BIOFIDELITY EVALUATION OF THE THOR ADVANCED FRONTAL CRASH TEST DUMMY

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ABSTRACT

Ten 48 km/h frontal sled tests were conducted to evaluate the biofidelity of the THOR dummy. Three replicate tests were conducted with THOR, three with a 50th percentile male Hybrid III dummy, and four with post-mortem human surrogates (PMHS). The tests, conducted in a buck representing a mid-size US sedan, included a force-limited three-point belt system with buckle-side pretensioner and a driver side airbag. Following the tests, select parameters were used to compare THOR's responses with those of the Hybrid III dummy and the PMHS. The results were mass scaled in order to account for size differences between the subjects. Based on cadaveric response corridors, the sled test results provided evidence that THOR is more biofidelic than the Hybrid III dummy. THOR lap belt loads, upper spine (T1) movement, head acceleration, and movement of the anterior chest wall were more similar to those of the PMHS than were those of the Hybrid III. However, THOR recorded less head forward movement and pelvic acceleration than did the PMHS.

Key Words: biofidelity, cadavers, dummies, frontal impacts, sled tests

THOR (Test device for Human Occupant Restraint) is the latest in a line of advanced frontal impact dummies developed by NHTSA's multi-year development program. Increased biofidelity with respect to the Hybrid III dummy has been the primary objective. THOR's predecessors include the Prototype 50M (upgraded 50th-percentile Hybrid III male dummy) and TAD (Trauma Assessment Device) (Schneider et al. 1992, Rangarajan et al. 1998a, Xu et al. 2000). Using biomechanical data that was unavailable to the Hybrid III designers in the 1970's, THOR's face, neck, shoulder, thorax, spine, abdomen, pelvis, and femurs have been redesigned to be more humanlike in response to impact loading.

- THOR's neck construction allows more translational "head lag" with neck flexion than does the Hybrid III.
- THOR's thoracic spine includes a flexible rubber link whereas the Hybrid III's spine is rigid from the lumbar spine to the neck.
- THOR's rib cage has been designed to approximate human rib geometry and structure better than that of the Hybrid III (Fig.1). Enhanced instrumentation includes a three-dimensional thoracic deflection measurement system.
- THOR's pelvic structure and flesh are substantially different from those of the Hybrid III. THOR's flesh is segmented to allow full range of motion at the hip joint. The Hybrid III's extension of pelvic flesh to the proximal thigh reduces the femur range of motion.

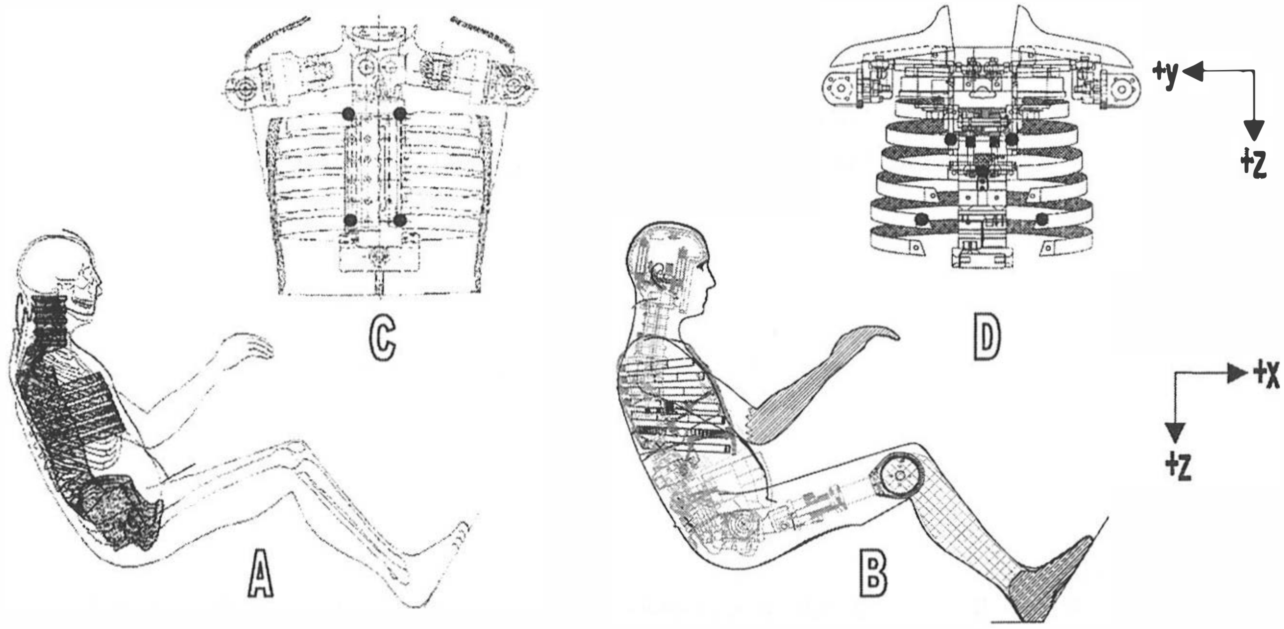


Figure 1. Test dummy construction. The left drawing (A) shows the Hybrid III dummy structure (shaded areas) overlaying a sketch of a 50th percentile male. Note the Hybrid III ribs are perpendicular to the spine while the human ribs angle downward as they do in THOR (B). Drawings C and D are front elevations of the Hybrid III and THOR chests, respectively. The dark circles indicate the anterior chest wall attachment points of the internal chest deflection linkages.

METHOD

Three replicate frontal sled tests were conducted for both the Hybrid III 50th percentile male and THOR dummies. Four additional tests were conducted using post-mortem human surrogates (PMHS) (Table. 1). The restraints used for the tests included a driver side airbag and a force-limited three-point belt system that incorporated a buckle-side pretensioner.

Table 1. Test Subject Information and Pre-Test Positioning

	Test ID	Sex	Mass (kg)	Age (yrs)	Height (cm)	Chest to wheel (cm)	Ave Knee to bolster (cm)
Hybrid III	HIII.532	M	84	-	173	29.0	7.1
	HIII.537	M	84	-	173	28.9	7.0
	HIII.538	M	84	-	173	28.6	6.7
THOR	THOR.539	M	80	-	173	30.5	4.0
	THOR.541	M	80	-	173	29.9	4.9
	THOR.542	M	80	-	173	30.2	4.9
PMHS	PMHS.533	F	63	67	163	30.1	6.4
	PMHS.534	M	51	47	175	30.2	4.3
	PMHS.544	F	56	59	169	30.5	4.4
	PMHS.545	M	73	67	184	30.8	3.5

EQUIPMENT: The tests were conducted using the sled system (Via Systems Model 713) at the University of Virginia's Automobile Safety Laboratory (UVA). The test fixture, or "buck", utilized in this test series was an approximation of the passenger compartment of a 1993 model year Ford Taurus. This configuration is typical of current automotive technology in terms of the spatial relationship between the occupant and the passenger compartment. The test buck was outfitted with an energy-absorbing steering column based on a modified production design. An adjustable knee bolster device was used to simulate the energy-absorbing characteristics of production knee bolster and dash assemblies. The seat was a Ford Tempo bucket seat equipped with an anti-submarining pan integral with the bottom cushion frame. The cushion frame and the anti-submarining pan were reinforced to allow multiple use while ensuring repeatable subject responses. Seat position was adjusted to accommodate the range of anthropometries required for PMHS. Positioning priority was given to maintaining a consistent chest to steering wheel hub distance for all occupants while providing realistic distances between the knees and knee bolster (Table 1).

The restraints used for the tests included a three-point belt system that incorporated a force-limited retractor (Autoliv ANG AGB 100 LL with custom 8.8 mm diameter torsion bar) and a 1 kN buckle-side pretensioner. This system was designed to maintain peak belt loads at the retractor in the range of 3.0 kN to 4.0 kN. The pretensioner system was deployed at 9 ms into the event and developed approximately 1.0 kN of tension in the belt at 15 ms. A non-depowered, tethered driver-side airbag (TRW, 1993 Taurus) was deployed 15 ms after the initiation of the impact event (T_0).

High speed photographic data was recorded by off-board high speed digital movie cameras arranged to record side views of the crash event for subsequent use in motion analysis. All cameras were operated at a speed of 1,000 frames/sec. Photo targets were placed at the occupant's ankle, knee, hip, shoulder joints, and the head center of gravity. A photo target also was placed on a bracket that allowed approximate tracking of the first thoracic vertebrae (T1).

Surrogates: Two 50th percentile male test dummies, the THOR and the Hybrid III, and four PMHS were used in the tests (Table 1). The PMHS were obtained through the Virginia State Anatomical Board with explicit permission given by the family to conduct impact biomechanics research. All test were approved by NHTSA's Human Use Review Panel (HURP) and all personnel involved in PMHS testing read and signed Ethical Treatment of Human Surrogate Forms supplied by the HURP. Screening of blood for Hepatitis A, B, C, and HIV was conducted with each PMHS prior to acceptance into the research program. The PMHS were preserved using a custom embalming technique (Crandall 1994). To simulate living conditions, pulmonary and cardiovascular pressurization was performed prior to testing.

Instrumentation: Sled deceleration and restraint belt loads were recorded for all of the tests. All test subjects were instrumented to record accelerations of body regions and chest deformation. The Hybrid III dummy instrumentation included a triaxial accelerometer mounted at the head center of gravity (CG), on the upper spine near the first thoracic vertebrae (T1), at the chest CG, and at the pelvic CG. Dynamic deformation data for the upper and lower thorax was determined using chestbands, non-invasive devices designed for the measurement of cross-sectional contours of the chest during an impact event (Eppinger 1984). The chestbands consisted of 40 strain gages mounted at 2.5 cm intervals in a steel band. Two chestbands were wrapped horizontally around the subject torso for each dynamic test. The chestbands were placed at the level of the second (upper band) and fifth (lower band) ribs. Additional chest deformation data was provided by an array of eight string potentiometers that recorded the x-y position of four points at the corners of the sternal plate with respect to the spine.

THOR instrumentation included triaxial accelerometers mounted at the head (CG), on the upper spine near T1, at the chest CG, on the lower spine near T12, and at the pelvic CG. Upper and lower chestbands were installed with the active area of the chestbands placed just above the level of the CRUX attachment points at the intersection of the sternal plate with the third and sixth ribs. These attachment points terminate in large bolt heads on the exterior of the sternal plate. In order to provide a more continuous bearing surface for the chestbands, we also padded the ribcage and the attachment points by installing strips of natural gum rubber, 7.6 cm wide x 0.95 cm thick, that encircled the ribcage. The dummy skin was installed over the gum rubber strips and, finally, the chestbands were positioned on top of the dummy skin. In addition to the chestbands, chest deformation data was provided by CRUXs, triaxial position sensors that provided the location of four points on the anterior thorax relative to the lower spine segment (Fig. 1).

Instrumentation for the PMHS included three uniaxial accelerometers installed on an angular rate sensor mounted on the rear of the head. Following the test, the triaxial head accelerometer data was transposed to the approximate center of gravity using information from the angular rate sensor. Similar instrumentation was used for the upper spine (T1). Triaxial accelerometers were mounted to the mid spine near the chest CG (T8 or T9), the lower spine, the second lumbar vertebrae (L2), and the posterior aspect of the pelvis. Upper and lower chestbands were taped to the chest at the level of the fourth and the eighth rib to provide chest deformation measurements about the ribcage, specifically near the heart and liver.

Electronic data was acquired at 10,000 samples/sec. using TRAQ-P, a DSP Technology Transient Acquisition and Processing System. The data was collected using IMPAX, a DSP technology PC-based data acquisition program. Raw force and acceleration data was processed by subtracting small initial offset values and filtering to either SAE J211-prescribed filter classes for the dummies or to NHTSA-prescribed filter classes for the PMHS.

THOR CRUX and Hybrid III string potentiometer data were processed using software provided by GESAC, Inc. and NHTSA, respectively. Output from the chestbands consisted of local curvature data from each chestband strain gage. This raw curvature data was filtered to SAE CFC-1000 and then chest deformation contours, calculated at each time step, were derived from this processed data using a variant of RBANDPC (Shaibani 1990). Using the contour data, local gage and sternum velocity were obtained using a four-point finite-difference approximation that is further filtered to SAE CFC-180.

RESPONSE SCALING, ALIGNING, AND PRESENTATION PROCEDURES: Due to the variability in subject geometry and inertial properties, the occupant responses were normalized to the standard anthropometry of the 50th percentile male weighing 75 kg. The normalization procedures of Eppinger et al. (1984) were used to perform the scaling. This procedure assumes that the mass density and modulus of elasticity are constant between test subjects. The scaling variable based on occupant mass (M) in kg is shown in equation (1).

$$\lambda = (75 / M_i)^{1/3} \qquad (1)$$

The scaled test parameters, denoted with subscript s, can then be expressed in terms of the initial parameters, denoted with subscript *i*, and the scaling factor (Eq. 2 to 6).

Velocity:	$V_s = V_i$	(2)
Acceleration:	$A_s = \frac{A_i}{\lambda}$	(3)
Length:	$L_s = \lambda \times L_i$	(4)
Time:	$T_s = \lambda \times T_i$	(5)
Force:	$F_s = \lambda^2 \times F_i$	(6)

Time-History Data Processing and Presentation: After the time-history data was mass and time scaled, we used an aligning procedure to further reduce variability due to test-to-test and subject-to-subject differences. The procedure involved shifting the curves in time in order to minimize the cumulative variance (Morgan, Marcus, and Eppinger 1981) (Fig. 2). In all cases, the alignment procedure required temporal shifts of less than 11 ms.

The next step involved the creation of response corridors. The data, having been mass and time scaled and aligned, was averaged for each subject type (Hybrid III, THOR, PMHS) and the standard deviation was calculated at each time step. The response corridor was created by plotting the average values ± one standard deviation (Fig. 3).

Subject Movement Data Processing and Presentation: The method used to process and present head and T1 movement data in spatial coordinates involved plotting the x and z coordinate trajectories of the head and TI. Analysis of high-speed video images provided the position at 10 ms intervals. The first step involved mass and time scaling the movement in the x and z directions with respect to time (See equations 4 and 5 above).

Response corridors were calculated using the average position at each time step for each subject. In order to define the bounds of the corridor, the standard deviation was calculated independently for movement in the x and z directions. Using the square root of the sum of the x and z standard deviations squared, we calculated a radius of uncertainty surrounding the average location at each time step. The radius was used to construct circles about each point. We formed the corridors by drawing upper and lower bound lines tangent to the circles (Fig 3).

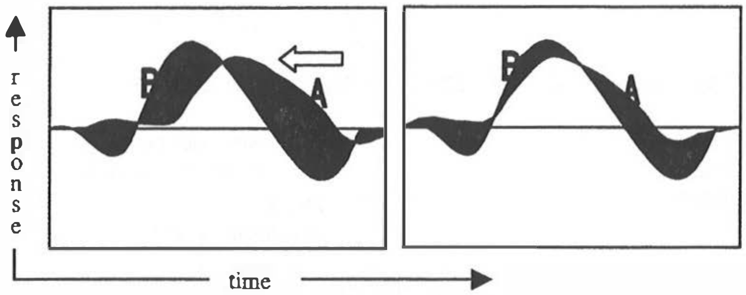


Figure 2. Aligning procedure. Curve A is shifted along the time axis toward curve B to minimize the cumulative variance between them (shaded area).

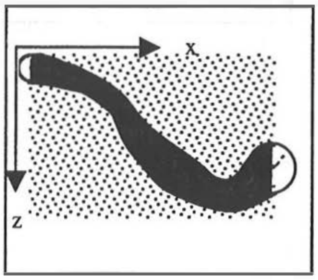


Figure 3. Creation of corridors for subject movement trajectories. The dotted line passes through the average position for each time step.

RESULTS

The testing hardware and occupant restraint systems performed consistently throughout the test series. There was little test-to-test variation in the change in velocity due to the impact (Delta V) (48.1 – 49.2 km/h) and the peak deceleration (16-17 g) (Fig. 4).

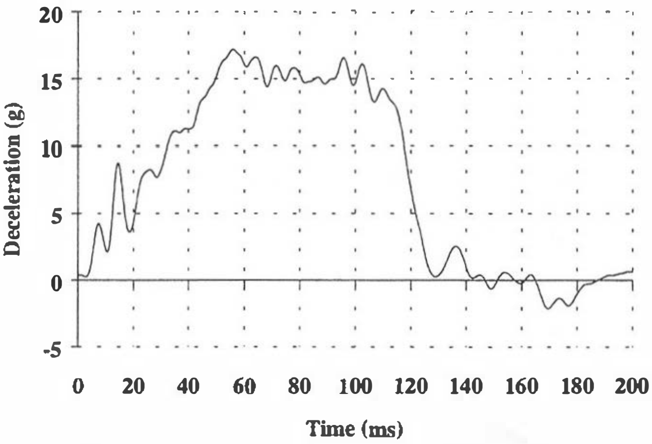


Figure 4. Typical sled deceleration time/history.

Figure 5 summarizes the sequence and approximate timing of events during the crash. The buckle-side pretensioner was deployed 9 ms after the crash began (T_0). This resulted in local maxima of the restraint belt loads at approximately 15 ms. Despite the use of a tethered air bag and a 30 cm nominal chest-to-air bag distance, a lobe of the deploying air bag struck the upper chest in all tests. This impact, referred to as "bag slap", began at 24-26 ms but subsided with continued filling of the air bag. The shoulder belt load rose rapidly from 50 to 70 ms as the subject moved forward. The load then remained constant or increased at a reduced rate from 70 to 120 ms during yielding of the force-limiting retractor. The knees began to load the knee bolster at approximately 60-65 ms. The lap belt load peaked at 80 ms. Peak head and chest resultant acceleration occurred between 90 and 110 ms. The forward movement of the head CG and the upper spine (T1) reached a maximum between 110 and 120 ms.

The following summarizes test results regarding restraint loads, subject movement, acceleration, and chest response.

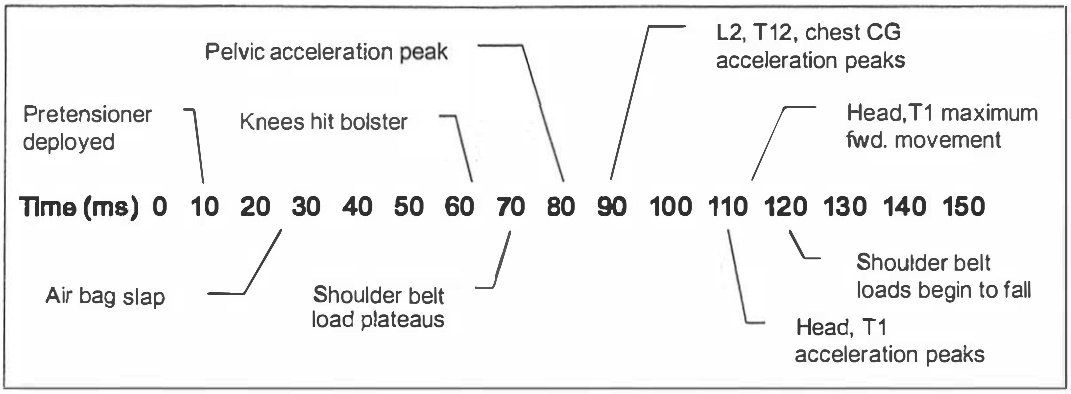


Figure 5. Event chronology. Times are approximate as they varied with test subject type in some cases.

RESTRAINT LOADS: Figure 6 presents shoulder and lap belt load-time histories. The shoulder belt loads are presented unscaled and unaligned because the force-limiting mechanism in the retractor spool produced loads that were generally insensitive to subject type or mass. Although the three-point belt used a single length of webbing for lap and shoulder restraint, the belt hardware effectively isolated the lap belt section from the shoulder belt section and its force-limited behavior. The lap belt loads were mass and time scaled and aligned as indicated above. The Hybrid III dummy developed much higher lap belt loads than did THOR or the PMHS.

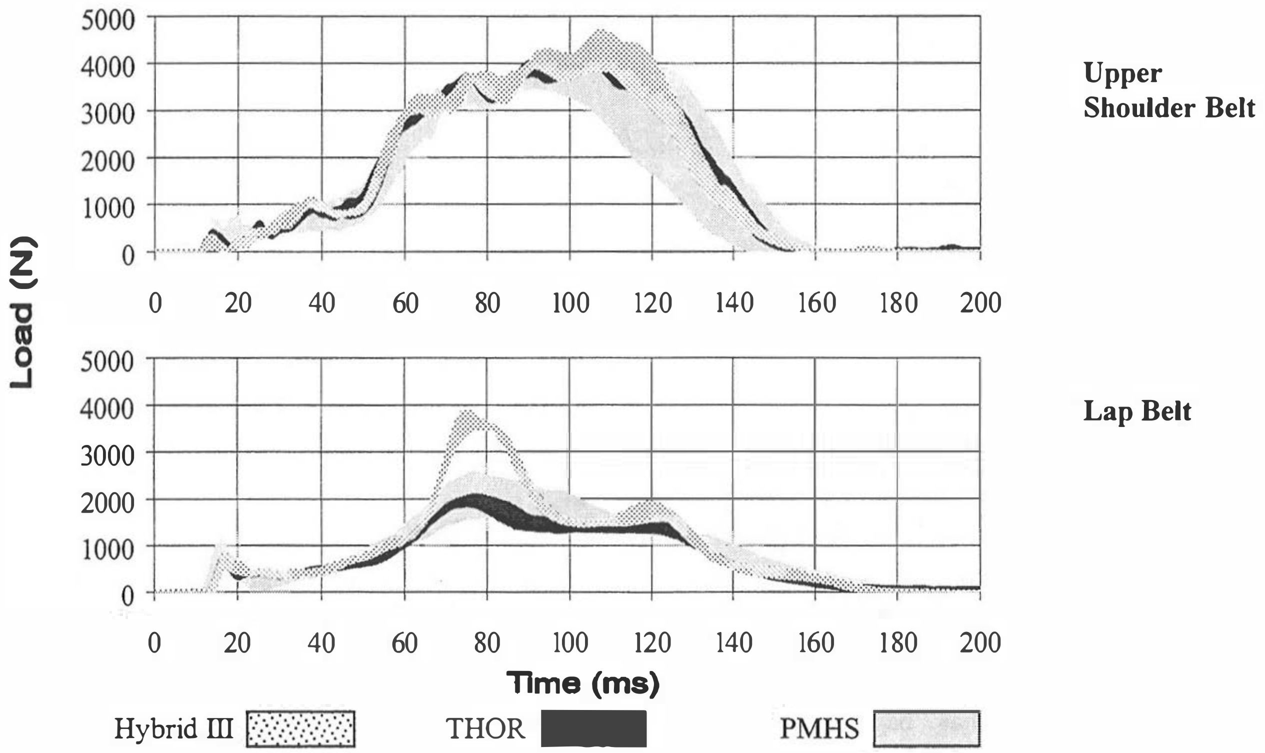
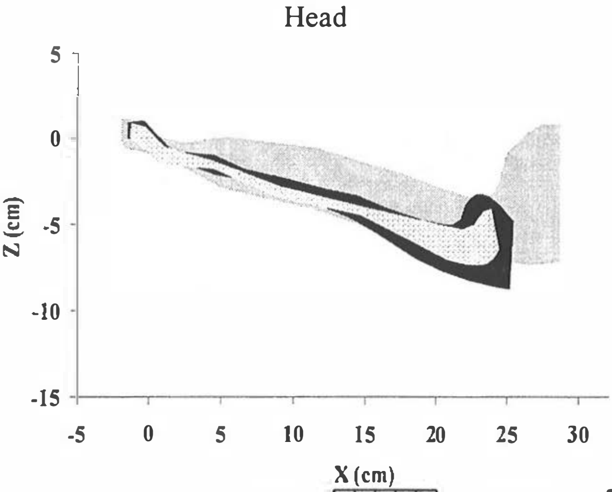


Figure 6. Restraint belt loading. The shoulder belt data was not mass-scaled.

SUBJECT MOVEMENT: Figure 7 plots the translation of the head and first thoracic vertebrae (T1) from T_0 through the beginning of subject rebound movement at approximately 120 ms. The general forward movement of THOR's head closely matched that of the Hybrid III and the trajectories of both dummy heads were very repeatable. After mass-scaling, the average PMHS maximum head forward movement was approximately 4 cm greater than that of the dummies.

The trajectory of THOR's surrogate first thoracic vertebrae, T1, was more similar to that of the PMHS than to that of the Hybrid III (Figs. 8 and 9). High speed video analysis indicated that the average PMHS T1 target moved down and then sharply up between 75 to 100 ms. THOR's T1 movement between 90 to 110 ms was similar but the upward movement was not as pronounced. In the Hybrid III tests, the T1 movement pattern was similar to the THOR and PMHS tests in that there was initial downward movement of T1. However, there was very little upward movement of T1 later in the event.

The greater upward movement of T1 in the PMHS and THOR tests was reflected in greater peak T1 z-axis accelerations that accompanied the movement. Scaled and averaged T1 z-axis acceleration was 21 g for the PMHS subjects, 15 g for THOR, and only 4 g for the Hybrid III.



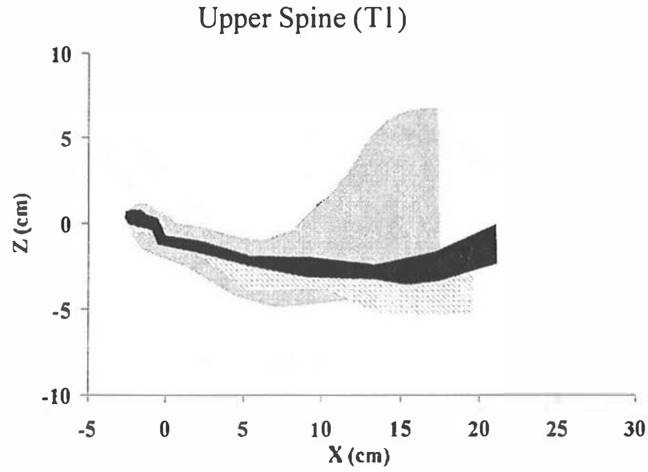


Figure 7. Head trajectory corridors. Forward movement is along the positive x-axis. Downward movement is along the negative z-axis.

Figure 8. First thoracic vertebrae (T1) trajectory corridors.

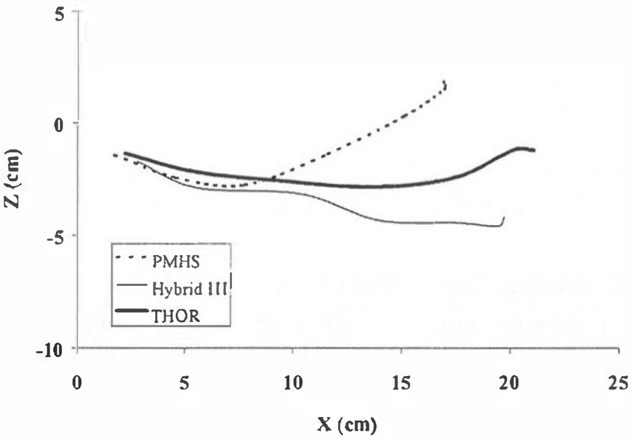
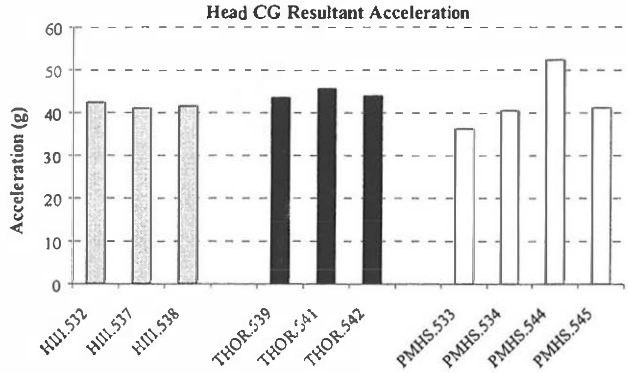


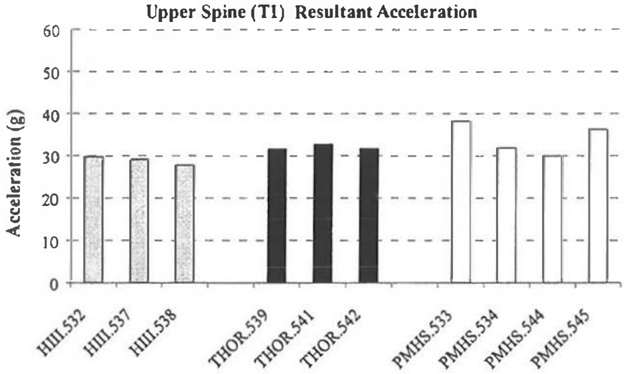
Figure 9. Average upper spine (T1) trajectories between 70 ms and 120 ms after T_o.

SUBJECT RESULTANT ACCELERATION: Figure 10 presents the maximum resultant subject accelerations after mass-scaling. Test-to-test repeatability was very good for both dummies. Repeatability was considered good for the PMHS despite the individual differences not accounted for by the mass scaling technique.

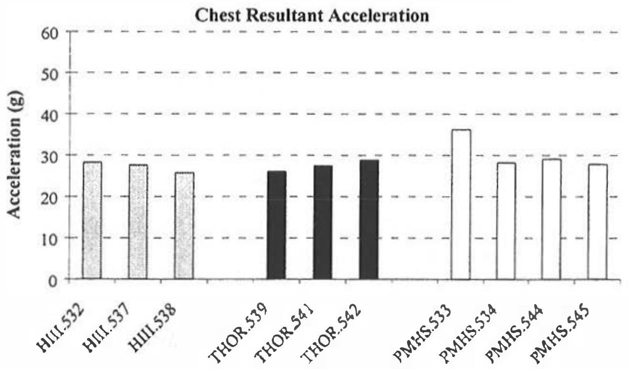
Figure 11 presents mass-scaled and aligned resultant acceleration-time histories. In general, the subject resultant acceleration traces were quite similar. THOR and Hybrid III head, upper spine (T1), and chest CG acceleration-time histories closely matched those of the PMHS. THOR lower spine (T12) accelerations closely matched those of the PMHS lower spine (L2), especially during the "loading" phase of the event (50 – 75 ms). THOR's resultant pelvic acceleration was significantly lower than that of the Hybrid III and was lower than the PMHS average. This was due to lower absolute z-axis acceleration (Table 2). THOR's z-axis peaks indicated differences in the direction of pelvic acceleration. THOR's average maximum value of +9 g indicated that the peak response reflected the pelvis being accelerated downward. In contrast, the Hybrid III and PMHS peak responses of –20 g and – 33 g indicated upward pelvic acceleration.



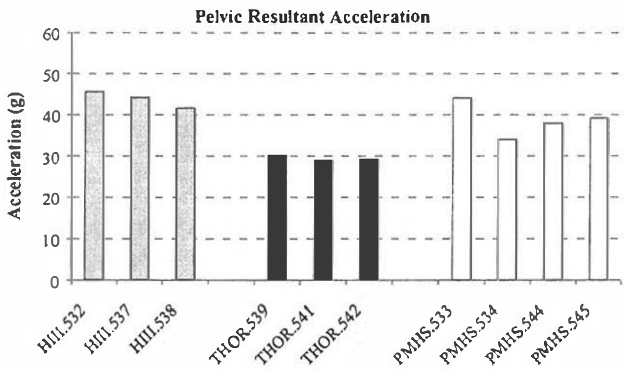
	HIII	THOR	PMHS
Ave. (g)	42	44	43
SD	0.7	1.1	6.9
CV (%)	1.6	2.5	16.2



HIII	THOR	PMHS
29	32	34
1.0	0.7	3.8
3.5	2.1	11.3



	HIII	THOR	PMHS
Ave. (g)	27	27	30
SD	1.3	1.4	4.0
CV (%)	4.8	5.0	13.1



HIII	THOR	PMHS
44	29	39
2.0	0.7	4.1
4.6	2.2	10.7

Figure 10. Peak resultant acceleration values after mass-scaling.
SD – standard deviation. CV – coefficient of variation = (SD/Average)*100.

Table 2. Average Mass-scaled Peak Z-Axis Pelvic Acceleration (g)

	Maximum (g)	Time (ms)	Minimum (g)	Time (ms)
Hybrid III	2	15	-20	87
THOR	9	67	-3	96
PMHS	5	19	-33	81

Acceleration-time histories of the head and chest reflected belt pretensioning and air bag slap effects. The buckle pretensioner created a peak response in the lower shoulder belt tension gage at approximately 15 ms. The most significant event that occurred before substantial forward subject movement was the air bag slap at approximately 25 ms. In response to the air bag slap, the THOR head acceleration exhibited rapid short duration peaks between 30 to 40 ms after T_0. The PMHS head acceleration exhibited a similar more pronounced series of peaks lasting from 25 to 60 ms. Hybrid III head acceleration remained below 5 g during these time periods. The PMHS chest acceleration exhibited a prominent 20 g spike occurring after the time of the air bag slap. Although there were spikes in the dummies' chest acceleration that occurred at the same time, they were roughly a third of the PMHS value.

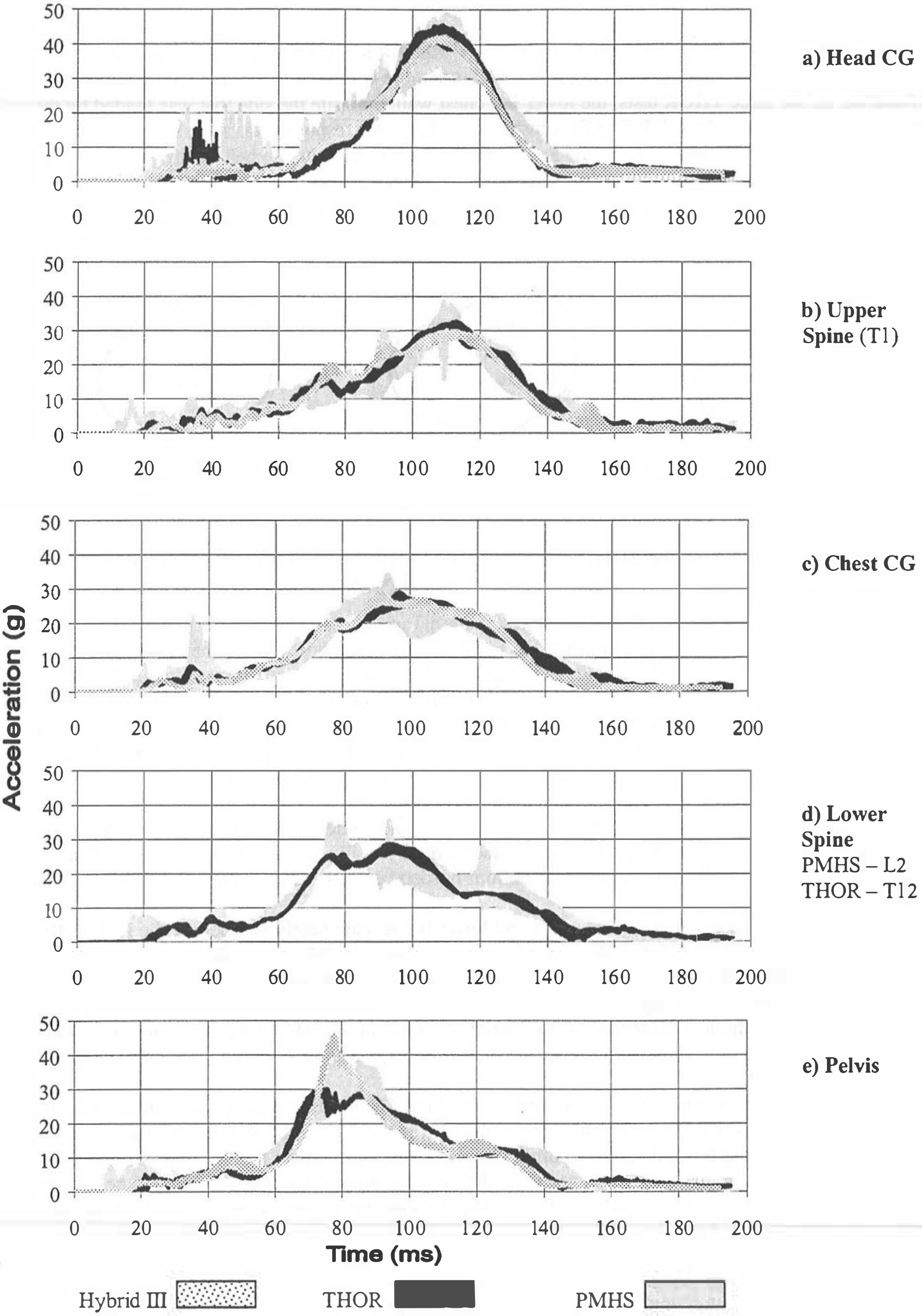


Figure 11. Subject resultant acceleration, mass and time scaled and aligned.

Movement of the Anterior Chest in Response to Restraint Loading: The movement of THOR's lower left anterior chest wall was more similar to that of the PMHS than of the Hybrid III. In the three THOR tests, the lower left chest wall, opposite the side that was loaded by the shoulder belt, moved outward during the crash event. A similar response was seen in 3 of 4 PMHS tests (Fig. 12). In the Hybrid III tests, the lower left chest wall, opposite the belt loading on the lower right side, moved toward the spine and no outward movement was recorded.

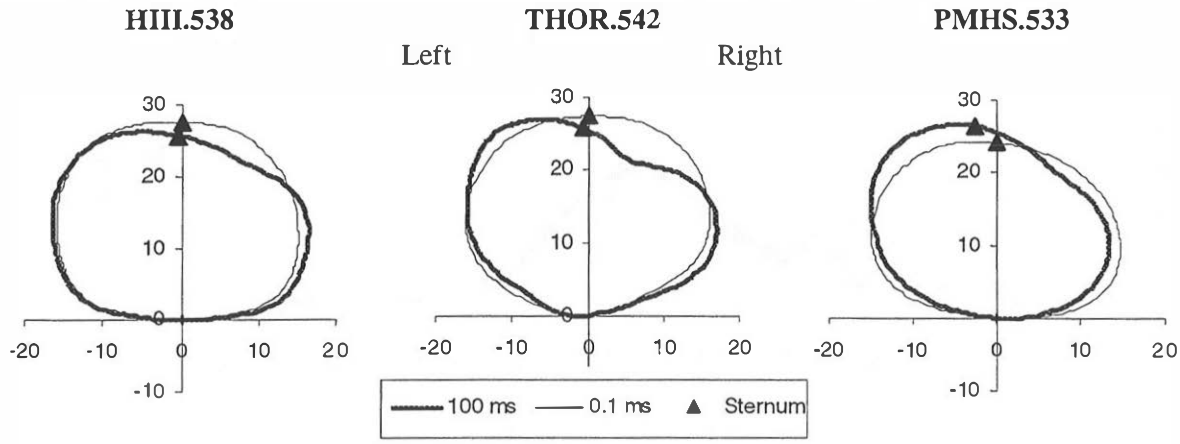


Figure 12. Representative lower Hybrid III, THOR, and PMHS chestband contours showing left side bulging out for THOR and PMHS.

Analyses of the results of the dummies' internal chest deflection instruments confirmed differences between THOR and the Hybrid III. Figure 13 plots the positions of the upper and lower CRUX attachment points for THOR and the string potentiometer attachment points for the Hybrid III in representative tests. The attachment points lie at the intersection of the ribs with the sternum. Lines drawn between the points define the upper and lower borders of the sternum (See Fig. 1). The plots represent a view of the chest from above the subject with the mid-sagittal plane and sternum located at the origin. Negative movement along the vertical axis indicates chest compression.

If the sternum moved as a rigid plate during the crash event, the CRUX attachment points and any lines drawn between them must lie on this plate or plane. When viewed from above, the upper and lower borders of the sternum may be displaced differentially relative to the spine, but must remain parallel to each other. If the top and bottom borders are not parallel, the "sternal plate" has deformed out-of-plane.

At 30, 80, and 140 ms after T_o, the upper and lower borders are nearly parallel for the Hybrid III. The attachment points on the right side have moved consistently farther inward than those on the left. THOR's upper and lower borders are not parallel at 30 ms and the upper left point is deflected more than the right. At 80 ms, the angle between the borders is reduced and the right-side deflections are greater than those of the left. Note that the lower left point has moved above the horizontal axis. This indicates the "bulging out" illustrated in Figure 12. At 140 ms, the angle between the upper and lower borders has increased.

Figure 14 illustrates this independent movement of the THOR attachment points with respect to time. The analysis to create this plot involved defining a plane (i.e., rigid plate) using the recorded positions of three of the four attachment points. We then calculated where the fourth point would be located in order to lie on the defined plane. The calculated position of the fourth point was compared to the recorded position (in the x-axis; see Figure 15). The greater the difference in the positions, the greater the out-of-plane behavior of the sternum. We arbitrarily chose to predict the position of the lower right attachment point. Positive values indicate that the lower right attachment point was farther away from the spine than predicted. Negative values indicate that the point was closer to the spine than predicted. A zero value indicates that the point lies within the defined plane.

The Hybrid III average value of the fourth point relative to the calculated plane is fairly constant and deviates from zero less that 0.5 cm, while THOR's average peak value varies from −1.5 cm at 30 ms to +2.25 cm at 140 ms. The difference in out-of-plane behavior between the two dummies that occurred between 15 ms and 50 ms coincided with belt pretensioning and air bag slap. The Hybrid III values were approximately zero during these events suggesting that the sternum displaced as a rigid plate. THOR values indicated significant out-of-plane behavior beginning at approximately 17 ms. By 23 ms, the value had reached -0.5 cm, presumably as a result of the pretensioning of the shoulder belt. After a brief plateau at 23 ms, the value sharply decreased to approximately –1.5 cm at 30 ms. This behavior could be explained by an off-center bag slap to the chest. CRUX deflection results indicate that the bag impact area was nearest to the upper left CRUX attachment point. THOR's second period of out-of-plane behavior began at approximately 80 ms and peaked at 140 ms.

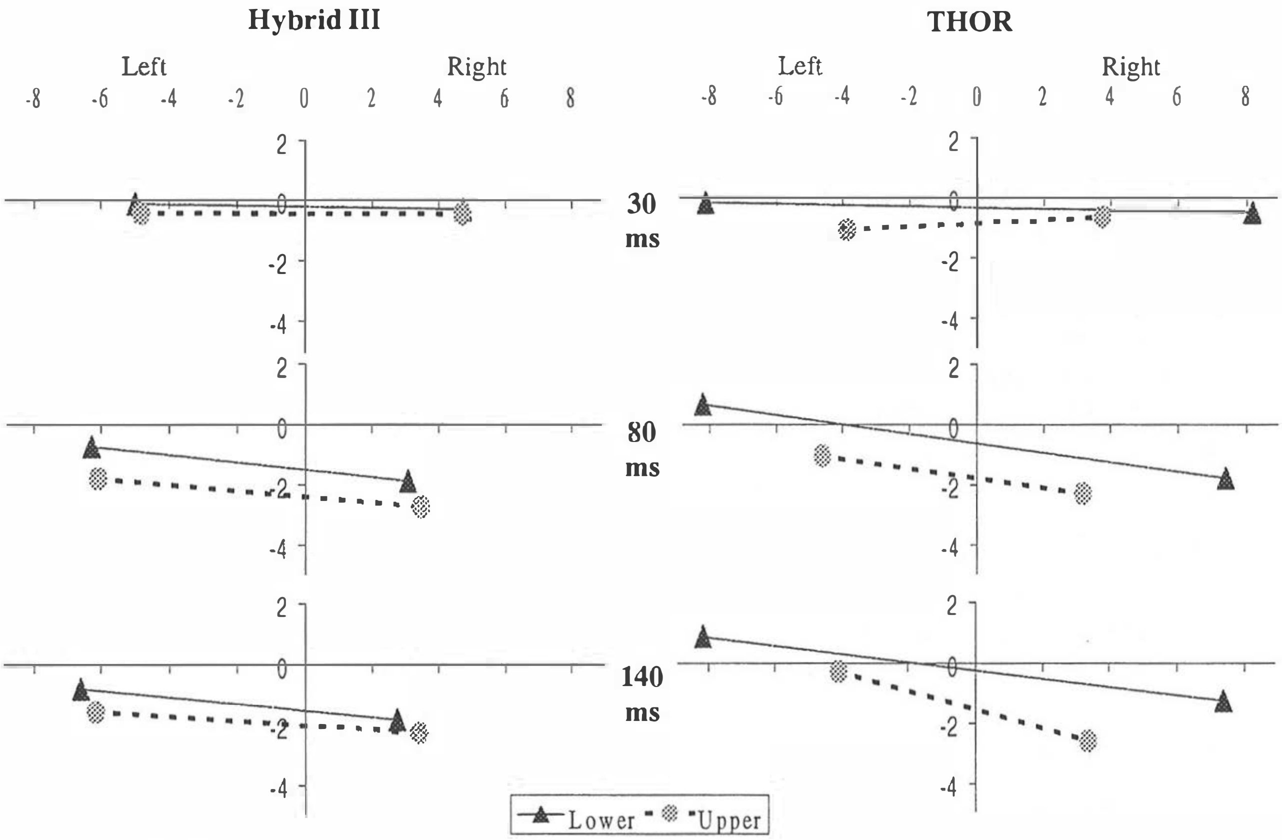


Figure 13. Movement in centimeters of the upper and lower anterior chest wall as viewed from above the test subjects in tests HIII.532 and THOR.539.

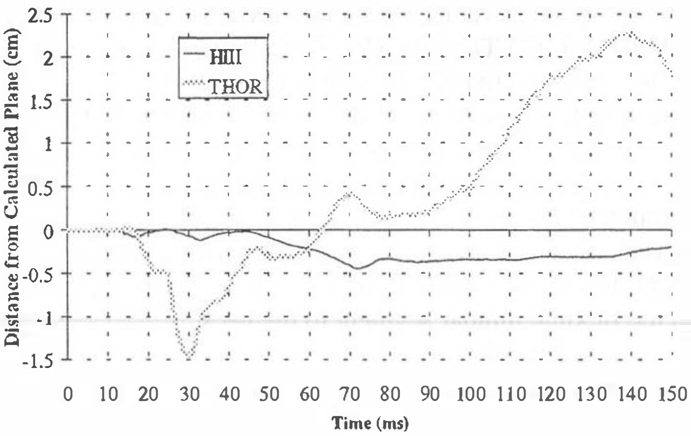


Figure 14. Out-of-plane behavior. The curves indicate the average distance of the lower right attachment point from the plane defined by the three other attachment points.

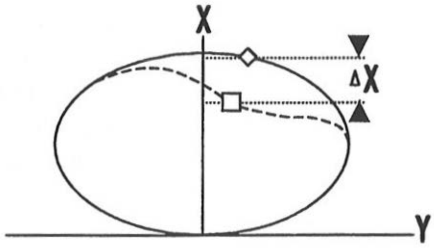


Figure 15. Definition of chest deflection: displacement along the x-axis toward the spine. The coordinate system is oriented so that "x" is in the direction of the sled travel. The diamond represents the location of a point on the front of the chest at T_0 and the square represents this point after chest deflection has occurred.

SUMMARY OF FINDINGS

1. The Hybrid III dummy developed much higher lap belt loads that did THOR or the PMHS.
2. Dummy head movement was similar and very repeatable test-to-test. The PMHS head trajectories exhibited more variability and the head moved approximately 4 cm further forward in the x-direction than did the dummies'. The trajectory of THOR's upper spine (T1) was more similar to that of the PMHS than to that of the Hybrid III. The greater upward movement of T1 in the PMHS and THOR tests was reflected in greater peak T1 z-axis accelerations that accompanied the movement.
3. Maximum resultant accelerations were repeatable from test-to-test for both dummies. PMHS repeatability was acceptable. In general, the resultant acceleration maxima and time history traces were quite similar. THOR resultant pelvic acceleration was significantly lower than that of the Hybrid III and was lower than the PMHS average due to low z-axis acceleration.
4. Data from internal instrumentation indicated that THOR's average peak upper chest deflection values were consistently lower than those of the Hybrid III.
5. The movement of THOR's lower left anterior chest wall was more similar to that of the PMHS than that of the Hybrid III. In the three THOR tests and three of the four PMHS tests, the lower left chest wall moved outward during the crash event. No such outward movement was recorded for the Hybrid III. An analysis of the dummies' internal chest deflection data suggested that the THOR sternum and anterior ribcage is more flexible than that of the Hybrid III.

DISCUSSION

Several THOR responses, including head acceleration, upper spine (T1) movement, anterior chest wall movement, and lap belt loads, were more similar to those of the PMHS than were those of the Hybrid III. Other THOR responses did not match those of the PMHS. Both dummies recorded lower peak forward head excursion than the PMHS. THOR's peak pelvic acceleration was substantially less than either the PMHS or the Hybrid III. We were unable to determine whether THOR's maximum chest deflection values were more PMHS-like than the Hybrid III.

RESPONSES SIMILAR TO THE PMHS: THOR's neck, torso, and pelvis produced responses to restraint system loading that approximated those of the PMHS.

Head acceleration results suggested that the THOR and the PMHS head and neck complexes were more sensitive than the Hybrid III's to events early in the crash such as restraint system pretensioning and air bag slap.

Substantial differences in torso construction between the two dummies contributed to THOR's more PMHS-like movement of the upper spine and anterior chest wall deflection. THOR's more biofidelic T1 movement may be due to the flexible joint in the dummy's thoracic spine. The Hybrid III spine is rigid from the flexible lumbar segment to the neck.

Chestband and internal deflection instrumentation results indicated that THOR's sternum and anterior ribcage is more flexible than that of the Hybrid III. The observed outward movement of THOR's lower chest wall, observed in the PMHS tests and considered an indicator of sternum flexibility, has been well documented for frontal sled tests with three-point belt and combined belt and air bag restraints (Schneider et al. 1992, Rangarajan et al. 1998a,b, Ito et al. 1998, and Xu et al.

2000). Improved response of the lower ribcage may improve injury assessment for soft tissue organs protected by the ribs such as the liver.

The out-of-plane movement of points defining the sternal borders is further evidence that THOR's sternum is more flexible than that of the Hybrid III and that it responded differently to restraint system loading. This finding is consistent with observable differences in sternal construction. THOR's sternal plate is much larger in area and fabricated entirely from flexible plastic. In contrast, the Hybrid III's sternum is comprised of a small rectangle of rigid plastic attached to the rib ends with a narrow, flexible plastic hinge.

Although we would expect out-of-plane sternal movement patterns in the PMHS, the chestbands did not provide enough information regarding the position of points on the anterior chest wall to conduct the necessary analysis. Unlike the dummies' internal instruments that recorded the position of points on the upper and lower anterior chest wall relative to a common rigid spine segment, the chestbands installed on the PMHS recorded anterior chest wall position relative to separate spine segments. These spine segments were free to move relative to one another and did not provide the common reference necessary to relate the position of points on the upper chest to those on the lower chest.

Lower lap belt loads for THOR were also reported by Rangarajan et al. (1998b) and by Honda R&D (1999). In comparative frontal three-point belt and air bag sled tests of the two dummies, Honda recorded lap belt loads of 3.2 kN for THOR and 5.9 kN for the Hybrid III. The authors attributed the difference to THOR's more flexible torso (due to the additional flexible joint in the upper spine). We attributed THOR's lower and more PMHS-like lap belt loads to its segmented-flesh pelvic construction that may have reduced the effective pelvic mass with respect to the Hybrid III. Rangarajan et al. (1998b) reached a similar conclusion.

RESPONSES DISSIMILAR TO THE PMHS: THOR's responses for head excursion and pelvic acceleration were notably different from those of the PMHS.

Both THOR and the Hybrid III recorded less forward excursion of the head and more forward excursion of the upper spine (T1). This may be due to the dummy's simplified spine that uses one (Hybrid III) or two (THOR) flexible joints to model the human thoracic and lumbar spine that is comprised of 19 intervertebral joints.

Although THOR produced lap belt loads similar to those of the PMHS, THOR's resultant pelvic accelerations were lower than those of both the PMHS and of the Hybrid III (THOR: 29 g, PMHS: 39 g, Hybrid III: 44 g). The differences in resultant pelvic acceleration likely were due to a number of interrelated factors including differences in accelerometer orientation, pelvic construction, pelvic rotation, and pelvic interactions with the lap belt, the seat, and/or the knee bolster (via femur loading). The most notable difference in pelvic response among the subjects was z-axis acceleration. THOR recorded a much lower absolute value than either the PMHS or the Hybrid III. Similar results were reported by Honda R&D (1999) for 56 km/h frontal barrier tests in which THOR recorded much lower absolute driver-side pelvic z-acceleration (THOR: 13 g, Hybrid III: 32 g).

After verifying our pelvic accelerometer data, we examined possible causes for the low THOR z-axis acceleration including differences in accelerometer orientation with respect to the laboratory reference frame and pelvic rotation about the y-axis. The results could be explained by assuming that the posterior tilt of the Hybrid III pelvis averaged 20 degrees more than that of THOR. Differences in pelvic tilt between the two dummies may have been due to THOR's more flexible lumbar spine and more mobile hip joint. Evidence of significant differences in the lower torso and pelvis of the two dummies included substantially different lumbar shear force (x-axis) and y-axis moment time-histories recorded in comparative Hybrid III and THOR 56 km/h frontal barrier tests (Xu et al. 2000). Confirmation that differences in pelvic tilt caused differences in pelvic z-axis acceleration would require more accurate monitoring of the pelvic accelerometer orientation.

Chest Deflection: We were unable to determine which of the two dummies best approximated the peak chest deflection of the PMHS due to a lack of confidence in the accuracy of the chestbands. A post-test analysis of THOR's chest deflection revealed a discrepancy between the peak deflection recorded by the chestbands and the CRUXs, internal instruments that recorded ribcage deflection accurately in prior tests. The upper chestband peak deflection (minus the estimated skin and rubber strip deflection) averaged 36% greater than that of the CRUX. Analysis of the Hybrid III data suggested that the chestband values averaged 20% greater than the actual peak deflection (Fig. 16). This analysis used the results from Hagedorn (1993) deflection tests that involved indentor loading of a Hybrid III torso instrumented with chestbands and string potentiometers. The results of tests in which the chestband contours accurately mapped the actual contours were used to develop a relationship between the actual overall deflection (for both the ribcage and the overlying padded skin) and the string potentiometer deflection (ribcage only). Using this relationship and the string potentiometers results in our Hybrid III sled tests, we estimated that the chestband overestimated the actual overall deflection by approximately 20%.

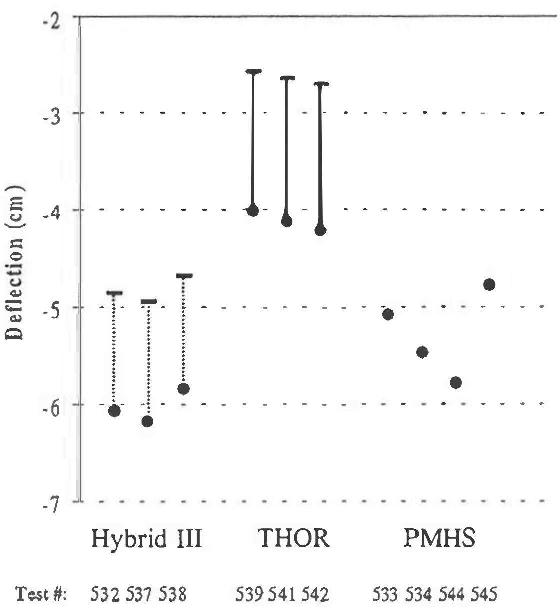


Figure 16. Chestband upper chest deflection at the CRUX location with the greater deflection. The peak values for each test (circles) were mass-scaled. The upper bounds (horizontal lines) indicate the peak value + 20% of the peak value for the Hybrid III and +36% for THOR. The tracked points corresponded with points 3.8 cm from the sternum centerline. These locations correspond to the equivalent location of the THOR upper CRUX attachment points on the anterior rib cage (Fig. 1) This point was on the right side for both dummies, but varied from the right to the left for the PMHS due to variation in the location of belt loading on the anterior surface of the chest. This variation in the point of belt loading resulted from differences in the subjects' anthropometry.

Although the chestbands overestimated chest deflection for both dummies, we were unable to assume that this relationship held for the PMHS. Prior studies do not corroborate the chestband's overestimation of peak chest deflection. Results from a chest deflection test conducted with a single PMHS statically loaded by a simulated roped shoulder belt indicated good agreement (within 3 percent) between the peak deflection recorded by the chestband and the actual deflection (Shaw et al. 2000). Reported results from five PMHS subjects who were subjected to dynamic shoulder belt loading indicated that the peak chestband deflection values were similar to those of the actual values (Tests THC 75, 77, 79, 91, 93; Cesari and Bouquet 1994). Bass, Wang and Crandall (2000) found that even small discontinuities on the surface of the ribcage could cause the chestband to substantially overestimate or underestimate actual chest deflection. Because PMHS ribcage surface features are very different from those of the dummies, this suggests the potential for differences in chestband response and calculations of chest deflection.

Although we were unable to compare the dummies with the PMHS with respect to peak chest deflection, the chestband results suggesting that THOR's upper chest deflected less than that of the Hybrid III were supported by data from the internal sensors (i.e., THOR CRUXs and Hybrid III central chest slider and string potentiometers).

Table 3 and Figure 17 summarize the internal deflection data from the tests reported in this paper as well as previous similar UVA tests and tests reported by other researchers. Each test series, comprised of one to three replicate tests with each dummy, subjected THOR and the Hybrid III to the same crash conditions. The dummies were seated in the driver position in all cases. With the exception of the full-width barrier test at 56 km/h (Xu et al. 2000), all were frontal sled tests. Other differences among the tests included Delta v (48 km/h or 56 km/h) and restraint system (three-point belt with air bag or force-limited three-point belt with air bag). Left and right chest deflection was compared using the CRUXs and string potentiometers. Deflection of the center of the sternum, recorded by the Hybrid III's chest slider, was estimated for THOR by averaging the deflection values of the upper left and right CRUXs. Errors using this method to estimate mid-sternal deflection may be substantial if THOR's flexible sternum is loaded locally. In the Hybrid III test reported by Xu et al. (2000), for which chest slider data was not reported, the center sternum deflection was estimated by averaging the deflection values of the upper left and right string potentiometers. Figure 17 presents the ratio of Hybrid III to THOR upper chest sternal deflection.

Table 3. Summary of Hybrid III and THOR Upper Chest Deflection Data from Internal Instruments (cm)

Test Ref.	Source	Test Site	Test Type	Restraint	Right		Center		Left	
					HIII	THOR	HIII	THOR a	HIII	THOR
A	UVA	Auto Safety Lab	48 S (3, 3)	3pt FL + AB	-3.5	-2.6	-2.7	-1.9	-2.3	-1.2
B	UVA	Auto Safety Lab	56 S (3, 1)	3pt FL + AB	-4.1	-3.5	-3.7	-2.6	-2.8	-1.7
C	Rangarajan et al. 1998a	Autoliv	56 S (3, 3)	3pt FL + AB	NR	-4.8	-4.0	-4.5	NR	-4.1
D	UVA	Auto Safety Lab	56 S (3, 2)	3pt + AB	-6.4	-5.3	-5.1	-3.5	-2.3	-1.6
E	Rangarajan et al. 1998a	Volvo	56 S (2, 2)	3pt + AB	NR	-4.4	-4.6	-3.5	NR	-2.6
F	Rangarajan et al. 1998b	VRTC	48 S (1, 1)	3pt + AB	NR	-4.5	-4.8	-4.1	NR	-3.6
G	Ito et al. 1998	JARI	48 S (1, 1)	3pt + AB	NR	-4.4	-4.8	-4.4	NR	-4.4
H	Rangarajan et al. 1998a	Autoliv	56 S (3, 3)	3pt + AB	NR	-4.9	-4.4	-4.4	NR	-3.9
I	Xu et al. 2000	GM	56 B (1, 1)	3pt + AB	-4.0	-4.3	-3.3[b]	-4.1	-2.5	-3.8

Key:

48 S (3, 3)	48 km/h frontal sled tests; 3 replicate tests with each dummy.
48 S (1, 1)	48 km/h frontal sled tests; one test with each dummy.
56 S (3, 1)	56 km/h frontal sled tests; 3 replicate tests with THOR, 1 with H3.
56 S (3, 2)	56 km/h frontal sled tests; 3 replicate tests with THOR, 2 with H3.
56 B (1, 1)	56 km/h NCAP barrier test; one test with each dummy.
3pt + AB	3-pt. Belt and Air Bag
3pt FL + AB	3-pt. Force-limited Belt and Air Bag
NR	Not reported.
a	Approximated by averaging the upper right and left CRUX values.
b	Approximated by averaging the upper right and left string potentiometer values.

Notes: Values for Xu and Rangarajan (b) tests were scaled from the report graphs. Comparison between the Hybrid III and THOR left and right sternal deflection values may be misleading due to the different locations of the string potentiometer and CRUX attachment points on the anterior chest. The string potentiometer attachment points are attached to the first rib whereas the CRUXs are attached to the second rib. The string potentiometer attachment points are approximately 1 cm closer to the mid-saggital plane (See Figure 1).

Five of the eight previous studies surveyed found that THOR's ribcage deflected less than the Hybrid III's, in support of the finding reported in this paper. However, there was substantial variation in the magnitude of the differences. In these five studies, the Hybrid III recorded 9 to 46 percent greater sternal deflection. UVA tests produced the greatest differences between the Hybrid III and THOR. For central sternum deflection, the UVA Hybrid III values for tests A, B, and D averaged 43% higher; test results from other labs (Fig. 17, tests E, F, and G) averaged 19% higher.

Three of the studies found either no difference between the dummies' ribcage deflection (Fig. 17, test H) or, in opposition to the findings reported here, that the Hybrid III ribcage deflected less than THOR's (Fig. 17, tests C and I). Xu et al. (2000) reported 4.6 m/s pendulum impacts to the torso that supported their sled test (test I) results indicating that the Hybrid III chest is stiffer that THOR's.

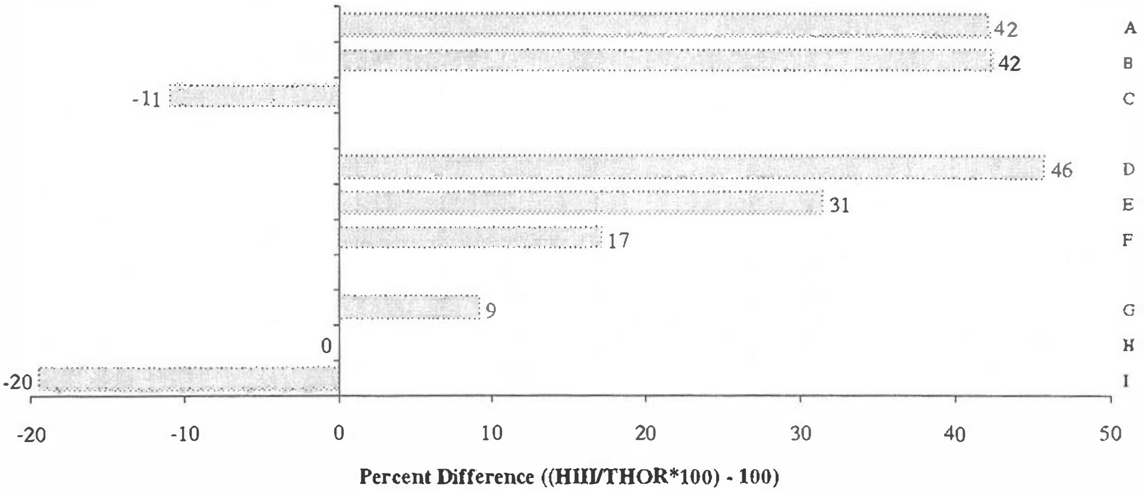


Figure 17. Percent differences between Hybrid III and THOR upper chest deflection values using the data for the center of the chest in Table 3. Refer to Table 3 for test series information.

We examined factors that may have accounted for the wide range of relative ribcage deflection values. Given that the one barrier test (I) produced the greatest relative deflections for THOR, it is reasonable to assume that the test conditions are a contributing factor. We found limited evidence that ribcage deflection results were related to the test site. The four test series reported by Rangarajan et al. (1998a,b) were conducted at three different laboratories, the Vehicle Research and Test Center (VRTC), the Volvo Safety Center, and the Autoliv Research crash facility. The values ranged from THOR deflecting 11 percent more than the Hybrid III to the Hybrid III deflecting 31 percent more than THOR (Fig. 17 tests C, H, F, E). However, in the two Rangarajan tests conducted at Autoliv, the values varied by only 11 percent (Fig. 17 tests C, H). In the three UVA test series, all of which were conducted at the Automobile Safety Laboratory, the values varied only 4 percent (Fig. 17 tests A, B, D). Potentially relevant test site differences included sled buck parameters such as steering column and knee bolster stiffness and restraint belt geometry. The crash pulse and restraint type, significant determinants of the test condition, did not appear to be related differences in deflection. The UVA results were very consistent across different sled pulses (56 km/h, 24g peak and 48 km/h, 17g peak) and restraint belt types (3pt. and force-limited 3pt.).

Differences in upper chest deflection are a function of variation in the normal loading produced by the upper torso restraint system, primarily the shoulder belt. The magnitude and location of shoulder belt loading varies with belt construction, orientation of the subject to the belt (a function of body proportions and sitting position), and/or shoulder movement.

In the UVA sled tests, THOR's pre-test shoulder position was approximately 4.5 cm higher than that of the Hybrid III. In comparison to the Hybrid III, THOR's shoulder complex is reportedly more human-like and allows increased forward movement in order to provide a more-biofidelic interaction with the shoulder belt (Rangarajan et al. 1998). Horsch et al. (1991) and Schneider et al. (1992) noted that the kinematics of the shoulder complex and its interaction with the shoulder belt have a significant effect on chest loading. The different initial shoulder position and greater forward movement of THOR's shoulder may have reduced the normal belt load on the upper chest.

Film analysis indicated THOR's left shoulder, which lay under the shoulder belt, moved approximately 5 cm farther forward with respect to the upper spine than did that of the Hybrid III at the time of peak upper chest deflection (~120 ms). Because THOR's shoulder moved further forward and was loaded more by the shoulder belt, this would have reduced the normal force on the chest and would have resulted in lower deflection values. In comparison, the Hybrid III shoulder complex moved forward much less than that of THOR. Therefore, a greater percentage of the shoulder belt load was born by the Hybrid III upper chest.

The discrepancy in THOR-Hybrid III chest response between the pendulum tests reported by Xu et al. (2000) and majority of the sled tests may be explained by differences in the way the chest is loaded. In pendulum tests, chest deflection was produced by a precisely controlled impact normal to the upper chest. Under this loading condition, the Hybrid III chest deflected less than THOR. In the sled tests, normal chest loading was a function of the complex interaction of the restraint system and the dummy. The sled test finding that the Hybrid III deflected more than THOR suggests that the pendulum test results provide incomplete information when predicting chest deflection in the crash test environment.

The results of this analysis suggest that THOR's chest deflection may be significantly affected by the interaction of the shoulder with the shoulder belt. We propose that forward shoulder movement and the resulting reduction in normal shoulder belt load may have been one of the factors that contributed to the different upper chest deflection recorded for the subjects.

THOR BIOFIDELITY: The finding that THOR responses such as head acceleration, upper spine movement, anterior chest wall movement, and lap belt loads were similar to the PMHS suggests improved biofidelity with respect to the Hybrid III. THOR responses that did not match those of the PMHS, such as peak pelvic acceleration, suggest a lack of biofidelity. However, a more comprehensive analysis of biofidelity is required that acknowledges that THOR's primary design objective is to approximate the responses of live humans. While PMHS responses provide valuable information regarding the responses of live humans subjected to injurious levels of impact loading, lack of muscle tone and other post mortem changes affect PMHS behavior in ways that are not fully understood. Estimations of live human response often involves a combination of PMHS data and data from volunteers subjected to non-injurious loading (SAE 1986). Therefore, defining biofidelity solely with respect to PMHS response may be misleading. For example, the fact that THOR recorded lower forward head movement than the PMHS may be due in part to a neck that was developed with information that included human volunteer data (Martinez et al. 1999). In this case, THOR's deviation from the PMHS response actually may represent enhanced biofidelity.

RECOMMENDATIONS

1. We recommend further study of the movement of T1 and its effect on upper spine acceleration. In comparison to the Hybrid III, THOR and PMHS recorded higher T1 z-axis acceleration that accompanied greater T1 upward movement. T1 acceleration is of special interest because it has been proposed as a component of a thoracic injury criterion (Kuppa and Eppinger 1998). T1 movement is also an important determinant of neck and head response.

2. A better understanding is required regarding THOR's relatively low upper chest deflection values. Shoulder belt loading patterns, torso construction, shoulder complex motion, and chestband installation should be examined. Better information is needed regarding the chest deflection values themselves since chest deflection is the second component of the thoracic injury criterion proposed by Kuppa and Eppinger (1998). Our investigation used chestbands, the only instrument able to measure chest deflections for both dummies and PMHS in a sled test environment. Unfortunately, absolute deflections derived from chestband data may involve inaccuracies under localized loading of the chest by a belt system (Hagedorn and Burton 1993, Shaw et al. 2000; Bass, Wang and Crandall 2000). We recommend continued development of the chestband. Initial efforts to investigate issues related to chest deflection should include quasi-static loading of the chest with an indentor and dynamic chest compression applied via a simulated shoulder belt using techniques similar to those reported by Backaitis and St-Laurent (1986). Test such as these would allow an in-depth investigation of chest deflection and chest deflection instrumentation.

3. Further investigation is needed to determine which factor or factors caused THOR's z-axis pelvic acceleration to differ from that recorded for the Hybrid III and PMHS. Pelvic response can

significantly affect acceleration results for the torso and head. The rotation of the pelvic structure and pelvic accelerometer about the y-axis should be recorded in future sled tests.

4. The conclusions regarding THOR's performance were based on the results of four PMHS sled tests using a single occupant restraint configuration. Despite mass and time scaling procedures, the PMHS results were more variable than those of the dummies. Additional testing with other restraints and test conditions is necessary before an overall assessment of THOR's biofidelity can be made.

CONCLUSIONS

The sled test results suggested that THOR was more similar to the PMHS than the Hybrid III dummy with respect to head acceleration, upper spine (T1) movement, anterior chest wall movement, and lap belt loads. THOR recorded less head forward movement and pelvic acceleration than did the PMHS. The tests did not provide the results necessary to compare dummy chest deflection values with those of the PMHS.

THOR's more PMHS-like head acceleration may be due to improved neck construction. THOR's more PMHS-like T1 movement was attributed to the flexible joint in the dummy's thoracic spine.

Substantial differences in torso construction between the two dummies resulted in THOR's anterior chest wall movement being more similar to the PMHS than was the Hybrid III. THOR's chest wall behavior under loading indicated that it was more flexible than that of the Hybrid III. THOR's upper ribcage deflected less than the Hybrid III. The difference between the Hybrid III and THOR seemed to be related to THOR's greater forward shoulder movement. Differences in shoulder belt loading patterns due to different body proportions, sitting position and/or shoulder movement, may have been contributing factors to the observed chest wall behaviors.

Head acceleration and chest wall movement results suggested that THOR and the PMHS were more sensitive than the Hybrid III to events early in the crash such as restraint system pretensioning and air bag slap. Differences in neck and torso construction are likely explanations.

We attributed THOR's lower lap belt loads to its segmented-flesh pelvic construction that may have reduced the effective pelvic mass with respect to the Hybrid III. No definitive causes were identified for THOR's low pelvic resultant and z-axis acceleration results.

For most of the responses evaluated, THOR was more similar to the PMHS than the Hybrid III. This indication of improved biofidelity should be examined relative to the responses of live humans to impact loading. Further study is required of head and upper torso movement, chest deflection, and pelvic acceleration. Carefully controlled component testing is recommended to explore the chest deflection and pelvic acceleration issues. Additional sled testing with other occupant restraints is also required.

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