5TH % FEMALE DUMMY UPPER EXTREMITY INTERACTION WITH A DEPLOYING SIDE AIR BAG

Stefan M. Duma
Jeff R. Crandall
Walter D. Pilkey
Automobile Safety Laboratory
University of Virginia

Kazuhiro Seki
Takashi Aoki
Honda R&D Co., Ltd., Japan
Paper Number 98-S5-W-20

ABSTRACT

This paper presents the results from experiments designed to characterize the upper extremity response of the small female during side air bag loading. A seat mounted thoracic side air bag was deployed statically using three different inflators. The aggressivity of the inflators varied in peak pressure and pressure onset rate. The 5^{th}% female HIII dummy was utilized in three positions which were chosen to maximize loading of the humerus and elbow joint. Two had the dummy positioned outboard with the forearm on the armrest, and the third had the dummy inboard such that the humerus was positioned horizontally in front of the air bag module with the forearm supported above the armrest. Instrumentation for the 5^{th}% female dummy included the fully instrumented SAE upper extremity with six axis load cells in the humerus and forearm as well as accelerometers and angular rate sensors attached to each segment. All inflators produced resultant humerus moments below published injury tolerance values for the small female with the more aggressive air bags producing higher responses. The upper extremity response was correlated to inflator peak pressure and pressure onset rate.

INTRODUCTION

Although driver side air bags have reduced the risk of fatal injuries in automobile collisions, they have increased the incidence of nonfatal injuries including upper extremity injuries. It is suggested that there may be a 40% increase in risk of serious (AIS 3) upper extremity injury to belted occupants with air bags versus those without air bags [NHTSA, 1996]. Kuppa (1997) showed that 1.1% of drivers who were restrained by only a seat belt experienced an upper extremity injury, versus 4.4% of drivers in the presence of a deploying air bag who experienced an upper extremity injury. Although air bag depowering is expected to have a beneficial effect on the rate of upper extremity injuries from driver side air bags, it is unclear whether or not the implementation of side air bags will provide a new upper extremity injury mechanism. Since side air bags have been installed in only a few cars, it currently is not possible to evaluate the upper extremity injury potential through the typical real world crash investigation techniques. Thus, experiments with instrumented dummies and cadavers are performed to better understand this interaction. The goal of this paper is to evaluate the interaction of the small female upper extremity with a deploying side air bag.

The interaction between a side air bag and the average male upper extremity was evaluated by Kallieris (1997) using both the HIII 50^{th} male dummy and five male cadavers. A seat mounted combination thorax-head bag was used with the upper extremity positioned in contact with the seat seam. Only one humerus fracture was recorded for all five tests. Thus, it was suggested that there is a low risk of upper extremity injury during side air bag deployment. In addition, the kinematic differences between the dummy and cadaver were significantly different given the poor biofidelity of the dummy shoulder joint.

HUMERUS INJURY REFERENCE VALUES

Before evaluating the dummy tests, it is useful to establish reference injury assessment reference values (IARV) for the humerus. Several studies have addressed the humerus bending strength and the results are presented in Table 1. The studies by Weber (1859) and Messerer (1880) are dated and involve populations that are likely different than the modern population. The 5^{th}% female injury criteria is best established by tests with female humeri. The research by Duma (1998) is the only study to use an appreciable number of female humeri (n = 12), moreover this is the only study to test the humeri dynamically with strain rates between 1 and 3 strain/second. Thus, the injury tolerance for the 5^{th}% female will be selected as the scaled value of 128 Nm. Since all but one of the humeri tested (n = 19) by Kirkish (1996) were male, the injury tolerance for the 50^{th}% male could be taken as the scaled value of 230 Nm.

Table 1: Published Humerus Tolerance Data

Author	Year	Male Bending Failure (Nm)	Female Bending Failure (Nm)
Weber	1859	115	73
Messerer	1880	151	85
Kirkish	1996	157 ± 41	84
	scaled	230 ± 65 (50^{th}%)	134 ± 38 (5^{th}%)
Kallieris	1997	137 ± 7	
Duma	1998		154 ± 24
	scaled	217 ± 32 (50^{th}%)	128 ± 19 (5^{th}%)

METHODOLOGY

Three types of seat mounted, thoracic side air bags were used that varied only in their level of inflator output. Table 2 outlines the relative differences in peak pressure and pressure onset rate between the three inflators as measured in a 1 ft^3 tank test. The inflators utilize hybrid technology and the bags have two vents on the outboard side.

Table 2: Side Air Bag Inflator Characteristics

Inflator Type	Increase in Peak Pressure Relative to Type 1B	Increase in Pressure Onset Rate Relative to Type 1B
1B	0%	0%
2A	23%	63%
3C	54%	160%

Dummy instrumentation included triaxial head and chest c.g. accelerometers. MHD angular rate sensors were attached to the head and upper spine to track body rotation throughout the event. Upper and lower chestbands were used to measure any possible chest deformation. All tests were captured with high speed color video (1000 fps) and high speed color film (3000 fps).

The 5^{th}% female dummy was also equipped with the SAE 5^{th}% female instrumented upper extremity. This device has been shown to be effective at characterizing the upper extremity response due under air bag loading [Bass 1998]. Although the arm was used to develop the injury criteria for the forearm under loading from a driver side air bag, this is the first published study to use it for the analysis of side air bags [Bass 1997]. The arm with the appropriate polarity is shown in Figure 1. The forearm is a single shaft incorporating a six-axis load cell located approximately mid-shaft. The elbow is a single degree of freedom joint which allows flexion/extension but not the pronation/supination rotation as in the human upper extremity. This loss of motion is not significant given the application and symmetry of the shaft. On the proximal side of the elbow joint, two strain gauges measure bending moments along the X and Y axis. The humerus is similar to the forearm with a single shaft design which includes a six-axis load cell. The shoulder joint allows for the three principle rotations of the shoulder, but the lack of clavicle/scapula movement accounts for the poor biofidelity of the dummy shoulder joint. Additional accelerometers and MHD angular rate sensors were added to the forearm and humerus.

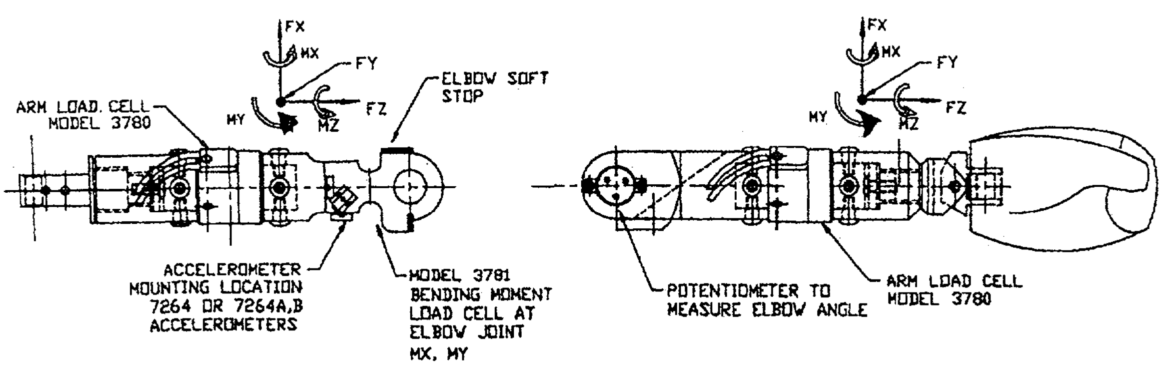

Figure 1: SAE 5^{th}% Female Instrument Upper Extremity

Initial Positioning

A new computer model for studying side air bag and occupant interaction was developed using the CVS/ATB multi-body dynamics program [Sieveka 1998]. The model employed standard ellipsoids to represent the type 2A side air bag. Using this model, several initial positions were developed that maximized loading to the humerus, elbow joint, and thorax, with emphasis on the humerus response. The instrumented 5^{th}% female HIII with the SAE instrumented arm were then used to evaluate the positions recommended by the simulation study. Figure 2 shows the peak resultant humerus moment for this first round of dummy tests. Since the primary goal was to maximize loading of the humerus, positions 11B, 5, and 8B were selected as the final three positions to be tested with each of the three side air bags.

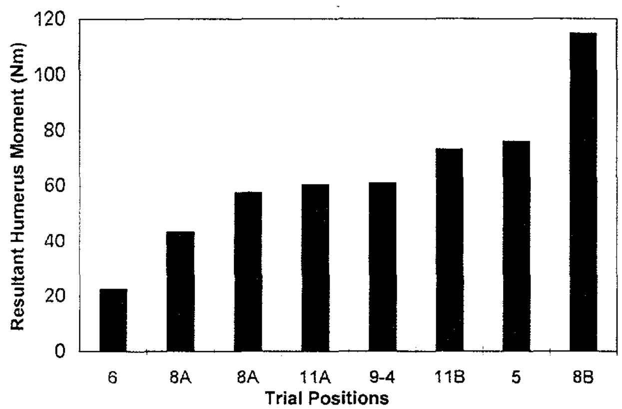

Figure 2: Peak Humerus Resultant (MX + MY) Moment for Dummy Trial Positions

The frontal and lateral views of the three worst case positions are displayed in Figure 3. The humerus, shoulder, and elbow are loaded in position 8B which has the humerus in contact with the seat back and the forearm resting on the armrest. In position 11B the humerus is placed across the path of the side air bag and the forearm is raised from the armrest. The dummy is moved slightly inboard to allow for proper upper extremity placement. This position is designed to maximize the load on the humerus and the elbow joint. Finally, in position 5 the dummy is placed completely outboard with the forearm on the armrest such that the air bag loads the humerus and posterior thorax.

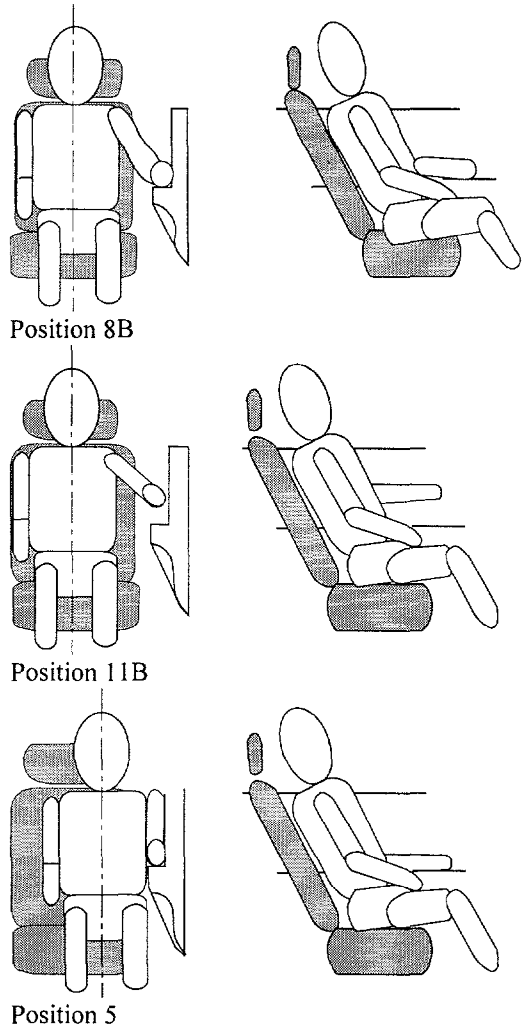

Figure 3: Three 'Worst Case' Positions

Preliminary tests suggested that an positioning accuracy of ± 2 mm along any axis was the tolerance needed to repeat the tests. Given the long and thin side air bag, any error in positioning would result in the air bag deploying along the path of least resistance. Thus, if the humerus is out of position by more than 2 mm in any direction, the air bag may not completely load the humerus. For this reason a FARO® arm was used for final positioning to remain within the allowed tolerances.

RESULTS AND DISCUSSION

The upper extremity response can be characterized using the kinematic and kinetic sensor data. Neck and torso rotations were insignificant for all tests given the stiffness of the dummy and the lack of direct loading to these body regions and therefore are not presented.

Kinematics

Although the SAE 5th% female arm contains a potentiometer at the elbow joint to measure forearm flexion/extension, there are no other angular motion sensors. For this reason an array of MHD angular rate sensors was installed on the dummy that allowed for the measurement of all upper extremity rotations. To compare the accuracy of the MHD sensors, the elbow flexion angle is plotted using both the potentiometer data and the MHD data as shown in Figure 4. Since both traces are nearly identical and the potentiometer wires failed in two of the nine tests, the MHD data will be used to discuss all rotations.

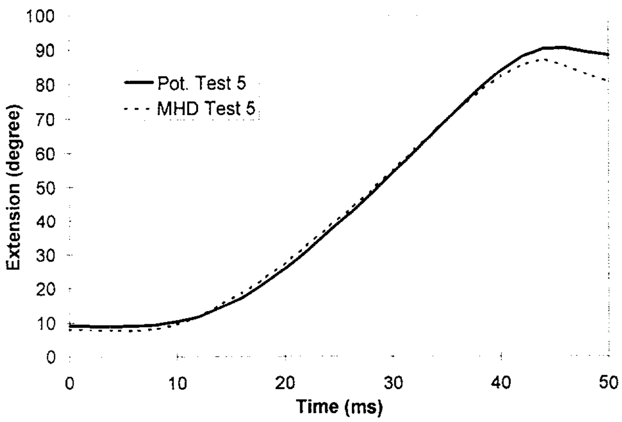

Figure 4: Potentiometer and MHD Data for the Elbow Flexion Angle in Test 5

The dummy shoulder joint consists of three rotations that are defined for this discussion as follows. Movement of the humerus in the sagital plane is defined as flexion when the humerus is moved forward and extension when the humerus is moved rearward. Rotation of the humerus in the frontal plane is defined as adduction toward the body and abduction away from the body. The neutral position for both flexion and adduction is with the humerus vertical and the distal end pointing down. The third rotation is called medial/lateral rotation. The neutral position has the humerus vertical and the elbow bent 90 degrees and pointing forward. Rotating the hand towards the body defines medial rotation, while rotating the hand away from the body is defined as lateral rotation.

A difference between the three positions can be seen by the amount of adduction at the shoulder joint. Figure 5 details this for all three positions with air bag 2A. The negative adduction angle is the same as a positive abduction angle. So, for positions 8B and 11B the humerus is initially abducted 37 and 64 degrees respectively. Despite the initial positioning difference, both position 8B and 11B rotate in a similar manner as the humerus rotates towards the body as shown by the increasing adduction angle. This motion corresponds to the air bag loading the posterior and lateral side of the humerus. However, in position 5 the air bag deploys between the thorax and humerus and forces the humerus against the door which results in the slight decrease in adduction angle.

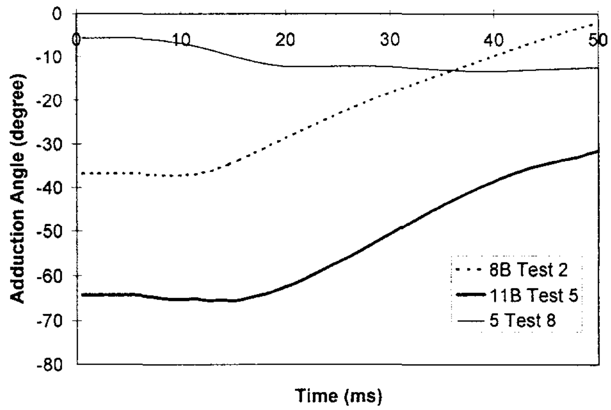

Figure 5: Shoulder Adduction for Three Positions with Air Bag 2A

A second kinematic difference between the positions is seen in the medial/lateral shoulder rotation. In Figure 6 the lack of shoulder rotation for position 5 is again seen as the medial rotation remains below 10 degrees for the duration of the test. The medial rotation for position 8B is nearly opposite that for position 11B. In position 8B the elbow is forced down which results in the negative medial rotation, whereas in position 11B the air bag forces the elbow slightly upward and therefore induces a positive medial rotation. Despite the separate direction, the magnitudes of the rotation for both positions 8B and 11B are similar at 29 and 35 degrees respectively.

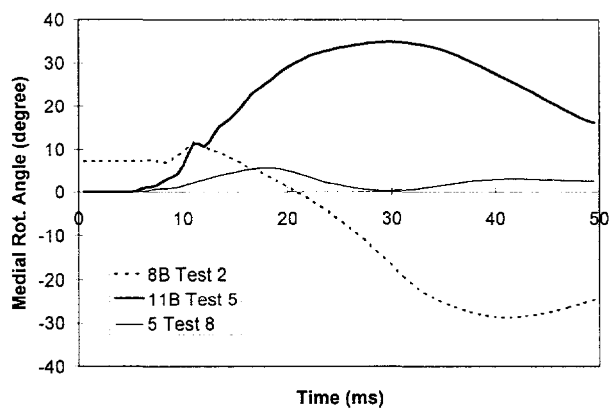

Figure 6: Shoulder Medial/Lateral Rotation for Three Positions with Air Bag 2A

While the air bag type had little influence on the shoulder rotation, it did affect the elbow flexion response. In position 11B the magnitude and rate of elbow flexion

Figure 7. The most notable difference is in bag 1B which produces approximately one half the elbow flexion response versus bags 2A and 3C.

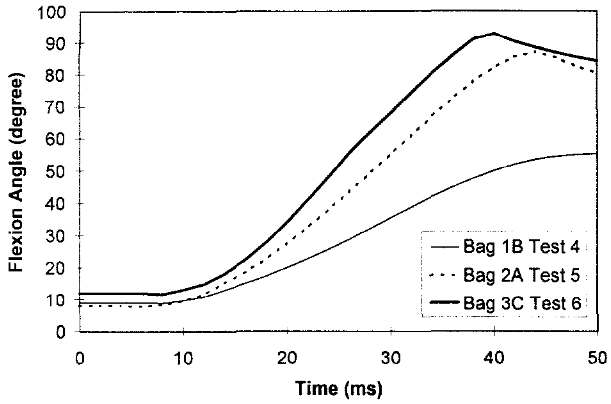

Figure 7: Elbow Flexion for All Air Bags in Position 11B

As seen with the shoulder rotations, elbow flexion is quite different among positions as Figure 8 details for all positions with air bag 2A. Again, position 5 reveals negligible upper extremity motion, while positions 8B and 11B follow similar rotations. When the initial flexion angle is considered for positions 8B and 11B, it can be seen that the air bag forces full elbow extension in position 11B versus a slightly flexed elbow for position 8B.

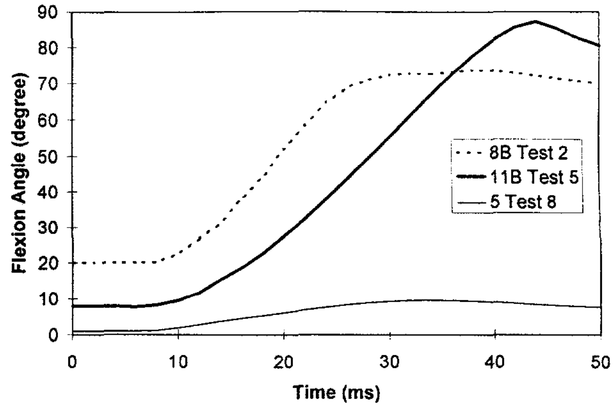

Figure 8: Elbow Flexion for Three Positions with Air Bag 2A

Kinetics

A summary of the upper extremity response for each test is presented in Table 3. A linear regression analysis was performed to identify any correlation between the peak sensor readings and inflator peak pressure and pressure onset rate.

Table 3: Peak Resultant Accelerations, Forces, and Moments for the Upper Extremity

Test	Position	Bag Type	Forearm Accel. (g)	Time (ms)	Humerus Accel. (g)	Time (ms)	Humerus Resultant FX + FY (N)	Time (ms)	Humerus Resultant MX + MY (Nm)	Time (ms)
1	8B	1B	170	9.2	207	8.7	720	8.4	59	9.0
2	8B	2A	207	7.5	220	10.0	1249	10.4	103	11.2
3	8B	3C	233	11.4	392	10.2	815	7.0	96	7.6
4	11B	1B	186	8.5	202	12.9	645	12.9	36	8.2
5	11B	2A	187	6.4	388	11.7	662	16.9	62	9.5
6	11B	3C	378	7.4	410	11.0	617	6.7	89	7.2
7	5	1B	102	9.4	187	14.1	550	14.7	73	17.2
8	5	2A	265	23.5	151	7.4	615	6.8	76	15.3
9	5	3C	343	22.2	213	7.8	1029	11.5	106	14.0

The peak forearm accelerations ranged from a low of 102 g to a maximum of 378 g for all air bags and all positions, and increased with increasing inflator aggressivity. The forearm accelerations for position 8B showed the best correlation to inflator properties with R^2 values of 0.96 for the peak pressure and 0.95 for pressure onset rate. No correlation was seen with the forearm accelerations in positions 11B or 5. In tests 8 and 9 the peak accelerations occur much later than the peaks for the other tests as a result of the different loading pattern. In position 5 the peak is affected by the upper extremity interaction with the door.

The forearm sheer forces FX and FY were insignificant given the type of air bag loading; however, the axial compression load FZ of the forearm demonstrates the early timing of the peak loads and air bag dependence as shown in Figure 9.

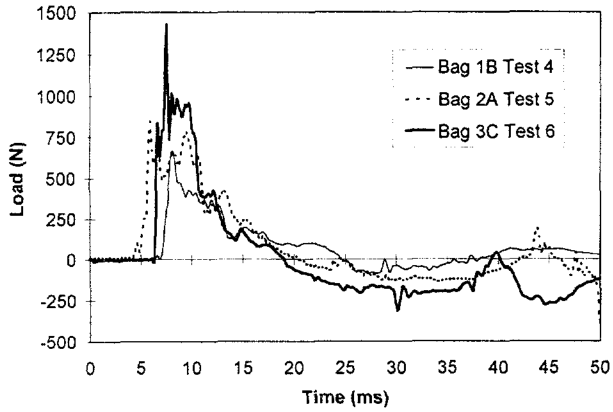

Figure 9: Axial Forearm Load (FZ) for All Air Bags in Position 11B

The humerus accelerations increased with increasing inflator aggressivity. In position 5 the humerus acceleration correlated reasonably well with the peak pressure and pressure onset rate with R^2 values of 0.92 and 0.90 respectively. The correlation was less with position 8B which gave R^2 values of 0.86 and 0.89 for peak pressure and pressure onset rate, while no correlation was seen with position 11B. The forearm and humerus accelerations are similar to those recorded by Kallieris (1997). Although Kallieris used a much larger side air bag, he also used average male cadavers so that the larger mass counteracted the larger bag to produce upper and lower humerus accelerations ranging from 193 g to 334 g.

The peak resultant FX and FY humerus forces ranged between 550 N and 1249 N with all peaks occurring before 14.7 ms as the air bag initially loaded the posterior humerus. The tests in position 8B resulted in a higher load of 1249 N with air bag 2A versus the higher aggressivity bag type 3C which gave a 815 N peak response. This trend was seen in position 11B where the resultant humerus sheer force was lower for bag 3C than for the 1B or 2A. Only position 5 gave humerus loads that correlated reasonably well with peak pressure and pressure onset rate with R^2 values of 0.90 and 0.92 respectively.

Dummy joint stops often dramatically increase the response due to high inertial accelerations when a particular joint reaches its limit. This is the case for the elbow joint in the SAE 5^{th}% female arm. Position 11B best illustrates this behavior as it is the position that induces the most extension in the elbow joint as seen previously in Figure 8. Using the strain gauge located at the distal humerus shaft, the bending moment MY for each air bag in position 11B is shown in Figure 10. Between 5 ms and 15 ms the response is due to the air bag contact, but the large peaks at 40 ms for bag 3C and 45 ms for bag 2A are a result of the elbow joint completely extending and reaching the joint stop. Due to the lack of known human elbow joint properties it is impossible currently to determine if this response is appropriate or not. Also, it is interesting to note that bag 1B does not impart enough energy to the upper extremity to cause the humerus and forearm velocities needed to see the peak at the joint stop.

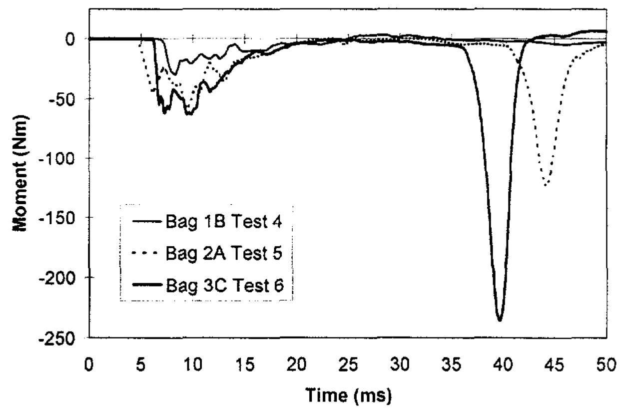

Figure 10: Elbow Moment MY for All Air Bags in Position 11B

The resultant humerus moments (MX, MY) presented in Table 3 were taken only during the first 30 ms given the uncertainty of the high values occurring after 30 ms as a result of the joint stop. As shown in Figure 11, the resultant humerus moment for position 11B demonstrates the same trend as seen with the elbow strain gauge, as bags 2A and 3C have peaks that correspond to the joint stop while the lesser bag 1B does not. Although the humerus moments in position 8B do not correlate with inflator properties, the humerus moments for positions 11B and 5 correlate very well. In position 11B, the resultant humerus moment correlates to peak pressure with a R^2 of 0.99 and to the pressure onset rate with an R^2

of 0.99. The correlation is less with position 5 with R^2 values of 0.87 and 0.90 for peak pressure and pressure onset rate.

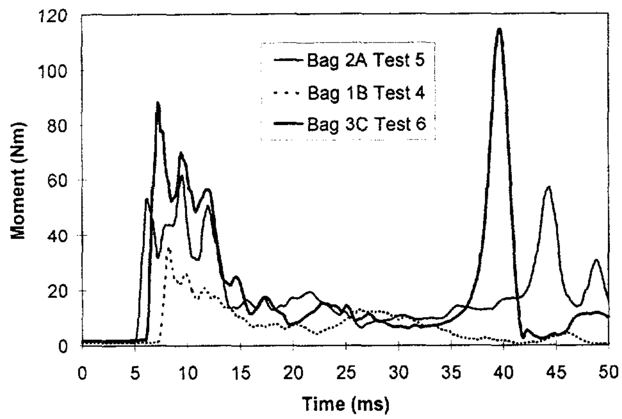

Figure 11: Resultant Humerus Moment (MX, MY) for All Air Bags in Position 11B

Using Duma's (1998) IARV of 128 Nm for the 5th% female humerus, Figure 12 was created to summarize the humerus response for the most severe tests. Since all values are well below 100% of the IARV, no humerus fractures are expected under loading by any of the three side air bags.

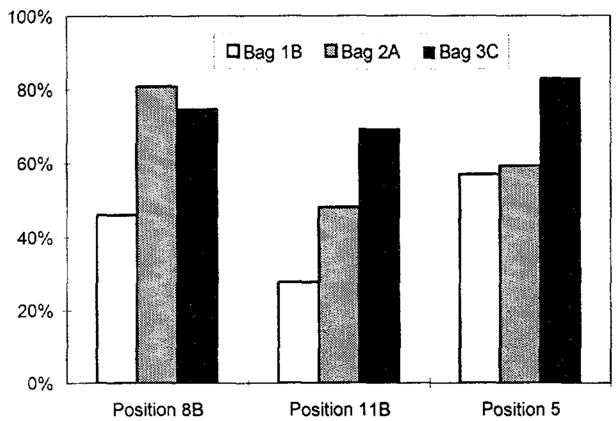

Figure 12: Percent of IARV for Resultant Humerus Moment (MX, MY) for All Air Bags in Three Positions

As expected the chest deflections for positions 8B and 11B were negligible. Only position 5 recorded any deflection of the upper or lower chestband as summarized in Table 4. Given that the injury reference value for peak sternal deflection by a distributed load is 53 mm, no thoracic injuries are expected for this test configuration [Mertz 1993].

Table 4: Peak Chest Deflection for All Air Bags in Position 5

Airbag Type	Upper Chestband (mm)	Lower Chestband (mm)
1B	4.1	2.1
2A	2.5	2.5
3C	12.1	3.1

CONCLUSIONS

The SAE instrumented upper extremity proved effective at evaluating the response under side air bag loading. Using the IARV of 128 Nm for the 5th% female, no humerus fractures are expected as a result of side air bag loading the upper extremity. Further tests are needed to compare the dummy response to that of a small female cadaver in order to examine the biofidelity of the SAE arm.

The MHD angular rate sensors proved useful in determining the rotations of both the shoulder and elbow joints. The large moments that were recorded in the humerus when the elbow reached the joint stop must be evaluated relative to human data to determine their relevance.

The upper extremity response correlated well with the inflator properties for certain positions and sensors. The peak pressure and the pressure onset rate correlated in the same manner for each comparison. Position 5 showed the best correlation with an increase in resultant humerus acceleration, force and moment corresponding to increase inflator aggressivity.

Slight changes in positioning have a significant effect on the occupant response. The long and narrow design of the side air bag allows it to travel the path of least resistance easier than the much larger driver side air bags. For this reason the positioning tolerance was established as ± 2mm for each axis. In addition, the more aggressive the air bag, the higher the tendency for the air bag to deploy to either side of the upper extremity rather than load it fully.

REFERENCES

Bass C.R., Duma S.M., Crandall J.R., George S., Kuppa S., Khaewpong N., Sun E., Eppinger R., The Interaction of Air Bags with Upper Extremity Test Devices, 16th International Technical Conference on the Enhanced Safety of Vehicles, Windsor, Canada, 1998.

Bass C.R., Duma S.M., Crandall J.R., Morris R., Martin P., Pilkey W.D., Hurwitz S., Khaewpong N., Eppinger R., Sun E., The Interaction of Air Bags With Upper Extremities, SAE Paper 973324, 41st Stapp International Car Crash Conference, 1997.

Duma S.M., Schreiber P., McMaster J., Crandall J.R., Bass C.R., Pilkey W.D., Dynamic Injury Tolerances for Long Bones of the Female Upper Extremity, International Research Council on the Biomechanics of Impact, Goteborg, Sweeden, 1998.

Kallieris D., Rizzetti A., Mattern R., Jost S., Priemer P., Unger M., Response and Vulnerability of the Upper Arm Through Side Air Bag Deployment. SAE Paper 973323, 41st Stapp International Car Crash Conference, 1997.

Kirkish S.L., Begeman P.C., Paravasthu N.S., Proposed Provisional Reference Values for the Humerus for Evaluation of Injury Potential. SAE Paper 962416, 40th Stapp International Car Crash Conference, 1996.

Kuppa S.M., Yeiser C.W., Oslon M.B., Taylor L., Morgan R., Eppinger R., RAID - An Investigation Tool to Study Air Bag/Upper Extremity Interactions. SAE Paper 970399, SAE International Congress and Exposition, 1997.

Mertz H.J., Anthropometric Test Devices in Accidental Injury: Biomechanics and Prevention by A.M. Nahum and J. W. Melvin, Springer-Verlag, New York, NY, 1993.

Messerer O., *Uber Elasticitat und festigkeit der Menschlichen Knocken*, Verkag der J.G. Cotta'schen Buchhandlung, Stuttgard, 1880.

National Highway Traffic Safety Administration, Third Report to Congress: Effectiveness of Occupant Protection Systems and Their Use, U.S. Department of Transportation, December 1996.

Sieveka EM, Crandall JR, Duma SM, Pilkey WD. Development and Application of a Side Airbag Computer Model Using a Multi-Body Dynamics Program, 16th International Technical Conference on the Enhanced Safety of Vehicles, Windsor, Canada, 1998.

Weber C., *Chirurgishe Erfahrungen and Untersuchungen*, Berlin, 1859.