Behavior of ATD, PMHS and Human Volunteer in Crash Test Ⅱ

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ABSTRACT: Quantifying the whole body response of ATD (anthropomorphic testing device) is important to study the behavior of belted occupant. In a previous study we studied the forward excursion of ATD, PMHS (Post Mortality Human Subject = Cadaver) with standard 3-point belt and Human Volunteer with energy absorbing shoulder belt, and concluded that the excursion of a human volunteer with tensed muscles in high speed frontal impact is less than the forward excursion of PMHS. In this work, we examined the behavior of a relaxed Human Volunteer with Inflatableband (an airbag equipped seat belt) in a high speed impact test which conducted in 1970`s and PMHS and ATD with Air Belt. We compared the result with previous study and also study the response of THOR dummy. Result demonstrated the forward excursion of relaxed human volunteer is smaller than PMHS, echoing the results from the previous research.

KEY WORDS: Safety, Occupant protection, Anthropomorphic Dummy, Crash Test, Safety Belt, Human Volunteer, Air Belt

1. Introduction

The whole body response of anthropomorphic test devices (ATD) is very important to evaluate the belted occupant behavior. We studied the excursion of ATD, PMHS and human volunteer in low and high speed frontal impact test in previous study using NHTSA`s previous experiment and our own test data [1]. In that study the initial position of the human volunteer was put their chin on their chest and whole body muscle tensed. We seek to expand this line of work and explore the responses of the high speed test with relaxed subjects. In the 1970s, the National Highway Traffic Safety Administration (NHTSA) and collaborators investigated the feasibility of inflatable belt restraints in the front seat of automobiles in a research program involving human volunteers and early generation dummies [2] [3]. In that study the human volunteers are requested to be relaxed and keep normal driving position. The objective of NHTSA`s research was to explore the possibility of an inflatable occupant protection system, and the difference in behavior between the relaxed human volunteer and ATD was not analyzed. To study the difference of behavior between them, we accessed the report and video of NHTSA`s human volunteer frontal crash tests, which were conducted at Southwest Research Institute, San Antonio, Texas. The team of CHOP (Children Hospital of Philadelphia), UVA (University of Virginia) and the lead author conducted research using ATD and postmortem human subjects (PMHS), both restrained with Air Belt [1], and in this paper we compare the result with NHTSA`s research. In addition, we compare the behavior of new ATD “THOR” with current ATD Hybrid III.

2. Method

Method of human volunteer tests at Southwest Research are discussed in 2.1 and our method is discussed in 2.2

2.1. Human Volunteer Test in Southwest Research

NHTSA`s tests with human volunteers were conducted at the Southwest Research Institute, San Antonio TX, in early 1970`s and final report was published in 1975.

Fig. 1 Inflatableband
System Description.
The Inflataband is an inflatable 3-point harness system that uses an argon stored gas system to inflate both the lap and shoulder belt. (Figure 1). Inflation is initiated by overpressure created by two electro pyrotechnic initiators. This overpressure ruptures a disc, allowing the gas to flow through the buckle assembly into the lap and shoulder belt portions. In the experiments, a 10-ms delay was introduced between the initiation of the sled deceleration and squib activation. The system requires an additional 10 ms from squib firing to inflate, so full inflation occurs 20 ms after sled deceleration begins. At full inflation, the belt is approximately 460 mm in circumference, but the inside pressure was not indicated. The gas is allowed to flow through the weave of the webbing portions, but the magnitude of the associated venting was not quantified.

Test Subjects.
The three subjects were selected by the original study authors from a panel of 12 volunteers recruited for the study. All volunteers were required to “pass a physical examination which included examination of heart, lungs, ENT, eyes, hearing, reflexes, muscle and joint motion and strength, pulse rate, B/P, certain anthropometric measurements, ECG after exercise, and psychologic evaluation.” Orthopedic abnormalities identified by x-ray, such as arthritis, calcified cartilage, and cortical thinning, were exclusion criteria. Table 1 describes the three subjects available for this study. They were all young adult males. The original test report describes a “rigorous program protocol” established by an institutional Committee for the Protection of Human Subjects. The makeup and purpose of this committee seem analogous to that of an institutional review board at a contemporary U.S. university.

Test Methods.
Each volunteer was tested several times, in increasingly severe exposures up to a sled change in velocity ($\Delta V$) of 52 km/h. Only the highest severity tests for the three subjects were available for analysis. These tests are described in Table 1 and in Figure 2. Test instrumentation included a triaxial accelerometer cluster mounted on the side of the head using an adjustable plastic headband. Thoracic acceleration was measured with a triaxial accelerometer array mounted, using a Velcro belt, over the right erector spinae muscle group. Shoulder belt load and bilateral lap belt loads were measured. Sled acceleration and toe pan loads were measured. The EKG and respiration of the volunteers were documented during the test. Four high-speed imagers documented occupant kinematics through the use of photo targets mounted on the wrists, elbows (lateral epicondyles), shoulders, head (temporoparietal), ankles (lateral malleoli), knees, and greater trochanters. The volunteers wore tight-fitting clothing described as “ski pajamas” with low-top tennis shoes. Ear plugs, a rubber mouthpiece, a wet suit hood, chamois for protection from fragments on the right arm, chest, and abdomen, and a foam pad over the abdomen were also used. The belt system was donned tightly, such that two fingers would fit snugly between the band and the clavicle and hip. The initial position was with the volunteer facing forward, centered in the seat, with his hands placed on his thighs just proximal to the knees with his thumbs facing medially and with his head in an upright “normal driving position”.

![Composite of 3 PMHS Tests](image1)

Figure 2. Sled deceleration pulses in human volunteer experiments with the Inflataband and in PMHS experiments with the Airbelt.

![Representative volunteer test sled pulse](image2)

Figure 3. “Airbelt” system evaluated in the experiments reported here.

| Subj | Age | Mass | Stature | Sitting height | Test | Sled $\Delta V$ |
|------|-----|------|---------|----------------|------|----------------|
| 13   | 21  | 66 kg | 179 cm  | 90.6 cm        | 976  | 52.0 km/h      |
| 24   | 24  | 80 kg | 182 cm  | 93.5 cm        | 978  | 52.0 km/h      |
| 28   | 28  | 75 kg | 177 cm  | 93.5 cm        | 979  | 51.7 km/h      |
| Avg. | 24.3| 73.7 kg| 179.3 cm| 92.5 cm       | 979  | 51.9 km/h      |

Peak sled decel = 20 g
Following the test, the volunteer was interviewed and examined to document subjective reaction and trauma. Subjective reports were completed immediately after the test and after 24 and 72 hours.

2.2. Airbelt Experiments with PMHS and ATD

System Description.
The Airbelt restraint system used in this study consisted of a three-point seatbelt with an airbag integrated into the upper portion of the shoulder belt (Figure 3). The restraint system had two retractors—one at the upper shoulder belt and one at the outboard lap belt. The upper shoulder belt retractor included a torsion-bar type force limiter, which limited the shoulder belt force to approximately 3 kN. The lap belt retractor included a pre-tensioner, which produced 2.3 kN pretensioning force (nominal). The shoulder belt and the lap belt were separate sections of webbing, each tied directly to an inboard anchor bracket upon which was mounted the Airbelt inflator. The belt-integrated airbag was conical in shape, with a maximum diameter of 160 mm. It was tapered such that the largest diameter was above or behind the shoulder, with the diameter decreasing to approximately 40 mm at mid-chest (Figure 3). This shape was chosen to increase the portion of the chest loading borne by the upper chest and shoulder (as opposed to the lower chest), to spread the load over a wider area of the upper chest, to help limit forward flexion of the neck, and to accommodate different sizes of occupants (the taper would tend to decrease the diameter of the airbag interacting with smaller occupants). The inflator was a hybrid type with a design pressure of 80 kPa. The total airbag volume was 7.2 liters. The unvented airbag was constructed of silicon-coated fabric with sealed seams and thus virtually no gas permeation. In the tests, both the airbag and the lap belt pretensioner were fired at 12 ms following the initiation of the test (t0).

Test Subjects.
The three male PMHS tested with the Airbelt are described in Table 2. Subjects were screened for HIV etc., and preexisting injury or pathologic anomalies prior to inclusion in the study. Pre-test, full-body CT scans were performed to identify pre-existing osseous pathologies. All PMHS were unembalmed and were preserved by refrigeration or freezing until the time of testing. The volunteer experiments were reported in the original test report, so the volunteers’ kinematics can also be evaluated quantitatively. The subjects’ pulmonary systems were pressurized (with compressed air) via tracheostomy to a nominal in vivo level (approximately 10 kPa measured externally) immediately prior to testing. The subjects’ cardiovascular systems were pressurized (with 6% hetastarch blood plasma replacement solution) via a cannula inserted into the left carotid artery to a nominal in vivo level (also approximately 10 kPa measured externally) immediately prior to testing. The pulmonary and cardiovascular pressures were maintained during the test using an onboard pressurization system. All PMHS test and handling procedures were approved by the University of Virginia Center for Applied Biomechanics (UVA CAB) Oversight Committee, which functions as an Institutional Review Board.

Test Methods.
Subjects were seated in the outboard, passenger’s side position of a sled buck representing the rear-seat of a 2004 mid-sized U.S. sedan. The front seat was removed to allow the capture of high-speed video. The seat cushion on which the PMHS sat was replaced after each test. PMHS positioning was performed following the procedure described by Forman et al. [5], which based the initial position on the rear-seat occupant posture study of Reed et al. [6]. Each subject was tested once at a nominal impact speed of 48 km/h. Tests were performed on a deceleration sled utilizing a hydraulic decelerator to shape the sled acceleration pulse (Figure 2). Subject instrumentation included triaxial accelerometers and angular rate sensors on the head and on the first thoracic vertebra (T1); triaxial accelerometers on the mid-spine (T7/8), lower spine (T12/L1), and pelvis; Lap belt tension was measured on the outboard side and shoulder belt force was measured using an inertially compensated load-cell mounted between the shoulder belt anchor and the rear deck. Off-board high-speed (1000 Hz) digital video was collected bilaterally and from two frontal views. Also the test with ATD conducted under same test condition as PMHS tests. The type of ATD used was the current HybridIII ATD (Test No. 1424–26) and newly developed THOR ATD (Test No. 193–95) [7].

3. Result

3.1. Inflataband Experiments
The initial position, the general occupant kinematics, and the restraint interaction can be seen in video images taken from one of the human volunteer tests (test 979) (Figure 4). Trajectories from the video collected during the 3 volunteer experiments were reported in the original test report, so the volunteers’ kinematics can also be evaluated quantitatively. The maximum forward excursion of the head ranged from 132 mm (Vol. 24) to 305 mm (Vol. 13) (Average 215mm) (Figure 5). The head followed a generally forward and downward trajectory relative to the
buck for the first 100-150 ms, dropping 58 to 168 mm. The maximum forward excursion of the shoulder ranged from 142 mm (Vol. 28) to 183 mm (Vol. 24) (Average 168 mm). Like the head, the shoulder generally displaced forward for the first 100 ms. Unlike the head, the shoulder generally continued to displace downward during rebound, reaching a maximum downward excursion of 94 mm to 120 mm around 150 ms. The maximum forward excursion of the hip ranged from 86 mm to 122 mm (Average 105 mm) and occurred around 100 ms post impact. The maximum forward excursion of the knee ranged from 89 mm to 179 mm (Average 138 mm) and occurred around 100 ms post-impact. Unlike the other body segment, the knee moved substantially upward relative to the buck, reaching a total upward
excursion of 120 mm to 200 mm before rebounding. The correlation of torso pitch angle with restraining force cannot be evaluated, unfortunately, because only peak belt loads and not time histories are reported in Burkes et al. report\[2\]. The maximum values reported by Burkes et al. \[2\] are listed in Table 3.

### 3.2. Airbelt Experiments

The initial position and the general subject kinematics and restraint interaction can be seen in video images taken from test 1428 (Figure 6). The maximum values of selected instrumentation signals of PMHS tests are listed in Table 4. Trajectories from the video collected during the 3 PMHS experiments are compared quantitatively to the Inflataband volunteer experiments in Figure 4. The maximum forward excursion of the PMHS head ranged from 347 mm (test 1428) to 521 mm (test 1429). Average was 437 mm (Figure 5), which was much greater than the excursion experienced by the volunteers with the Inflataband. The maximum forward excursion of the PMHS shoulder ranged from 330 mm (test 1428) to 409 mm (test 1429), average was 381 mm and again much greater than the volunteers with the Inflataband. The maximum forward excursion of the PMHS hip ranged from 134 mm to 179 mm. Average was 153 mm, and occurred 90-100 ms post-impact, which was approximately 20 ms earlier than head/shoulder rebound. The maximum forward knee excursion of the PMHS ranged from 150 mm to 245 mm, Average was 195 mm and was nearly coincident in time with maximum forward hip excursion. The knee motion was predominantly forward with little vertical displacement. The trajectories correspond to a whole-body response that is closer, compared to the Inflataband, to what is typically seen with a contemporary seatbelt system.

Fig. 7 shows the kinematics of Hybrid III and THOR. The behavior of both ATD looks similar, but THOR’s torso is a little flexible than

| Table 4 – Maximum Values in PMHS Tests with the Airbelt System |
|---------------------------------------------------------------|
| Subj | Test | Head \(^{+}\) Acceleration | Torso \(^{+}\) Acceleration | Lap Belt Force (Outer) | Shoulder Belt Force | Res. Angular Velocity (head) |
|------|------|--------------------------|--------------------------|---------------------|------------------|-----------------------------|
| 481  | 1427 | 32.2 g                   | 27.8 g                   | 6.13 kN             | 3125 N           | 25.6 rad/s                  |
| 461  | 1428 | 32.9 g                   | 34.9 g                   | 4.45 kN             | 3178 N           | 20.4 rad/s                  |
| 482  | 1429 | 29.3 g                   | data failure             | 3.79 kN             | 3428 N           | 15.8 rad/s                  |
| Avg. |      | 31.5 g                   | 31.4 g                   | 4.79 kN             | 3244 N           | 20.6 rad/s                  |

Resultant Acceleration, CFU 1000.
the torso of Hybrid Ⅲ, resulting in more of an arched spine, similar to the PMHS. It is difficult to compare the THOR with the human volunteer, because the difference of torso angle is too large, due to the existence of force limiter in the THOR tests.

4. Discussion

The combination of the PMHS experiments reported here and the classic human volunteer experiments of NHTSA provide an interesting opportunity to compare the behavior of PMHS and Human volunteers. The difference in test subjects (young, living males vs. older PMHS) and the test condition made it difficult for direct comparison, but overall comparison of forward excursion is possible.

The forward excursion The difference in frontal excursion between human volunteer and PMHS was 222 mm at head, 213mm at shoulder, 48mm at hip and 57mm at knee. Similar to our previous study, PMHS’s excursion is greater in all body segments. But before we make these findings need to be considered in light of the differences between volunteer tests at SWRI and the PMHS and ATD tests at UVA.

1) While the shape of the deceleration pulse was similar, there is a difference in test speed (51.9 km/h vs 48.3 km/h) (see Fig.2).
2) PMHS test used the Air Belt with Force limiter, but the Torso Angle, degrees Submerging Observed

Table 5: Maximum Forward Displacements*

| Test # | 1263 | 1262 | 1264 | Std. Dev. | 1386 | 1387 | 1389 | Std. Dev. |
|--------|------|------|------|-----------|------|------|------|-----------|
| Head X | 50   | 48   | 53   | 51.3      | 3.5  | 56   | 60   | 57        | 2.3       |
| Shoulder X | 32   | 27   | 29   | 28        | 1.4  | 45   | 47   | 39        | 4.1       |
| Hip X  | 38   | 19   | 22   | 21        | 2.1  | 30   | 15   | 18        | 4.3       |
| Knee X | 30   | 22   | 21   | 22        | 0.7  | 30   | 15   | 18        | 8.1       |
| Torso Angle, degrees Submerging Observed | -19 | -18  | -21  | -20       | 2.1  | 0.8  | 27   | 29        | 18        | 15       |

* All displacements are relative to the initial positions. The torso angles are the maximum absolute torso angles. All displacements are scaled, and are in cm.

We studied the increase of forward excursion by impact speed in previous study[1], and it can be estimated that there is a 17mm decrease at human volunteer shoulder excursion for 3.6km/h impact speed decrease. (Fig.8).

The influence of belt force limiter to the amount of forward excursion must be examined. UVA also conducted tests with PMHS with a standard seat belt and a seat belt with force limiter and pre-tensioner at 48km/h. [8]. Table 5 compares the result of both conditions. The force limiter increased the excursion about 160mm even the presence of a pre tensioner. Also Takata conducted preparation tests before PMHS tests, which included seat belts with force limiter and without force limiter using the Hybrid Ⅲ dummy (Table 6, Shot 1 and shot 2 ). The forward excursion of shoulder in this tests was 227mm without the force limiter and 293mm with the force limiter. The difference was 66mm for ATD (Hybrid Ⅲ). From these two result, 66mm with ATD and 160mm with PMHS, we can estimate the amount of force limiter influence. Together, the total of the influence of the difference of test speed and the existence of force limiter on the forward excursion is still smaller than the forward excursion difference between human volunteer and PMHS. Therefore, we
remain confident in the conclusion that the human volunteer has reduced forward excursion compared to the PMHS and we attribute this difference to the response of muscle. Even when volunteers are requested to be relaxed as was the case in these tests, their muscles react at the moment of impact. The lower limb muscle tensing transmits load to toe pan, supports the mass and acceleration of the hip, and reduces forward excursion. In the volunteer experiments the toe pan loads were 7.1, 7.7, and 5.6 kN for the three volunteers. Thus, it seems reasonable to conclude that active muscle tensing in the volunteers created a situation that the forward excursion of volunteer was generally smaller than PMHS.

Add to the forward excursion consideration we can see the difference of neck flexion of human volunteer, PMHS and ATD.

The Neck flexion difference between human volunteer and PMHS

Fig.8 shows the maximum neck flexion of human volunteer, PMHS and ATD. The amount of human volunteer’s neck flexion is smaller than PMHS even when considering the difference in torso lean. The neck flexion of the ATD looks larger than volunteer and PMHS. In a previous study we examined the maximum neck flexion of ATD was smaller than PMHS. (Fig.9) The neck of PMHS demonstrates substantial flexion with the forehead almost touching the thigh, but Hybrid III’s neck only bend slightly more than horizontal. The neck of THOR bend more than Hybrid III but still smaller than PMHS. Probably the reason of this different result is the interference of Air Belt and face of subject. In Fig.8 the neck of human volunteer bend to side (left), and the torso of PMHS leans to left. This movement to the side is not seen for the ATD. This movement to side made the contact of the face more perpendicular to the Air Belt. So the inflated Air Belt prevent more flexion of the neck of human volunteer and PMHS. In contrast, with a standard seat belt the neck flexion is dictated by the neck performance and torso flexibility of subjects. THOR simulated the behavior of human volunteer and PMHS better than Hybrid III, but the movement to the side need to be refined.

5. Conclusion

We had concluded in previous study that the forward excursion of a human volunteer with tensed muscles in high speed frontal crash
loading condition is less than the forward excursion of PMHS. In this paper, we examined the difference of forward excursion between PMHS and ATD restrained by an Air Belt and human volunteer who relaxed before impact restrained by Inflataband. Even in the context of variation in test speed and the existence of belt force limiter, the forward excursion of human volunteer was smaller than that of PMHS. This result is similar to our previous study where we evaluated human volunteers with tensed muscles, restrained by standard belt with energy absorbing webbing. Thus providing evidence of the generalizability of this conclusion.

We tested the new THOR dummy in similar condition and compared it to current Hybrid III dummy. We found that the THOR dummy simulated human volunteers and PMHS better than Hybrid III, but it still needs refinement in its off-axis movement to the side.

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