each occupant.

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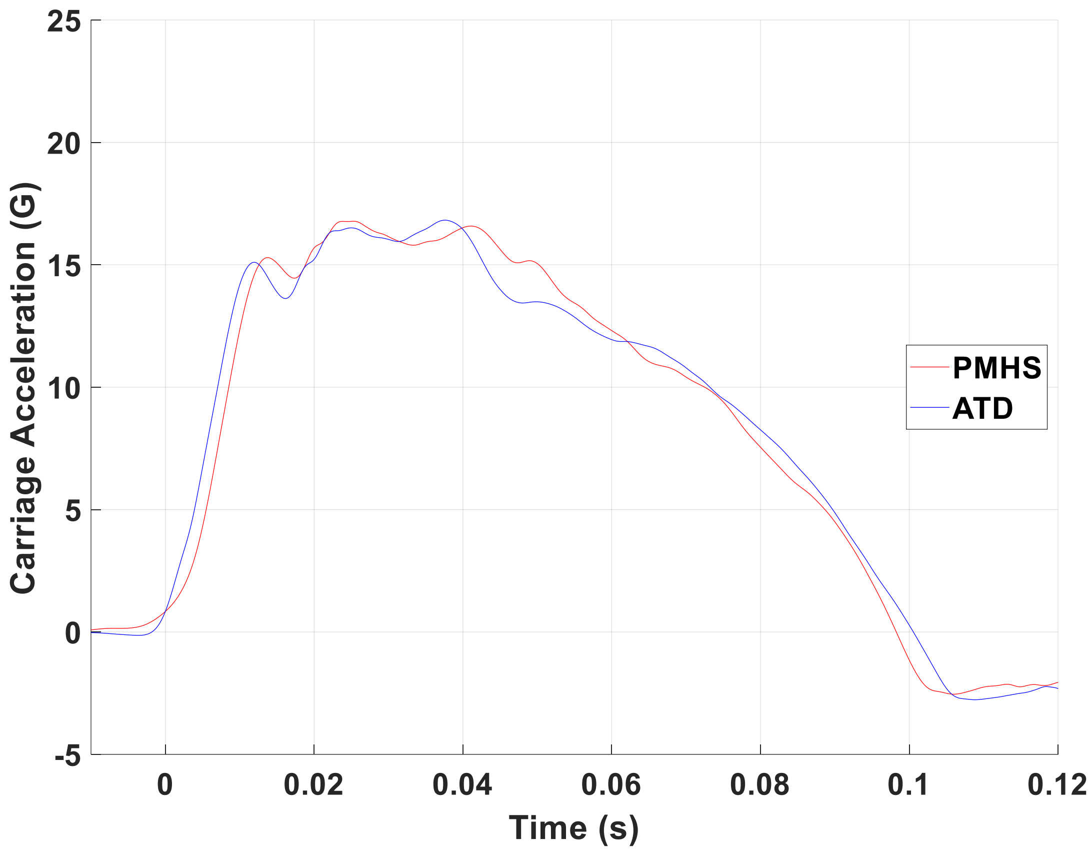

Figure 8. The carriage accelerations of Exposure 1b are shown for the PMHS and ATD where one test is shown for each occupant.

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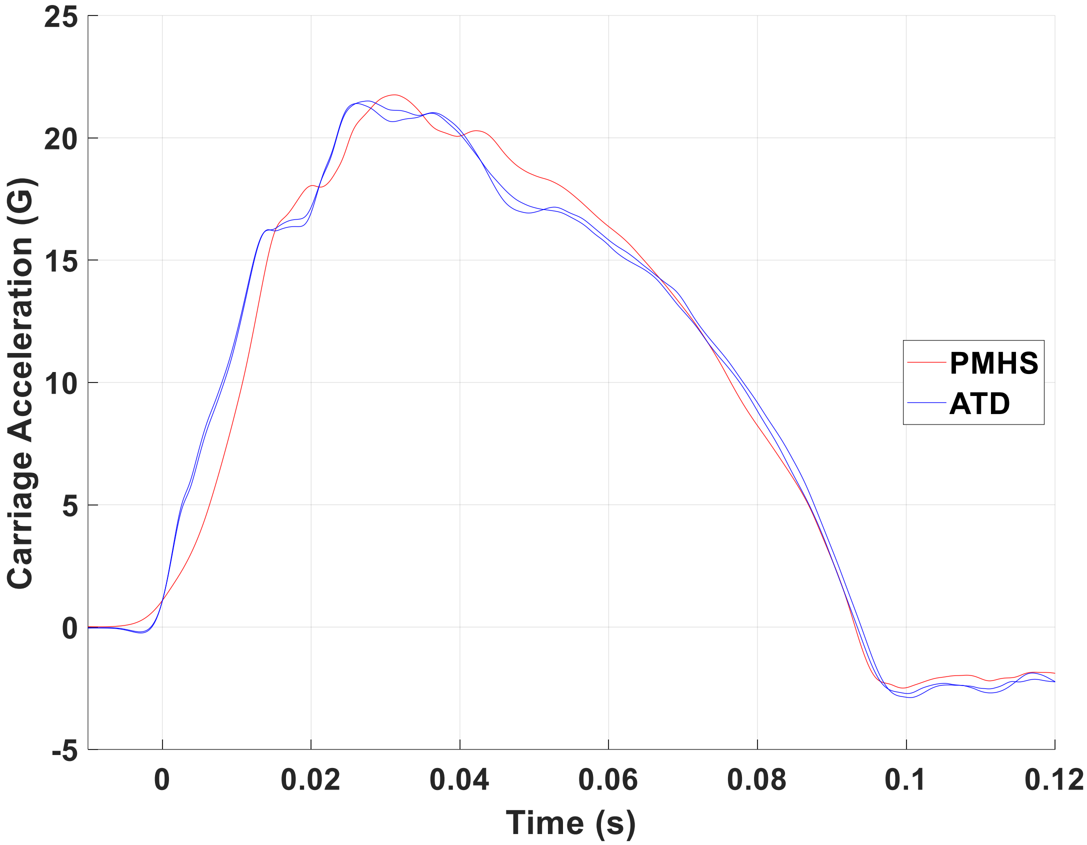

Figure 9. The carriage accelerations of Exposure 2 are shown for the PMHS and ATD where two tests are shown for each occupant (see note on Table 2).
Note. Instrumentation failure on one of the PMHS tests.

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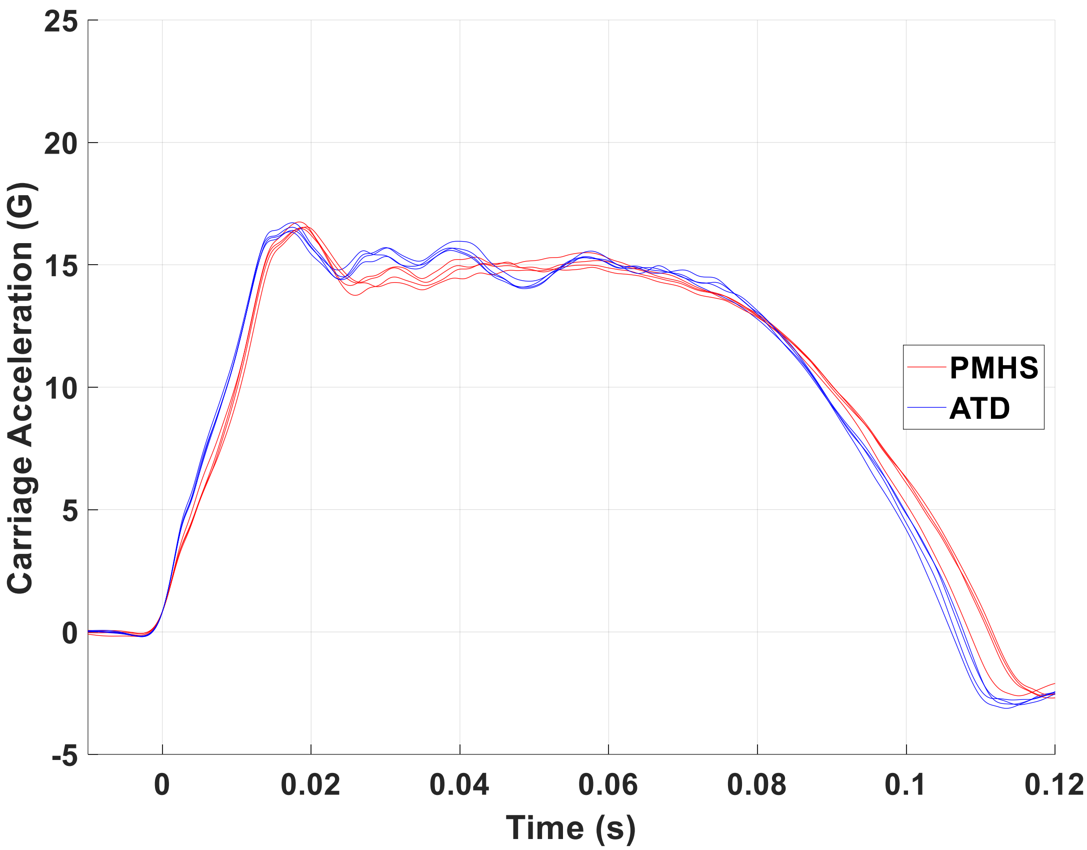

Figure 10. The carriage accelerations of Exposure 3 are shown for the PMHS and ATD where four tests are shown for each occupant.

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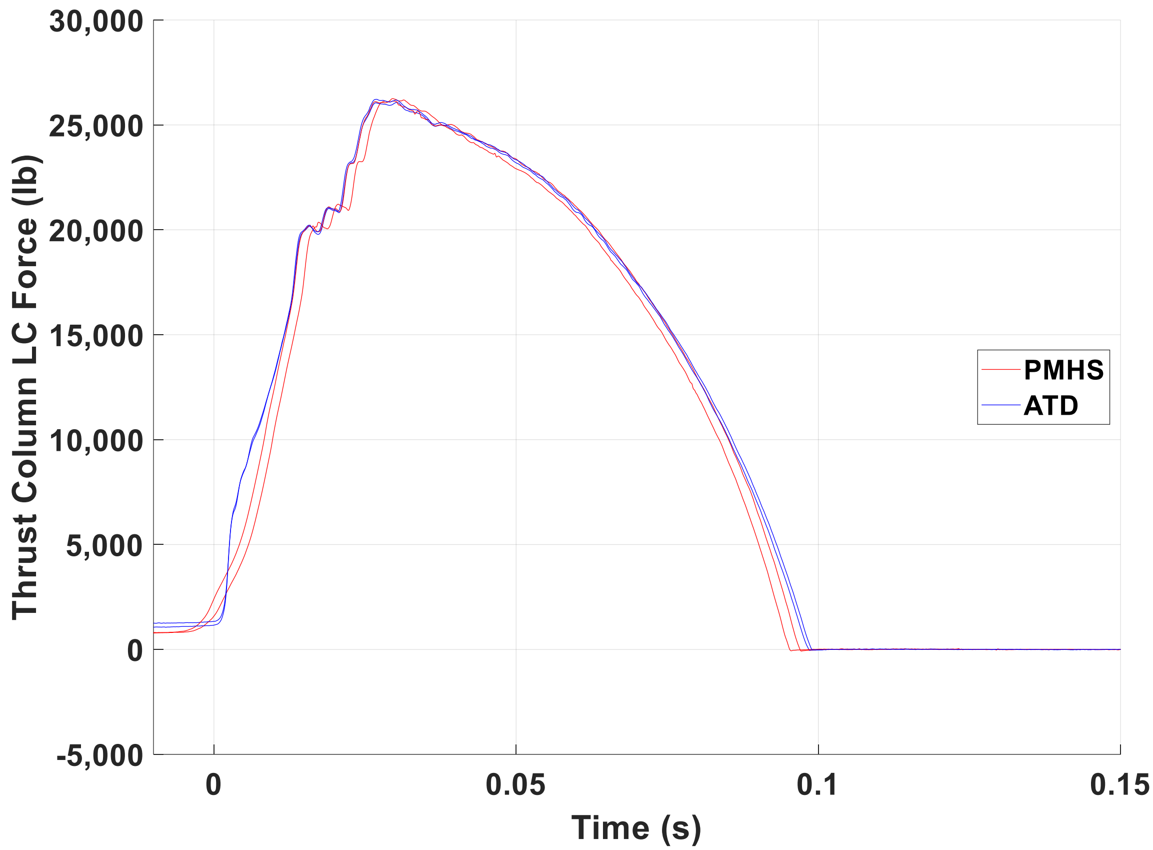

Figure 11. The thrust column input force to the carriage for Exposure 2 for the PMHS and ATD was used to verify matching conditions where carriage acceleration data was absent for one PMHS test.
Note. The approximate static weight of the carriage resting on the thrust column can be seen before Time = 0.

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The lumbar axial data collected for each matched-pair test were grouped by exposure (Figure 12 through 15). Overall, peak lumbar axial force (the minimum value observed) ranged between -1068 and -1438 lb (Table 3) with an average of -1270 lb (±109 lb). Peak loads occurred between 0.02 and 0.06 seconds (s) of the initiation of carriage motion for all exposures. A total of 10 out of the 14 runs resulted in a double peak in lumbar axial compressive force during the exposure. The overall lowest observed force was recorded regardless of whether the signal was unimodal or multi-modal.

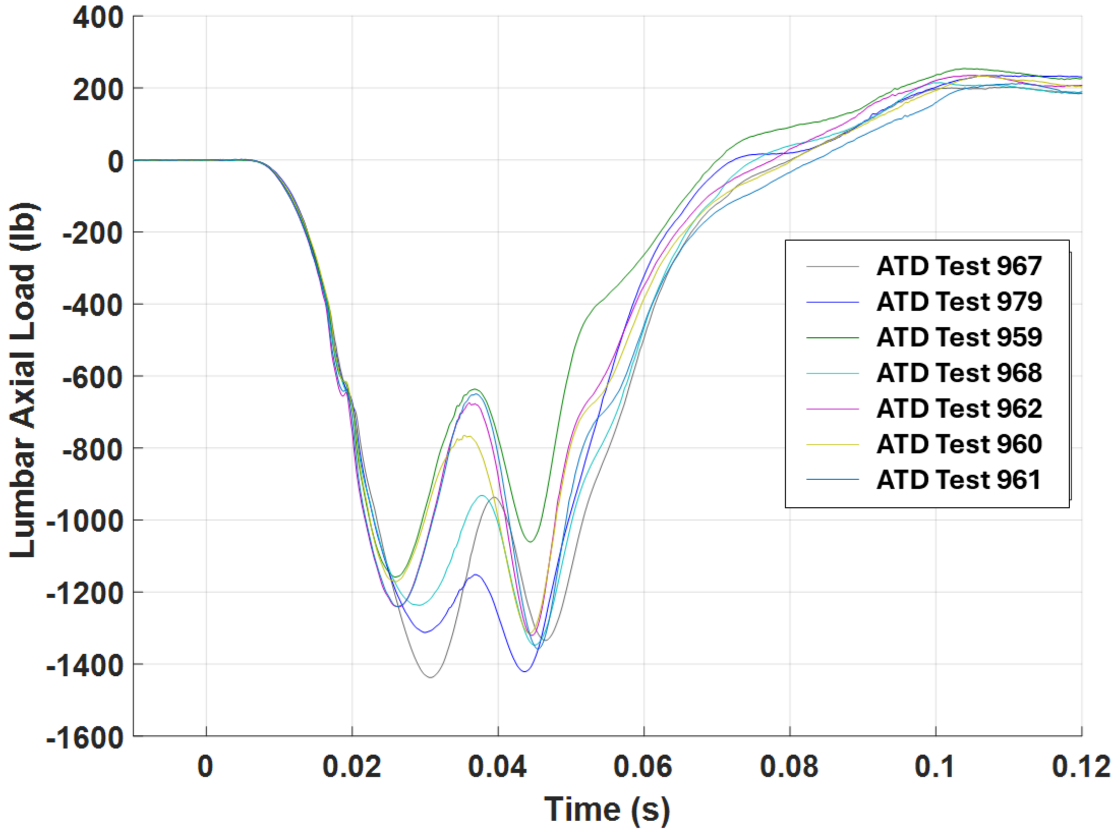

Figure 12. The lumbar axial force versus time data were recorded for each matched-pair run at Exposure 1 for the HIII-95M-PED. Positive axial values represent lumbar tension loads and negative values represent lumbar compression loads.

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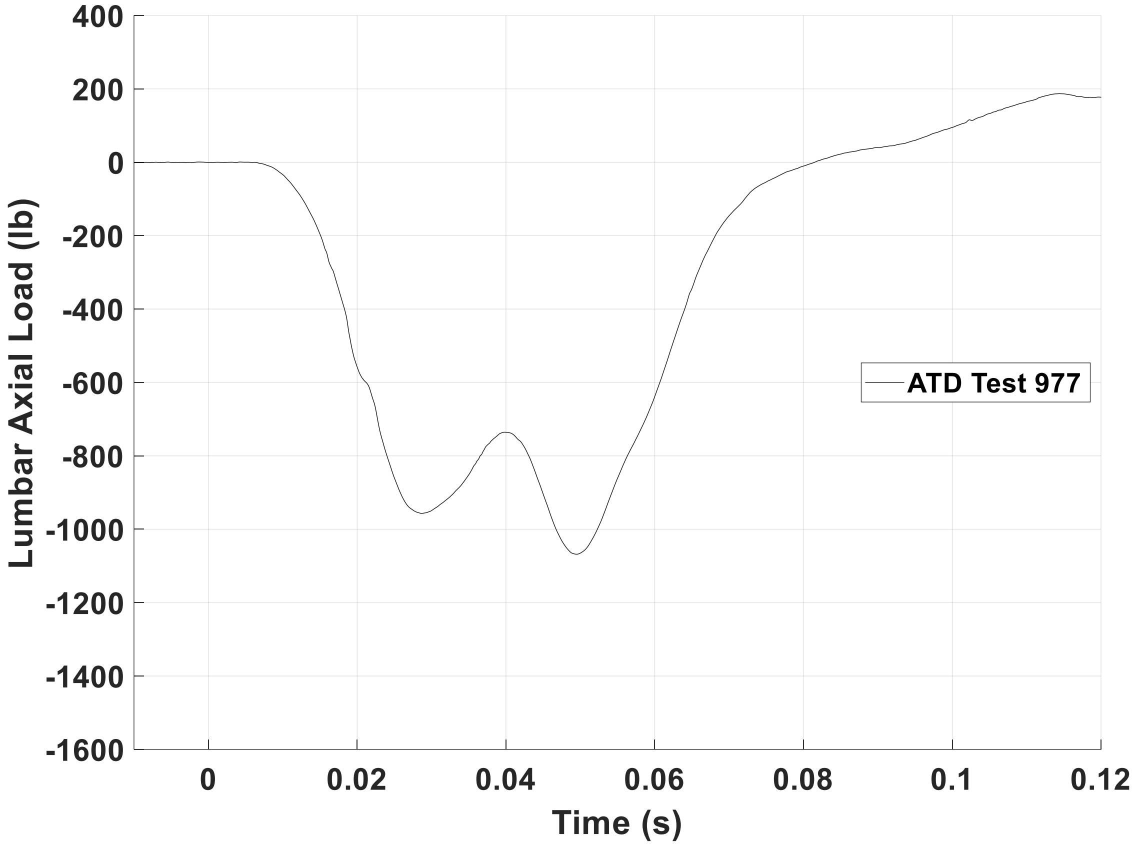

Figure 13. The lumbar axial force versus time data were recorded for the matched-pair run at Exposure 1b for the HIII-95M-PED. Positive axial values represent lumbar tension loads and negative values represent lumbar compression loads.

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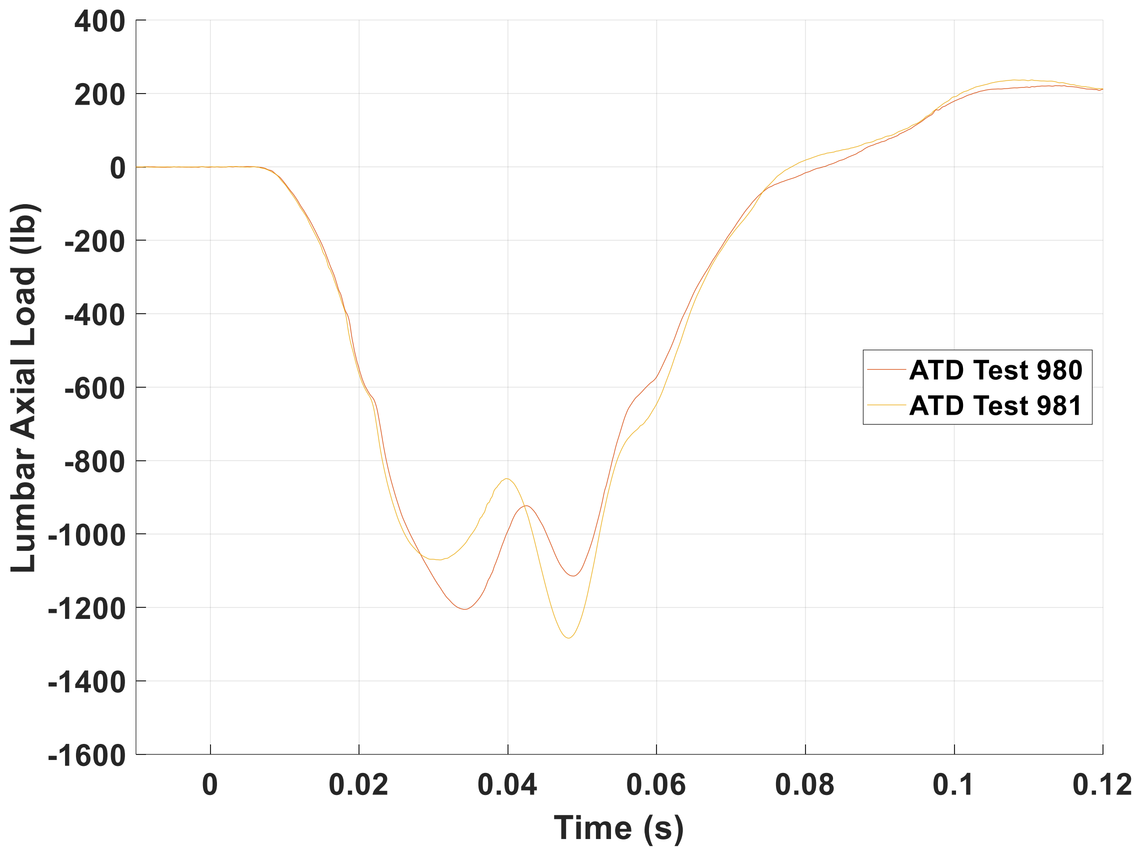

Figure 14. The lumbar axial force versus time data were recorded for each matched-pair run at Exposure 2 for the HIII-95M-PED. Positive axial values represent lumbar tension loads and negative values represent lumbar compression loads.

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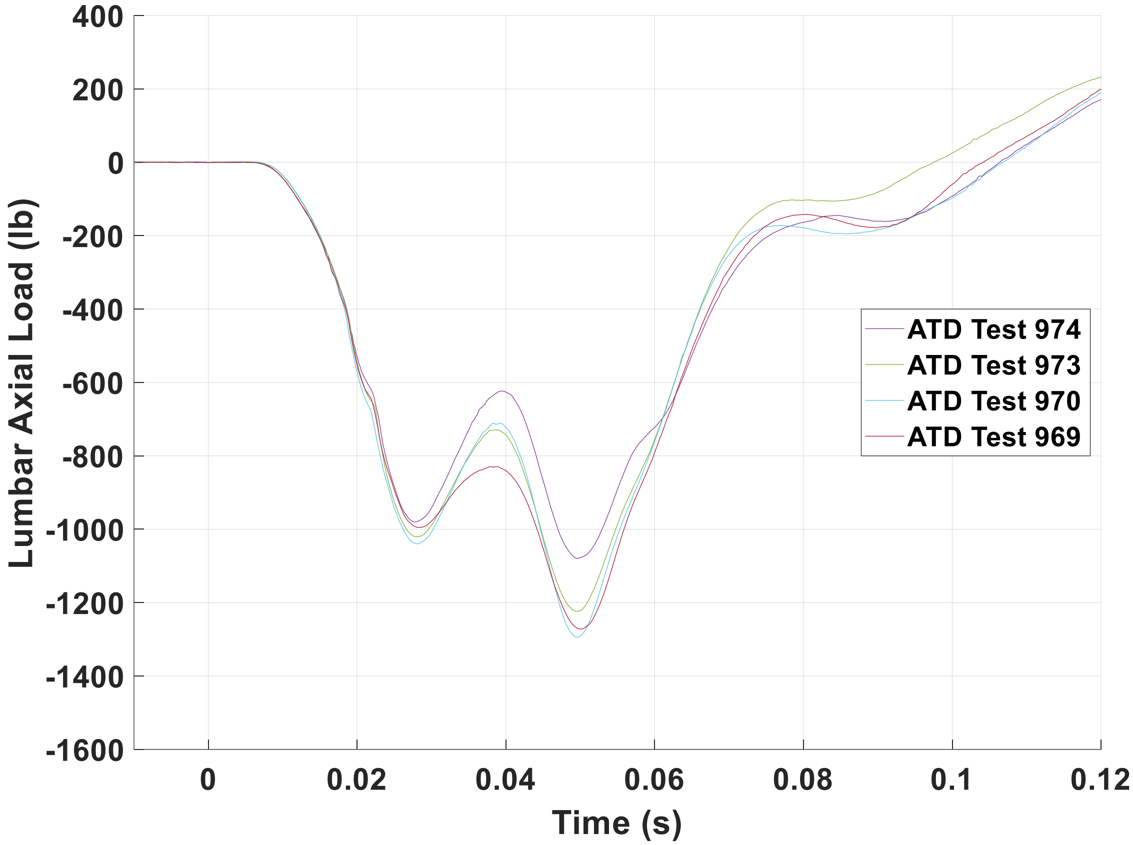

Figure 15. The lumbar axial force versus time data were recorded for each matched-pair run at Exposure 3 for the HIII-95M-PED. Positive axial values represent lumbar tension loads and negative values represent lumbar compression loads.

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Table 3. Peak Axial Compressive Lumbar Loads for the HIII-95M-PED Used for Survival Analysis

Prior Male PMHS Testing			**Matched HIII-95M-PED**
PMHS ID	Exposure	Highest AIS*	Lumbar Axial Load (lb)
1	1	2	-1438
2	1	2	-1421
3	1	3	-1158
4	1	2	-1347
5	1	3	-1320
6	1	1	-1315
7	1b	3	-1068
8	1	3	-1358
9	2	3	-1205
10	2	3	-1284
11	3	1	-1079
12	3	1	-1224
13	3	2	-1294
14	3	2	-1272

*Column denotes highest AIS in the thoracolumbar and/or pelvic regions (Lafferty et al., 2020)
Note. Negative values represent a compressive lumbar load.

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The data were treated as uncensored for the development of the moderate or greater (AIS 2+) thoracolumbar IARCs and the three underlying distributions (i.e., log-normal, log-logistic, and Weibull) presented AICs of 140.7, 140.5, and 138.8, respectively. The lowest AIC, the Weibull distribution, was chosen by the USAARL researchers as the best for the IARC (Figure 16). Age and bone mineral density were not previously found to be significant during the male PMHS survival analysis conducted by Lafferty et al. (2020) and were therefore not evaluated as covariates. The IARVs were identified at 5, 10, 20, 25, 50, 75, 90, and 95% injury risk on the AIS 2+ IARC. The NCISs were calculated at those discrete values and were all found to be below 0.5 and considered "good." A 10% risk of moderate (or greater) thoracolumbar spinal injury (AIS 2+) was selected by the team to define the IARV for injury risk assessment for the 95th percentile Soldier (Table 4).

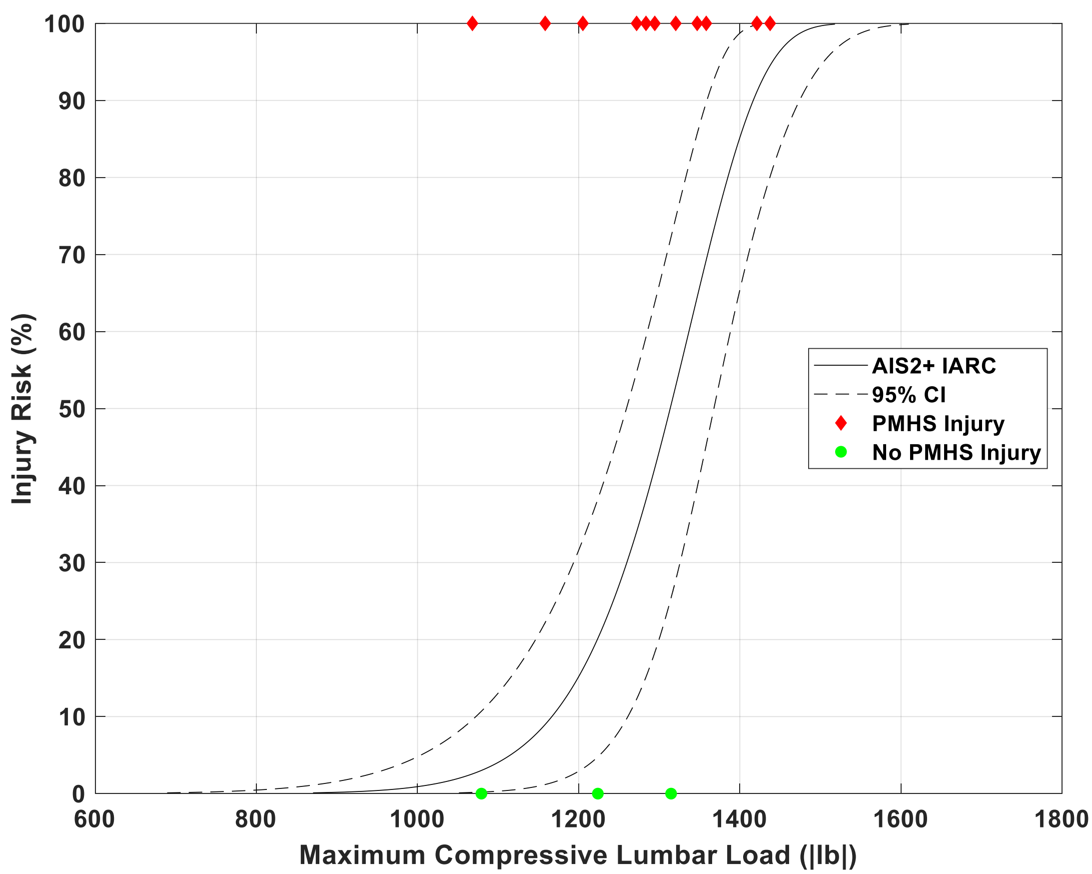

Figure 16. The IARC and associated 95% CI for the maximum vertical compressive lumbar loads of the HIII-95M-PED for AIS 2+ thoracolumbar injury risk is shown. The ATD peak compressive lumbar load values are plotted (◊/O) against PMHS AIS 2+ outcome (0% for no AIS 2+ injury and 100% for AIS 2+ injury). Lumbar axial compressive load is negative by SAE convention; however, they are plotted here as absolute values for survival analysis functionality.

Table 4. Compressive Lumbar Load AIS 2+ IARVs for the HIII-95M-PED

AIS 2+ Injury Risk (%)	Compressive Load (lb)	NCIS
5	1115	0.20
10	**1167**	**0.16**
20	1223	0.13
25	1243	0.11
50	1314	0.08
75	1372	0.07
90	1417	0.08
95	1440	0.09

Note. A 10% risk of AIS 2+ thoracolumbar injury was selected as a recommended IARV (indicated by the bolded and boxed row). Lumbar axial compressive load is negative by SAE convention; however, they are listed here as absolute values for survival analysis functionality.

Discussion

There is a need within the military crashworthiness community to provide large male occupant protection recommendations to seat designers and program managers in rotary-wing mishaps. However, military seat specifications do not require biomechanical response measurements in the test surrogate (DoD, 1986). Military specifications should be revised to include ATD performance requirements so that injury risk for all Soldier populations can be properly assessed.

The HIII-95M was selected to represent large aircraft occupants; however, the pedestrian pelvis was used instead of the standard seated molded pelvis due to its immediate availability. It should be noted that there was no visual degradation or wear to the pelvis upon post-test inspections. The study team noted in a previous test series that the standard seated molded pelvis of the adult 50M HIII ATDs create a pinch point between the ATD's aluminum ischial tuberosities and the rigid seat pan that result in observable rupture of the ATD's skin material, which is termed "punch through" by the ATD industry. No "punch through" issues were noted throughout testing using the HIII-95M-PED pelvis.

Due to the ATD configuration used, these results are unique to this pelvis size and configuration; therefore, the resulting IARC and IARV should not be applied to other HIII ATDs or other pelvis configurations unless further testing justifies such application. One such difference in ATD pelvic response was that nearly all lumbar axial compressive LC signals were bimodal, in that they had two distinct peaks (Figure 12 through Figure 15). The maximum lumbar axial compressive force was used for the data analysis, regardless of whether it was the first or second identifiable peak, to match the methodology used during the PMHS testing by Lafferty et al. (2021). The bimodal behavior was unique to the current test series, which used a pedestrian style pelvis, and was not observed during the HIII-50M ATD testing previously conducted with a seated molded pelvis (Lafferty et al., 2020).

It was noted that the HIII-95M-PED IARV of 1167 lb falls at a logical place, size-wise, with respect to the previously reported HIII-5F (909 lb) and HIII-50M (1135 lb) ATD IARVs. The tight range of these IARVs could be explained by the fact they were developed from the same PMHS injury results. Additionally, the reported FAA-HIII 50M IARV (1223 lb) had a higher magnitude than the HIII-95M-PED IARV. This could be due to subtle differences in ATDs (e.g., construction, instrumentation, segmental mass distributions) or their responses to vertical loading. A review of ATD suitability conducted by Flath et al. (2022) noted that the pedestrian lumbar spine was composed of a stiffer variant of butyl rubber than the HIII-50M and FAA-HIII. It is postulated that despite the straight spine, the articulating pelvis of the HIII-95M-PED mitigates the load through the lumbar spine which could account for the lower IARV when compared to the FAA-HIII 50M. Moffatt et al. (2003) incorporated the pedestrian pelvis into HIII ATD testing since increased hip mobility offered more realistic ATD kinematics than the standard pelvis.

The IARC was developed using the peak axial compressive lumbar force as uncensored data in accordance with the methodology previously used (Lafferty et al., 2020; Lafferty et al., 2021). Further, uncensored data were determined to be most suitable for this study because the documented PMHS injury severities observed during post-test autopsy were a direct result of the entire exposure, eliminating the need for time of fracture. Censoring the data points associated with injury in any other fashion (left- or interval-censored) would force assumptions to be made about injury initiation and severity progression that cannot be confirmed. Furthermore, peak load has been commonly used as an injury prediction metric in previous research (Nightingale et al., 1996; Pintar et al., 1998; Ochia et al., 2003; Arun et al., 2014; Stemper et al., 2015; Stemper et al., 2018) and has been used as uncensored data points for survival analysis (DeVogel et al., 2019). Future work could be considered to improve IARC development by leveraging covariates like bone mineral density, age, weight, and interval censoring. However, both bone mineral density and age were evaluated as covariates in the development of the IARCs developed by Lafferty et al. (2020), but neither were found to be significant during survival analysis ($p < 0.05$). Furthermore, although the recommended IARV was developed from older specimens, this is general practice within the biomechanics community and typically results in conservative injury thresholds.

Future military rotary-wing aircraft and crashworthy seat development efforts should include injury assessment for Soldier populations. Military specifications should be updated to revise the current seat performance requirements. Including injury assessment for seat designers and program managers into the requirements will enable them to improve seated occupant protection. Occupant protection recommendations by the researchers who conducted this study provide guidance to improve occupant protection for the 95th percentile male during high vertical accelerative loadings. In the interest of providing a single pass/fail IARV limit, the authors suggest the use of IARVs assuming a 10% risk of an AIS 2 or greater injury (AIS 2+). An occupant exposed to AIS 3 would suggest more severe spinal injuries, with potential for more detrimental cord involvement. These serious injuries would certainly degrade a Soldier’s ability to self-egress or perform critical duties. AIS 2 type injuries still include any number of vertebral body fractures and minor injury to the spinal cord. In fact, an individual with a series of adjacent level vertebral body factures would have a very unstable spinal column; however, this set of injuries would only be coded as having AIS 2 level injuries (Appendix, Table A2). As such, the authors suggest only allowing a 10% risk of AIS 2 or greater injury (Lafferty et al., 2020). These metrics are only reflective of spinal injury and do not control for other chest injuries, like rib fractures. Further work that includes testing with similar stature PMHSs for the 95th percentile male to update or validate the IARCs is needed.

Conclusions

Military specifications should be revised to include ATD performance requirements that properly assess injury risk for all Soldier populations. The IARVs for the lumbar load cell of the HIII-95M-PED presented in this report will guide the assessment of injury risk for the seated 95th percentile male in vertical accelerative loadings during helicopter mishaps.

The IARCs were developed using the lumbar load cell in the HIII-95M-PED to assess vertebral body fractures categorized by AIS 2+ injuries. A 10% risk was chosen on the IARC to determine the IARV needed to mitigate the risk of a vertebral fracture in an impact event. It is recommended that vertical compressive lumbar loads should not exceed 1167 lb when testing with an HIII-95M-PED ATD to control for a 10% risk of AIS 2+ injury.

Recommendations

- An instrumented ATD intended for injury assessment should be required for current and future military rotary-wing and tilt rotorcraft aircraft and crashworthy seat development efforts. The ATD chosen should be consistent with the anthropometry of the population being tested and should have military relevant injury assessment capabilities.

- A 10% risk of moderate (or greater) thoracolumbar spinal injury (AIS 2+) is recommended in establishing IARVs for seat development and injury risk assessment.

- When a seated HIII-95M-PED ATD is exposed to dynamic vertical loading conditions, an axial compressive lumbar load performance limit of 1167 lb is the IARV recommended for a 10% risk of moderate and severe thoracolumbar spinal injury (AIS 2+).

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Appendix A. Post-Mortem Human Subject (PMHS) Test Data

Table A1. Demographics and Exposure Parameters for All PMHS Tests Reproduced from Lafferty et al., 2020

	PMHS ID	Age (yrs)	Lumbar BMD (mg/cm^3)	Weight (lb)	Stature (in)	Max. Accel (G)	Total Change in Velocity (ft/s)	Onset Rate (G/s)
Phase I: Exposure 1	1	73	85.5	184	69	21.0	41.6	1395
	2	63	131.4	198	70	21.4	41.9	1488
	3	70	156.6	202	76	21.5	42.0	1416
	4	55	135.3	185	71	20.7	41.1	1395
	5	39	122.4	166	65	21.5	42.0	1419
	6	39	160.8	118	71	21.3	42.0	1350
	7	67	115.7	158	69	21.4	42.1	1395
	Mean ± Std	58 ± 14	130 ± 26	173 ± 29	70 ± 3	21.3 ± 0.3	41.8 ± 0.4	1408 ± 42
Phase II: Exposure 1b	8	54	125.5	244	73	16.5	35.7	1154
Phase II: Exposure 2	9	72	64.1	167	72	21.5	41.6	1116
	10	58	74.4	175	70	*	*	*
Phase II: Exposure 3	11	64	123.9	169	75	16.3	41.9	1034
	12	63	166.5	217	72	16.1	42.1	1008
	13	65	89.4	167	72	16.2	42.3	976
	14	37	127.3	188	74	16.2	41.9	994
	Mean ± Std**	57 ± 14	126 ± 32	185 ± 23	73 ± 1	16.2 ± 0.1	42.0 ± 0.2	1003 ± 25

Note. Data acquisition system failure of all carriage mounted accelerometers during testing.
Note. Mean and standard deviations were calculated for Exposure 3 tests only.

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Table A2. PMHS Injury Counts

	PMHS ID	Vertebral Fractures	Disc Rupture	Disc Laceration	Spinal Ligament Laceration	Rib Fracture	Other Injuries	Highest Spine AIS*
Phase I: Exposure 1	1	1	0	0	0	3+	0	2
	2	5	0	0	0	3+	0	2
	3	3	0	0	0	3+	1	3
	4	5	0	0	0	3+	0	2
	5	5	1	0	0	3+	0	3
	6	0	1	0	4	1	0	1
	7	3	2	0	3	3+	1	3
Phase II: Exposure 1b	8	10	2	0	0	3+	0	3
Phase II: Exposure 2	9	3	1	0	0	3+	2	3
	10	1	2	0	0	2	2	3
Phase II: Exposure 3	11	0	0	0	2	1	1	1
	12	0	0	0	1	0	0	1
	13	1	0	2	2	3+	1	2
	14	2	0	0	0	0	1	2

Note. Column denotes highest AIS in the thoracolumbar and/or pelvic regions.

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