PMHS Restraint and Support Surface Forces in Simulated Frontal Crashes

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ABSTRACT: This paper documents restraint and support surface interaction recorded for eight 50th percentile post mortem human surrogates (PMHS) in 40 km/h, 14 g frontal sled tests. Three-point restraint belt loads and inertially compensated loads and moments for rigid seat, knee bolster, and footrest are reported. This information will facilitate computational modeling of PMHS response and assist in efforts to evaluate the biofidelity of frontal crash dummies.

KEY WORDS: human engineering, biomechanics, occupant safety

1. INTRODUCTION

Biomechanical response corridors are often used in the development of anthropomorphic test devices (ATDs) and in the validation and development of finite element (FE) models[1-3]. The corridors based PMHS responses provide a metric for assessing the current biofidelity of the ATD or FE models and also provide targets for further development and refinement.

In order to successfully evaluate ATDs and FE models a defined and repeatable testing environment is needed. A testing environment that has defined boundary conditions removes any variability that would otherwise be introduced by the interaction of the PMHS and any compliant surfaces found in the testing environment. In frontal crash simulation these surfaces can include the seat and knee bolster. In order to minimize the influence of the compliant vehicle interior the seat was approximated by a rigid, planar surface and the knee bolster was made of a stiff aluminum channel. These design elements of the testing environment allowed for an accurately defined PMHS-support surface interaction (see Fig. 1) and aided in the overall goal of providing data that can be used in further ATD and FE model development[4].

This test series was composed of eight, approximately 50th percentile male PMHS tested in a 40 km/h frontal sled test with a 14 g impact acceleration restrained with a 3-point belt. The testing configuration was developed to evaluate and assess the biofidelity of the thoracic response of restrained ATDs in a frontal crash. The goal of the test series was to measure the kinetics and kinematics of a restrained occupant in a test condition that approximates a frontal crash. While some of the results of this test series have been reported previously in the form of skeletal kinematics[4] and chest deformation[5] the goal of this study is to document and present restraint and support surface interactions including the upper and lower shoulder belt tensions, lap belt tensions, seat loads, left and right knee bolster loads, and footrest loads for use in ATD and FE model response assessment and validation.

2. METHOD

Eight PMHS tests were conducted in a 40 km/h frontal impact environment. The condition was selected to represent a repeatable pure frontal collision in the severity range of typically conducted regulatory and assessment frontal impact tests. The PMHS were chosen to approximate the height and mass of a 50th percentile male. The eight tested PMHS had an average height of 179.3 cm and an average mass of 75.5 kg. Additional specimen information is shown in Table 1.

Table 1 PMHS Specimen Information

<table>
<thead>
<tr>
<th>Test #</th>
<th>1294</th>
<th>1295</th>
<th>1358</th>
<th>1359</th>
<th>1360</th>
<th>1378</th>
<th>1379</th>
<th>1380</th>
<th>Ave.</th>
<th>St. Dev.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
<td>76</td>
<td>47</td>
<td>54</td>
<td>49</td>
<td>57</td>
<td>72</td>
<td>40</td>
<td>37</td>
<td>54</td>
<td>4.9</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>70</td>
<td>68</td>
<td>79</td>
<td>76</td>
<td>64</td>
<td>81</td>
<td>88</td>
<td>78</td>
<td>75.5</td>
<td>2.8</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>178</td>
<td>177</td>
<td>177</td>
<td>184</td>
<td>175</td>
<td>184</td>
<td>179</td>
<td>180</td>
<td>179.3</td>
<td>1.16</td>
</tr>
<tr>
<td>Cause of Death</td>
<td>Pancreatic Cancer</td>
<td>Coronary Artery Disease</td>
<td>CVA and Atrial Fibrillation</td>
<td>Lung Cancer</td>
<td>Neoplasm of Brain</td>
<td>Cancer</td>
<td>Cardiovascular Disease</td>
<td>Seizure Disorder</td>
<td></td>
<td></td>
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</table>
The procurement and use of the PMHS was done in accordance with the ethical guidelines established by University of Virginia Oversight Committee. The PMHS were preserved through freezing prior to testing. The PMHS were also tested to ensure that they were free of HIV and Hepatitis B and C.

The test fixture was developed as a reasonable analog to a front passenger seating location and 3-point shoulder and lap belt anchoring position of a mid-size U.S. sedan. The belt anchoring positions are reported relative to the outer lap belt anchor location in Table 2 (in the coordinate system shown in Fig. 1).

Table 2 Coordinates of belt attachment locations (measurements reported in centimeters)

<table>
<thead>
<tr>
<th>Location</th>
<th>X</th>
<th>Y</th>
<th>Z</th>
</tr>
</thead>
<tbody>
<tr>
<td>Outer lap belt anchor location</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Inboard lap/shoulder belt anchor</td>
<td>13.3</td>
<td>-54.6</td>
<td>-7.6</td>
</tr>
<tr>
<td>D-Ring anchor point</td>
<td>-21.0</td>
<td>3.2</td>
<td>-99.7</td>
</tr>
</tbody>
</table>

The test fixture was also designed with the goal of providing a repeatable and reproducible testing environment. The repeatability of the test fixture was achieved through the use of a rigid planar seat and an adjustable matrix of cables used to support the torso and head. The lap and shoulder belts were joined near the left hip in a location that approximates the position of a stalk-mounted buckle. The shoulder and lap belts were constructed of 48 mm-wide restraint webbing (6-8% elongation, 6000 lbf (26,689 N) minimum tensile strength). The belts were replaced after each test but the belt adjustment hardware was reused. A stiff knee bolster was positioned to be in contact with the lower shank of the leg at the location of proximal tibias at the time of impact. The positioning of the knee bolster and an aluminum footrest with ankle straps were used to restrict the movement of the pelvis and lower extremity. The snug lap belt, the stiff (aluminum) channeled knee bolster, and stiff footrest were designed to restrict the motion of the pelvis and lower extremity while allowing the torso to pitch forward in order to mimic the kinematics of real-world crashes.

Instrumentation used in these tests and described in this paper include 6-axis load cells at the seat, right knee bolster, and footrest. A 5-axis load cell was used in the left knee bolster location (moments about the X axis (MX) were not measured). The tensions in the upper and lower shoulder belt as well as the lap belt were also measured. The upper shoulder belt tension was measured near the D-ring while the lower shoulder belt tension was measured near the joining location of the lap and shoulder belt. The lap belt tension was measured near the right anchoring location. An accelerometer was fixed to the sled to measure the acceleration of the sled during the impact.

2.1 Data Processing and Presentation

Instrument data for these tests were collected at 10,000 Hz with an onboard data collection system, TDAS (Diversified Technical Systems Inc.). The data were hardware-filtered to 3000 Hz, debiased, and then subjected to post-processing filtering as prescribed in the SAE J211(6) filtering standard of channel frequency class (CFC) 60 for the belt loads and CFC 60 for the support surface load cell forces and moments. The instrument data are reported in accordance with the coordinate systems specified by SAE (Fig. 1).
The forces and moments reported in this study and measured at the seat, knee bolster and footrest load cells are the forces and moments acting on the PMHS at interaction surface between the PMHS and the test hardware. The data were then truncated to -10 to 250 ms for presentation. For a given parameter the response data were averaged for each point in time and then a ±1 standard deviation corridor was generated around the average response curve. After the data was collected the effect of the acceleration of the mass attached to the load cells was removed through the process of inertial compensation. This was accomplished by conducting a test with no subject in order to record the inertially generated forces and moments acting on the affected load cell coordinate system axes (mass compensation for forces: X-axis for the seat load cell and X and Y-axes for the left and right knee bolster load cells and the footrest load cell; mass compensation for moments: Y-axis for the seat, left and right knee bolster and footrest load cells). These measured forces and moments with no occupant were then subtracted from the tests conducted with the PMHS to remove the forces and moments generated by the acceleration of the testing hardware.

Additional information regarding the setup and pre-test subject positioning is available in the previous presentation of the results of this test series in Shaw et al[4-5].

The variability in the data was investigated using the coefficient of variation (COV):

\[
COV = \frac{St. Dev.}{Avg.} \times 100
\]

where the St. Dev. is the standard deviation of the peaks in the signals and Avg. is the average of the peaks.

3. RESULTS

The acceleration of the sled during the impact was observed to be qualitatively repeatable while closely approximating the 14 g target value (Fig. 2).

The upper and lower shoulder belt tensions also exhibited repeatability with COV values less than 15% (Table 3). The lap belt tension was observed to be slightly less repeatable with a COV value of 26% (Table 3)

Table 3 Peak sled acceleration and belt loads.

<table>
<thead>
<tr>
<th>Belts</th>
<th>Sled Acceleration $\max$ Avg 14.7</th>
<th>Upper Shoulder $\max$ 6753</th>
<th>Lower Shoulder $\max$ 5596</th>
<th>Lap $\max$ 861</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>St. Dev 0.4</td>
<td>940</td>
<td>570</td>
<td>226</td>
</tr>
<tr>
<td></td>
<td>COV 3%</td>
<td>14%</td>
<td>10%</td>
<td>26%</td>
</tr>
</tbody>
</table>

Fig. 2. Buck acceleration with the average value shown in the solid line with a ±1 standard deviation bound show in the dashed lines. The individual data traces from each subject are reported in light gray.

The upper and lower shoulder belt have a similar behavior resulting in a narrower ±1 standard deviation corridor than the lap belt (Fig. 3). The behavior of the lap belt signal is less uniform causing the larger corridor.

Fig. 3. Belt tension forces (upper and lower shoulder and lap belt) with the average force shown in the solid line with a ±1 standard deviation bound show in the dashed lines. The individual data traces from each subject are reported in light gray.
When looking at the XZ plane of the impact event, the primary loading plane of the seat and footrest load cells, the loading is more uniform between tests and moves out of the vibrational and noise oscillation levels that are observed when examining the loading that occurs in the lateral (Y) axis (Fig. 4). There are larger Y-axis loads (Fig. 5, FY) that are observed in the left and right knee bolsters due to the interaction between the lower shank of the legs and the aluminum channel that is restraining the lower extremities.

The tests were found to be repeatable with COV values of less than 20% for the majority of the primary loading directions and in the predominant polarity (Table 4). The average value reported is the average of the peaks of the individual data traces along with the standard deviation of those peak values.

The minimum forces in the X and Z directions of the seat and foot rest load cells were less variable with COV values of less than or equal to 21%. This is similar to what was observed in Fig. 4 and Fig. 5 where the sensor values are more uniform in the XZ plane than the values are along the lateral axis (Y-axis). The minimum forces in the X and Y directions were more repeatable in the knee bolsters with COV values of 20% or less.

Fig. 4. Force from the seat and foot rest load cell with the signal average in a solid line and ± 1 standard deviation in the dashed line. The individual data traces from each subject are reported in light gray.
4. DISCUSSION

The test hardware and testing procedures utilized in this test series were able to produce repeatable upper and lower shoulder belt loads and forces in the primary loading directions of seat, knee bolster and footrest load cells. The tests showed good repeatability within the often highly variable dataset of PMHS testing. Similar or higher levels of variability are seen in test fixture/PMHS interaction with approximate COV values of 10-45% (7). These reported COV values are considered good within the PMHS testing environment though they are higher than values that are considered "acceptable," 10%, or even "good," 5%, in ATD repeatability requirements (8).

5. SUMMARY AND CONCLUSIONS

This study evaluated the restraint loads generated during a 40 km/h, 14 g simulated frontal impact with eight approximately 50th percentile male PMHS. The reported restraint forces and moments were from belt tension gauges placed on the shoulder and lap belt along with load cells at the seat, the knee bolster and footrest. The results were reported in graphical format with the average of all the recorded data traces and a corridor established by plus or minus one standard deviation from the average at each time value recorded in the test. The resulting corridors are valuable in assessing the performance of ATDs and FE models in the tested condition presented in this study.
The use of the rigid test fixture lead to a uniform and defined set of boundary conditions yielding repeatability that was reasonable for PMHS, especially for main axis loading.

6. ACKNOWLEDGEMENTS

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REFERENCES