THE INTERACTION OF AIR BAGS WITH UPPER EXTREMITY TEST DEVICES

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ABSTRACT

This study examines and compares the response of two upper extremity test devices under driver-side air bag deployment to contribute to the development of dummy surrogates for the investigation of primary contact forearm injuries during air bag deployments. The first of these test devices, the SAE 5th Percentile Female Arm (SAE Arm), is an anthropomorphic representation of a small female forearm and upper arm that is instrumented with load cells, accelerometers and potentiometers to enable the determination of upper extremity kinematics and dynamics. The second, the Research Arm Injury Device (RAID), is a simple beam test device designed for detailed investigation of moments and accelerations resulting from close contact in the initial stages of air bag deployment. The RAID includes strain gauges distributed along its length to measure the distribution of moment applied by the air bag deployment.

The study used four air bags representing a wide range of aggressivities in the current automobile fleet. The upper extremity position was a 'natural' driving posture when turning left with one hand across the steering wheel. The forearm was positioned directly on the air bag module with the forearm oriented perpendicular to the air bag module tear seam. For the SAE Arm, the humerus was oriented normal to the steering wheel. Tests with the SAE Arm were performed both with the arm attached to a 5th Percentile Female Hybrid III dummy and with the arm mounted to a universal joint test fixture. The RAID was mounted to an articulated test fixture. In addition to the dynamic tests, a detailed comparison of the inertial properties of each of the test devices with the inertial properties of a typical small female was performed.

Forearm response from both test devices confirmed the levels of air bag aggressivity determined using previous cadaveric injury results. In addition, logistic risk functions for forearm fracture were developed using existing cadaver studies and the moment response of each test device. These risk functions indicate that for 50% risk of ulna or ulna/radius fractures, the SAE arm peak forearm moment is 61 N-m (+/- 13 N-m standard deviation) while the RAID peak forearm moment is 373 N-m (+/- 83 N-m standard deviation). For 50% risk of fracture of both the ulna and the radius, the SAE arm peak dummy forearm moment is 91 N-m (+/- 14 N-m standard deviation) while the RAID peak forearm moment is 473 N-m (+/- 60 N-m standard deviation).

INTRODUCTION

Although the use of air bag systems as supplemental restraints has significantly decreased the risk of fatality in automobile collisions, there is evidence of increased risk of non-fatal injuries including burns, abrasions, and eye injuries owing to air bag deployment. In addition, case studies suggest that upper extremity injuries, including severe fractures, may be caused by air bag deployment [c.f. **Marco 1996**, **Freedman 1995**, **Huelke 1995**, **Kirchoff 1995**, and **Roth 1993**]. Kuppa *et. al.* analyzed several accident databases to determine the incidence of upper extremity injury for accidents with and without a driver-side air bag deployment [**Kuppa 1997**]. They found that 1.1% of drivers who were restrained only by a seatbelt experienced an upper extremity injury. In contrast, 4.4% of drivers experienced upper extremity injuries in the presence of a deploying air bag.

Two modes of injury have been suggested to explain this increased incidence of upper extremity injuries with air bag deployment. The first type is a flinging type of injury in which the air bag propels the arm into an object in the vehicle (e.g. b-pillar, roof, and occupant's head). The second type is primary contact with the air bag or air bag flap; this injury may occur, for example, while executing a left turn with a continuous motion of the right hand, placing the forearm directly over the module. It is the latter group, primary contact injuries, that is the subject of the current study.

Case studies and NASS data suggest that these severe upper extremity injuries occur predominantly in women. It may be hypothesized that this represents the effects of three factors: 1) as women are generally shorter in stature than men, they drive closer to the steering wheel/air bag module, 2) women experience an age-related loss of bone mineral density, and 3) women have generally smaller bones and, hence, lower ultimate bone strength.

To investigate the upper extremity/air bag interactions causing these injuries, Saul *et al* used an instrumented 50th Percentile Male Hybrid III upper extremity to examine injury from direct contact **[Saul 1996]**. Using strain gauges and accelerometers, they found that bending moments and accelerations of the forearm could be accurately recorded. Moreover, a correlation was found between these values and the air bag's inflator properties, flap, and steering wheel orientation. In addition, the forearm bending moment response of the instrumented SAE 5th Percentile Female Arm (SAE arm) under primary air bag contact has been correlated with cadaveric injury to produce an injury risk function for small females **[Bass 1997]**.

In addition, the Research Arm Injury Device (RAID) was developed by Conrad Technologies Inc. and NHTSA to investigate the interaction between a deploying air bag and an upper extremity in close proximity to the air bag **[Kuppa 1997]**. They found that the two most significant determinants of peak measured bending moment were the orientation of the arm with respect to the air bag module and the separation distance between the two. Maximum moments were recorded when the forearm was positioned perpendicular to the air bag module. This situation occurs, for example, when making a left turn with the right hand. In this situation, the right and left sides of the air bag are at the 1 and 7 o'clock positions respectively, while the hand and elbow are at the 10 and 4 o'clock positions respectively. The maximum moments also decreased as the distance between the air bag and the forearm was increased from 1.3 cm to 7.6 cm.

It is likely that a specific air bag design is developed with a view toward total restraint system effectiveness. As different passenger automobiles have different physical sizes and stiffness, this results in installed air bags of different deployment properties (e.g. pressure-time histories, module design, and deployment characteristics) among vehicle models. Four OEM air bag types were used in this study; these air bags were identified using RAID testing as representing a wide range of aggressivities in the current passenger car fleet. Using a previously created coding scheme **[Bass 1997]**, these systems are termed System H, System K, System J, and System L air bags. The System H and System K air bags produce relatively more aggressive air bag deployments, the System J air bag produces a moderately aggressive deployment, and the System L air bag produces a relatively less aggressive deployment. In addition, the System H air bag has been identified in case studies as producing primary contact upper extremity injuries under certain circumstances.

The principal goal of this study is to examine the suitability of both the SAE arm and the RAID in characterizing the forearm forcing during air bag primary contact using OEM air bag systems. In addition, this study quantifies the dynamic response of dummy upper extremities under air bag deployment in a 'worst-case' position. Also, the study investigates factors that affect injuries in cadaveric upper extremities and develops a correlation of these injuries with dummy response using the SAE Arm and the RAID. As there are a number of design factors that may influence the upper extremity injury potential of a given air bag, including inflator properties, air bag properties and module properties, we have chosen to focus on dummy and cadaveric response criteria as the most effective measure of injury risk.

The testing was performed in two major parts. The first includes tests of the RAID test device under air bag deployment in a representative 'worst-case' position for air bag deployment. The second is a study of the same set of deploying air bags into the SAE arm mounted on a Hybrid III dummy and tests with the SAE arm attached to a universal joint arm fixture developed for cadaveric studies **[Bass 1997]**. This second series of tests involves forearm positioning similar to that prescribed for the RAID testing.

TEST DEVICES

Several instrumented dummy arms exist that are appropriate for use in arm/air bag interaction studies; these include the 50% Male Hybrid III Instrumented Arm **[Saul 1996, Johnston 1997]**, the Research Arm Injury Device (RAID) **[Kuppa 1997]**, and the SAE 5th Percentile Female Instrumented Arm (SAE arm) **[Bass 1997]**. As the epidemiogical analysis of air bag-induced upper extremity injuries suggests that small females suffer injuries at a much greater rate than males, this study investigates the use the SAE arm and the RAID as suitable dummy surrogates for the development of risk functions using previously reported small female cadaveric injury studies **[Bass 1997]**.

A diagram of the SAE arm is shown in *Figure 1*. Pronation/supination of the forearm is provided by a single degree-of-freedom axial 360^0 rotation in the wrist. The forearm is a single shaft incorporating a six-axis load cell located approximately mid-shaft. The elbow is a single degree-of-freedom clevis joint allowing elbow flexion/extension with a soft joint stop in each direction. This elbow motion may be measured using a potentiometer incorporated into the elbow. In addition, strain gauges to measure two bending axes are located in the distal humerus. The humerus is a single shaft with a six-axis load cell approximately midshaft. At the proximal end of the humerus, two degree-of-freedom rotations are allowed by a 360^0 axial rotation at the top of the humerus shaft and a clevis joint at the shoulder. In addition to the existing instrumentation on the SAE arm, the current study added a distal triaxial accelerometer and a single-axis MHD angular rate sensor mount located one third of the distance from the wrist to the elbow. Additional accelerometer mounting locations in the elbow were not used.

Motions allowed by the SAE arm listed in *Figure 2* are approximately anthropomorphic with the exception of pronation/supination and shoulder motions. For pronation/supination, the existence of a single shaft forearm limits both the availability and the utility of forearm rotations located outside the wrist. Though the predominant flexion/extension motions and upper humerus rotations are represented in the SAE dummy shoulder, the human shoulder has three degrees-of-freedom in rotation and limited translation that is not seen in the dummy.

In contrast, the RAID, shown in *Figure 3*, has a more limited range of motions. Developed as an investigative tool to study primary contact arm/air bag interactions, the RAID is constructed of a 3.2 mm thick aluminum tube of 51 mm diameter with a two degree-of-freedom clevis joint to allow rotational motion along two axes. The mass of the tube (1.6 kg) was chosen to approximate a 50th percentile male human forearm. To simulate the effects of a hand, a small additional mass (0.5 kg) is attached to the free end of the RAID. The length of the RAID was selected as 460 mm to protect the pivot attachments from the deploying air bag. The RAID instrumentation includes five stations of diametrically opposed strain gages to measure moments along two axes. In addition, rotations are measured by two angular potentiometers, and triaxial accelerations are measured at the approximate mid-length of the RAID. The RAID is covered with 20 mm of foam and rubber skin similar to that on the Hybrid III mid-forearm.

As the RAID incorporates simple two-dimensional rotation, the RAID simulates only the forearm degrees of freedom associated with elbow flexion/extension and shoulder abduction/adduction. So, while the RAID may be appropriate for primary contact with a deploying air bag, it is likely not appropriate for later interactions involving additional upper extremity degrees of freedom.

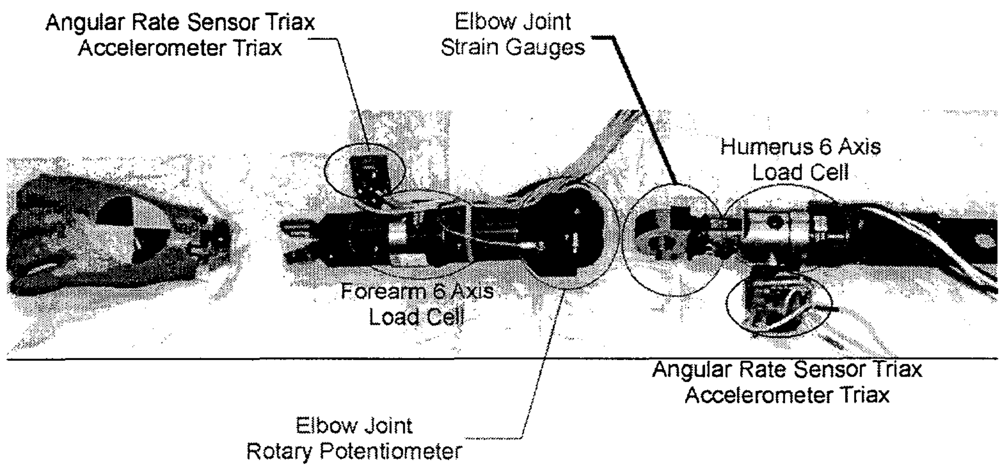


Figure 1. *Picture of the SAE 5th Percentile Female Instrumented Arm (SAE Arm).*

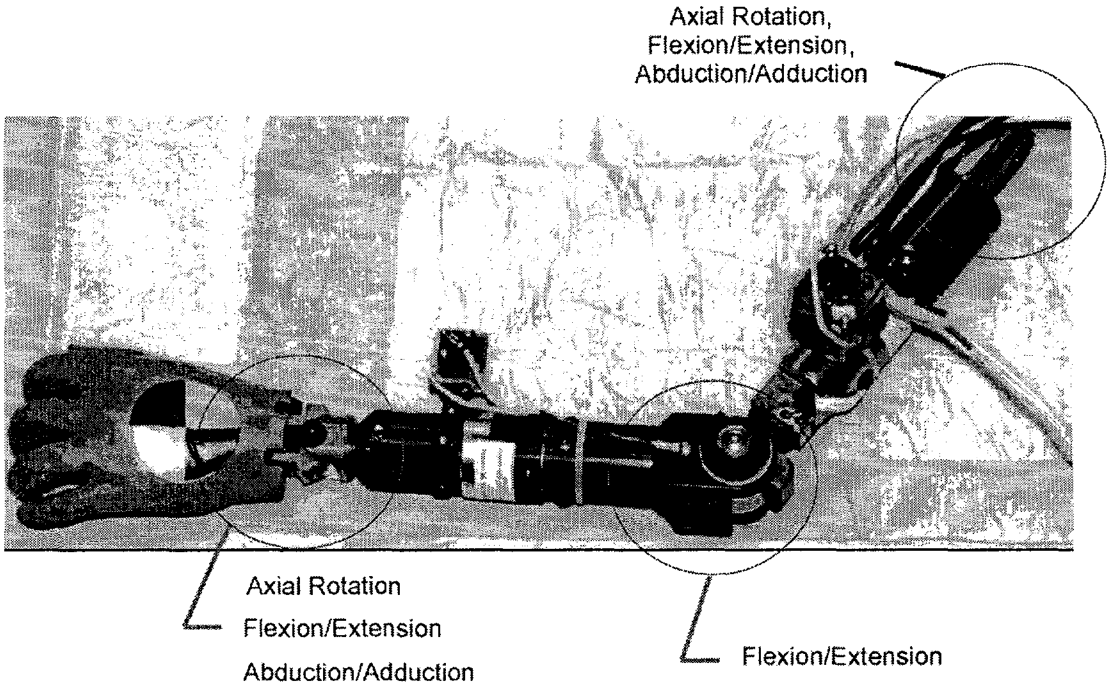


Figure 2. Motions of the SAE 5th Percentile Female Arm.

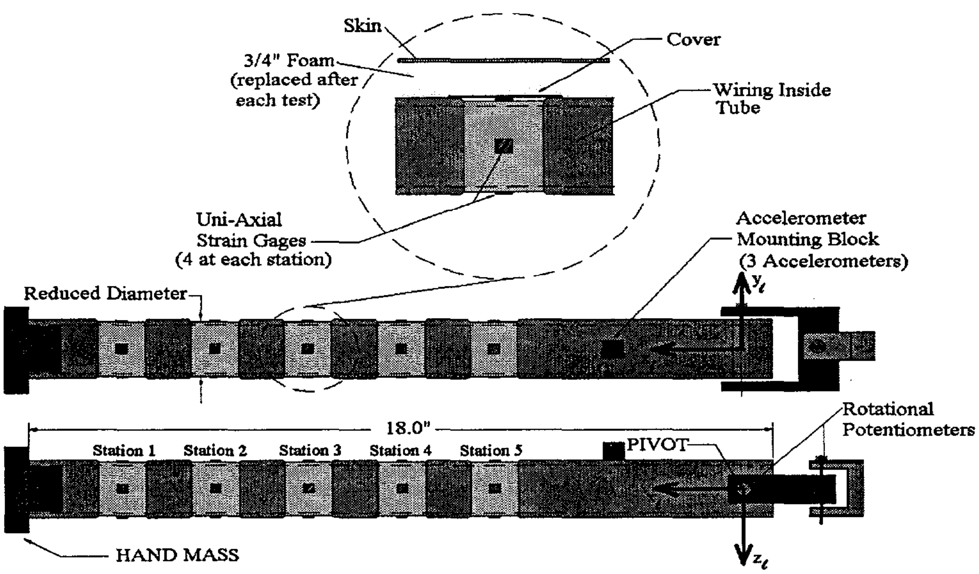


Figure 3. Research Arm Injury Device (RAID).

A comparison of the segment masses of the SAE arm and the RAID with the 5th and 50th percentile female population are shown in **Error! Reference source not found.**. The SAE arm is substantially heavier than the reference 5th percentile female population but is similar to the reference 50th percentile female population. The RAID, however, was designed to simulate a 50th percentile male. So the RAID is substantially heavier than the forearms of the reference female populations.

Reference forearm and hand lengths shown in **Error! Reference source not found.** were derived from an anthropometric study on 1905 USAF women **[McConville 1979]**. For the human population, the forearm length is taken to be the distance from the tip of the olecranon to the tip of the ulna styloid process, and the hand length is the distance from the ulna styloid process to the middle finger of the outstretched hand. The dummy arm measurements are taken from the rotation centers.

The total forearm/hand length of the SAE arm of 405 mm is comparable to the 5th percentile female length of 397 mm but over 20 mm less than the forearm/hand length of the reference 50th percentile female. In contrast, the total length of the RAID (460 mm) is much larger than the forearm/hand length of the reference female population and is comparable to the forearm/hand length of a 50th percentile male population (491 mm) but significantly larger than the forearm length of a 50th percentile male population (299 mm). So, the SAE dummy forearm is similar to a 5th percentile female population in length but a 50th percentile female population in mass ,while the RAID is, by design, similar to the 50th percentile male in mass and forearm/hand length.

***Table 1.** Comparison of Reference and Dummy Arm Anthropometry*

	Mass (kg)			Length (mm)		
Arm	**Forearm**	**Hand**	**Total**	**Forearm**	**Hand**	**Total**
Reference 5th % Female[1]	0.71	0.28	0.99	227	170	397
Reference 50th % Female[1]	0.90	0.36	1.26	244	184	428
Reference 50th % Male[1]	1.3	0.5	1.80	299	192	491
SAE Instrumented 5th % Female Arm	1.08	0.41	1.49	240	165	405
RAID[2]	1.6	0.5	2.1	460		460

Three-wire torsional pendulum studies were performed on the segments of the SAE arm to determine inertial properties in the principal axes. Axes of rotation passing through the segment center of gravity define all moments, and the reference female forearms are oriented in the neutral position. The x and y principle axes of the dummy and reference female forearms are approximately normal to the anatomical axis running along the forearm. The z principle axis is approximately tangential to this axial axis. Moments of inertia were calculated for the RAID assuming a uniform aluminum cylinder.

For primary contact injury under air bag deployment, kinematics observed in previous dummy and cadaver studies [c.f. **Bass 1997**] indicates that there is no significant motion of the humerus prior to peak moments or cadaveric injury. So, the inertial properties of the upper arm are negligible in the investigation of surrogate response under primary contact air bag deployment. Also, the dynamic significance of pronation/supination (axial rotation of the forearm) motions is minimal in primary contact studies, so the principle moment of inertia about the axial axis is of limited significance in this study.

For the SAE arm, the influence of the mass of the centrally located load cell on the x and y principle moments of inertia is clear. Though significantly heavier than the reference 5th percentile female reference population, the SAE arm has x and y moments of inertia that are comparable to the reference 5th percentile female population. A significant portion of the mass of the SAE forearm is included in this load cell. The z (axial) moment of inertia of the SAE arm, however, is larger than the 5th percentile female owing to the size of the SAE arm. In addition, owing to the substantial mass of the SAE hand, principle moments of inertia in the x and y axes are much larger than those of the reference 5th percentile female population, and are more comparable to those of the 50th percentile female.

For the RAID, the length is significantly greater than the forearm of either female reference population or that of a 50th percentile male. So, the moment of inertia of the RAID forearm segment is substantially larger than that of either reference female population. In air bag tests, this

[1] **[McConville 1979]**
[2] RAID is single segment.

will likely result in lower peak velocities and possibly much larger moments. This imposes an additional limitation on the use of the RAID in the investigation of 'flinging' injuries in which maximum velocity plays an important role in injury mechanics. The RAID axial moment of inertia (z axis) of 1040 kg-mm^2 is commensurate with a 50th percentile male value of 1180 kg-mm^2 **[McConville 1979]**. The 'hand' mass of the RAID can be considered to be concentrated at the end of the RAID for the purpose of this study as the test device allows only rotations about the other end.

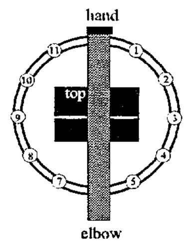
Table 2. *Principle Moments of Inertia*

Arm	Forearm Principle Moment of Inertia (kg mm^2)			Hand Principle Moment of Inertia (kg mm^2)		
	x	y	z	x	y	z
Reference 5th % Female[3]	3100	2900	410	340	280	94
Reference 50th % Female[1]	4700	4600	630	530	430	150
Reference 50th % Male[1]	8850	8610	1180	1040	850	290
SAE Instrumented 5th % Female Arm	2800	2300	550	970	450	55
RAID	28200	28200	1040	NA	NA	NA

EXPERIMENTAL SETUP

Both the RAID and the SAE arm attempted to attain a 'worst-case' test condition and hence a 'worst-case' response under air bag deployment. The test position selected is roughly a 'natural' driving position in a one-armed left turn maneuver modified for enhanced repeatability and 'worst-case' behavior. The SAE forearm was placed directly on the air bag module with the forearm oriented perpendicular to the air bag tear seam as shown in *Figure 4.* The distal third of the SAE forearm was placed over the module tear seam, and the humerus was oriented normal with respect to the plane of the steering wheel. In this configuration, the dummy fingers do not reach the steering wheel for any of the OEM air bags tested. Positioning was maintained using frangible tape.

This position represents the 'worst case' or most vulnerable position for four reasons. First, previous RAID testing indicated that bending moments were maximized when the test device was oriented perpendicular to the air bag tear seam **[Kuppa 1997]**. Second, RAID bending moments under air bag deployment were found to decrease as the test device was moved away from the module. Though the RAID was placed at distances 13 mm and greater from the air bag module, out-of-position thoracic testing **[Melvin 1993, Bass 1998]** suggests that positioning directly on the air bag module may constitute a worst case for certain occupant/air bag interactions. Third, the distal third of the human forearm is the weakest location in bending with the lowest combined polar moment of inertia of both the radius and ulna, providing the greatest risk of fracture. Fourth, the humerus oriented normal to the steering wheel provides a support for the forearm under air bag deployment forcing the initial center of forearm rotation to be about the elbow with a relatively long moment arm.



Config. V

Figure 4. *Test Configuration - Arm Relative to Steering Wheel.*

Eight of ten SAE arm tests were performed on a universal joint test fixture diagrammed in *Figure 5.* The fixture is comprised of two components. The first supports the steering wheel/air bag module on a five-axis load cell. The second mounts the arm to a four degree-of-

[3] **[McConville 1979]**

freedom universal joint. A five-axis humerus load cell was mounted at the interface between the SAE arm and the universal joint at the shoulder. For the fixture tests, the center of rotation of the universal joint was located at a position equivalent to the center of rotation of the Hybrid III shoulder joint relative to the humerus. The remaining two tests were performed with the SAE arm attached to the Hybrid III 5th percentile female dummy.

One possible objection to the use of the test fixture is that, for experimental convenience, the location of the point about which the shoulder rotates is fixed in space. In a natural driving condition, the shoulder is relatively free to translate in response to forcing. This translationally fixed shoulder was examined using the Articulated Total Body (ATB) lumped-mass simulation program as shown in *Figure 6*. The figure shows a comparison of the humerus axial force for a subject with a shoulder fixed in translation versus a shoulder free to translate under the action of a deploying air bag. There is little difference in humerus response between the two cases, especially in the crucial initial deployment period. This result justifies using a shoulder that is fixed in translation for the experimental setup.

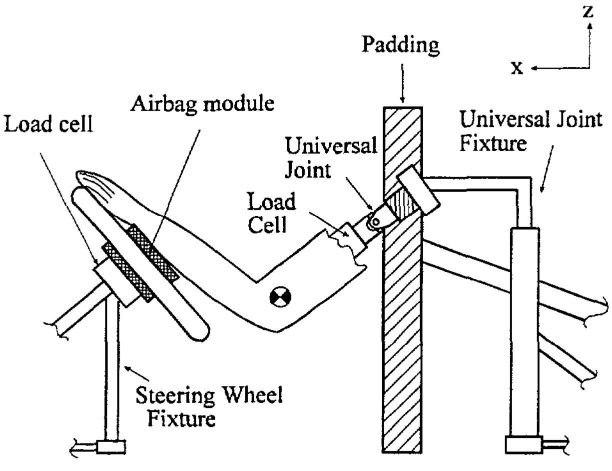


Figure 5. *Arm/Air Bag Test Fixture.*

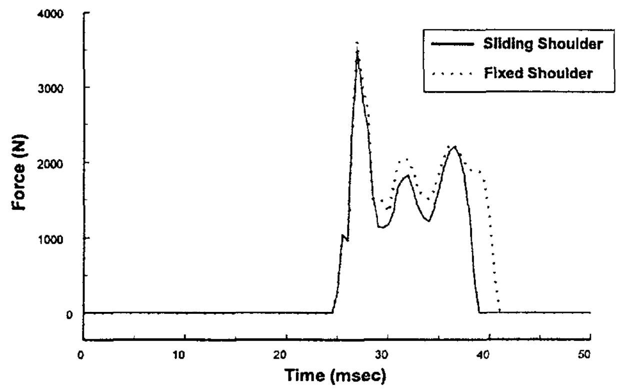


Figure 6. *ATB Simulation of Fixed vs. Sliding Shoulder.*

A side view of the test setup with the RAID is shown in *Figure 7*. The RAID hangs vertically in front of the steering wheel and rotates at the mounting pivots. The test device may be translated in three dimensions to achieve desired positioning with respect to the air bag module. For this study, the distance from the surface of the RAID to the plane of the steering wheel rim was set to 13 mm to achieve 'worst case' response. Positions closer to the steering wheel were not investigated. The steering wheel was oriented as shown in *Figure 4* with the RAID perpendicular to the air bag tear seam. As with the SAE arm tests, a five-axis load cell was located behind the steering wheel to measure reaction forces. The time of air bag cover opening was determined using break wires over the tear seam. In addition, a backstop with foam padding was used to stop the RAID after the test.

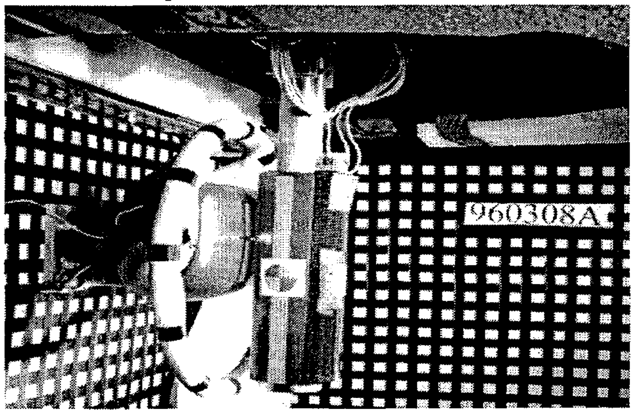


Figure 7. *Side View of RAID.*

EXPERIMENTAL RESULTS

Four OEM air bags that are representative of a wide range of air bag aggressivities in the current automobile fleet were used in the testing. These air bag systems, in order of decreasing aggressivity identified in previous RAID testing [**Kuppa 1997**], are denoted System K, System H, System J, and System L. The air bags were mounted in original equipment steering wheels appropriate for the air bag tested. Inflator performance of each air bag system from tank testing (60 L tank) is shown in *Figure 8*. Tank tests for System J are not available. Tests on the remaining inflators confirm the ordering of aggressivity suggested in the RAID testing. The System K inflator is very aggressive with a high peak pressure and a high pressure onset rate. During this study, several System K air bags burst around the vent holes during deployment. System H inflators are also very aggressive with peak pressures slightly lower than those seen in System K inflators but with a high pressure onset rate. The System L inflator is relatively non-aggressive with a very low peak pressure and onset rate.

There are significant differences in the air bag modules, especially the location of the module tear seam as shown in *Figure 9*. The tear seams for the System K and System L modules are approximately mid-way between the top and the bottom of the module. In contrast, the System H module has a very large and heavy flap with a low tear seam. This large flap has been found to provide some protection during the initial air bag deployment to cadaveric arms under air bag deployment [**Bass 1997**]. System J has a relatively small vertically oriented tear seam with wide side flaps. The air bags are all similar in height and width, and the steering wheels are similar in dimension. Only the System J air bag is untethered.

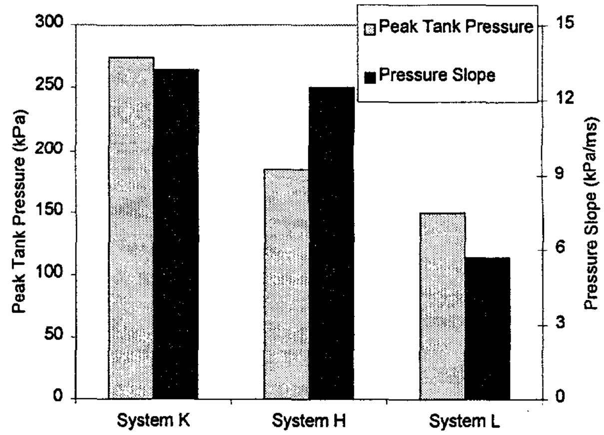


Figure 8. *Static Tank Pressure and Pressure Slope Curves – Values Based on a 60 L Tank, Pressure Slopes Derived from Maximum 10 ms Values.*

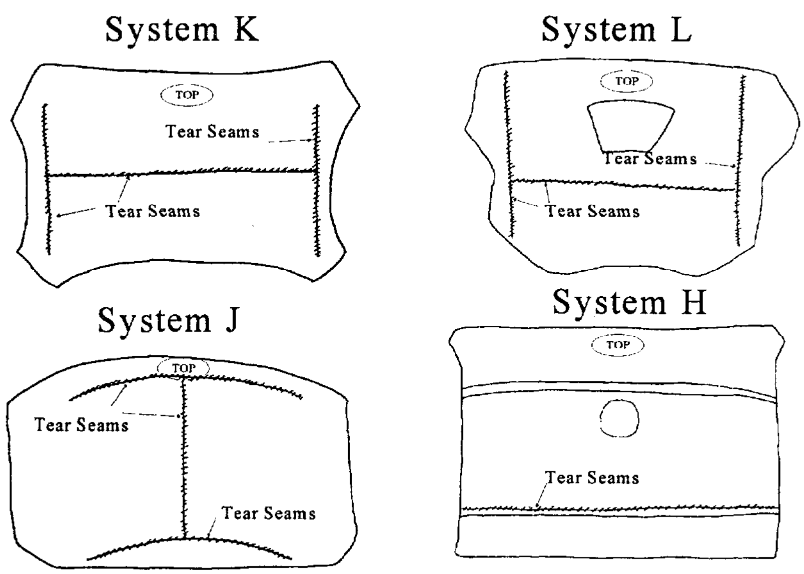


Figure 9. *Sketches of Air Bag Module Covers Indicating the Tear Patterns.*

Table 3. *Characteristics of Air Bag Modules (All Measurements in mm)*

Air Bag System	System H	System L	System K	System J
Diameter of steering wheel	380	380	387	397
Location of module plane wrt. wheel	3.2 above	9.5 above	6.4 above	6.4 above
Distance from top of rim to tear line	259	238	222	200
Vertical height of module	178	152	152	152
Horizontal width of module at seam	203	171	191	216
Distance from top of module to seam	138	91	78	108
Thickness of flaps	3.2	4.4	5.1	3.2
Vertical height of air bag	686	699	635	660
Horizontal width of air bag	686	635	660	635
Number of tethers	4	2	3	none
Length of tethers	267	279	318	--

System K, System H, System J, and System L air bags were each tested twice with the SAE arm mounted on the test fixture used in previous cadaveric tests. In addition to the fixture tests, one System H air bag and one System L air bag were deployed into the SAE arm mounted on a 5th Percentile Hybrid III dummy. For the RAID, one test was performed using each of the air bag systems in this study. In addition to these tests, several repeatability tests were performed with the System K and System J air bags.

A typical deployment for both test series begins with a bulge in the air bag module following air bag initiation. Then, the air bag deploys through a scored tear seam oriented perpendicular to the forearm. In the initial stages of air bag inflation with the SAE arm, there is no significant humerus motion, and the forearm begins to rotate about the elbow until it reaches the joint stop. After the elbow reaches the joint stop, the humerus begins rotating toward the center of the steering wheel. This continues until the SAE arm hits the dummy in the Hybrid III tests or the backstop in the fixture tests. For the RAID, the deployment rotates the arm until the arm contacts the padded backstop. For the SAE arm mounted to the Hybrid III, there is no substantial shoulder movement until the air bag deploys into the dummy chest. Moment time histories from both test devices suggest that the greatest forces on the forearm occur during the air bag punch-out and shortly thereafter.

All air bags deployed normally except for one of the System K air bags in the SAE arm testing. As seen in a previous cadaveric test series [**Bass 1997**], the System K air bag suffered large tears during the deployment originating at the reinforced seam around the peripheral vent holes. In spite of these holes, the air bag appeared to inflate fully.

Peak resultant forearm bending moments for both the SAE arm and the RAID are presented in ***Error! Reference source not found.***. For repeated tests within each air bag system using the SAE arm, these moments are consistent, showing a maximum difference of approximately 10%. Peak humerus axial loads for the SAE arm are not as consistent since they are associated with the details of the air bag/elbow joint stop interaction at times greater than injury times identified in cadaveric tests. So, these peak humerus axial loads are not generally relevant to primary contact injuries.

Peak moment values for the RAID are much larger than those measured using the SAE arm. This is likely the result of the RAID having greater mass and moments of inertia than the SAE arm. In addition, the ordering of aggressivity quantified using peak moments of the System K and System H air bags is reversed in the RAID from that found using the SAE arm. This is likely the result of a heavier air bag module cover with a lower air bag seam than the rest of the test devices. As the SAE arm testing placed the distal third of the forearm on the air bag tear seam while the RAID maintained uniform radial placement with respect to the steering wheel, the larger, lower flap of the System H air bag tends to increase peak moments for the RAID relative to the SAE arm. However, the peak moments derived from testing using the SAE arm and testing using the RAID, compared in *Figure 10*, show a correlation coefficient of 0.90 indicating similar peak moment response.

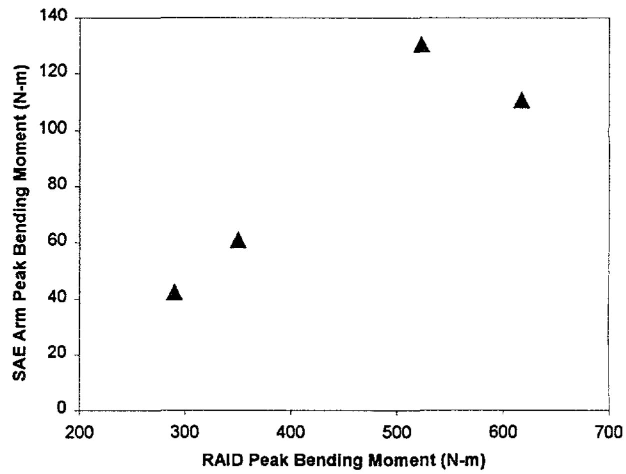


Figure 10. Peak Forearm Moments of the SAE Arm and RAID.

Forearm moment time histories for each of the systems tested are plotted in *Figure 11* for the SAE arm and in *Figure 12* for the RAID. As expected, the peak SAE arm forearm bending response of System K and System H is significantly greater than that seen in System J or System L. Large peak moments after 15 ms are associated with the SAE arm elbow reaching the joint stops. Interestingly, the peak forearm moments from the System K tests are much earlier than those seen in the System H tests. High-speed video analysis indicates that while peak bending moments occur during module cover/arm interactions for System K, the peak moments for System H occur after the time that the arm interacts with the module cover. This indicates that while the module cover may play a role in injuries, module cover interaction may not be necessary for such primary contact injuries.

For the RAID, the timing of forearm moment peaks is similar to those found with the SAE arm. The System K air bag has the earliest peak, and the System H air bag has the latest peak moments. Though the order of peak forearm moment was switched between System L and System J air bags for the RAID and SAE arm, the timing of these peak moments was similar. Both the SAE arm and the RAID show peak moments for System H after the time of arm/module cover interaction.

In contrast, with the System K air bag, the second moment peak appears later for the SAE arm than for the RAID. Because the acceleration of the SAE arm is much greater than that of the RAID, the RAID is closer to the inflator when the air bag emerges from the module cover, producing earlier peak moments. The kinematics of the SAE arm appear to be more consistent with cadaveric test results. In addition, the SAE arm moment peaks generally maintain the order of aggressivity found in previous cadaveric testing.

As expected from the inertial properties, the peak accelerations shown in **Error! Reference source not found.** using the SAE arm are substantially larger than the peak accelerations found using the RAID even though the accelerometers were placed in similar locations. The RAID has a 40% greater forearm/hand mass and nearly ten times the forearm lateral moment of inertia. For the SAE arm, the accelerations generally maintain the order of aggressivity found in previous cadaver tests. The relatively aggressive System K air bag demonstrated over twice the peak forearm acceleration than the other three air bag systems tested. The System H and System J air bags showed comparable peak accelerations; however, the more aggressive System H air bag delivered approximately 10% more peak impulse to the distal forearm than the System J air bag during the first 15 ms of deployment. The similarity of peak accelerations with dramatically different peak moments may be accounted for by differences in air bag deployments between System H and System J. From high-speed video, the System J air bag appears to deploy in a smaller forearm area than do the System H air bags. One likely source of this difference is the lack of tethers in the System J air bag. This concentration of air bag deployment may lead to increased risk of fracture relative to a tethered bag. In addition, the System H air bag deploys generally more distally than the System J air bag when accounting for the difference in SAE forearm position with respect to the steering wheel. This effect is not present in the RAID tests as the test device was not adjusted radially to account for the differences in air bag tear seam location. The less aggressive System L air bag demonstrated peak accelerations and impulses that were substantially lower than the other air bag systems.

Table 4. *SAE Arm and RAID Peak Response Data*

Value	Test Device	System K	System H	System J	System L
Peak Forearm Moment (N-m)	SAE arm	131	111	61.0	42.5
	RAID	522	617	350	280
Peak Distal Accel. (g's)	SAE arm	450	187	208	137
	RAID	137	183	65	57
Peak Humerus Axial Load (N)	SAE Arm	2110	1660	1680	940
	RAID	NA	NA	NA	NA

Measured shear loads in the SAE dummy forearm were relatively low, under 800 N for all tests. Such shear loads are unavailable in RAID instrumentation. These low forearm shear loads are likely the result of the center of pressure of the air bag deployment being close to the center of the load cell.

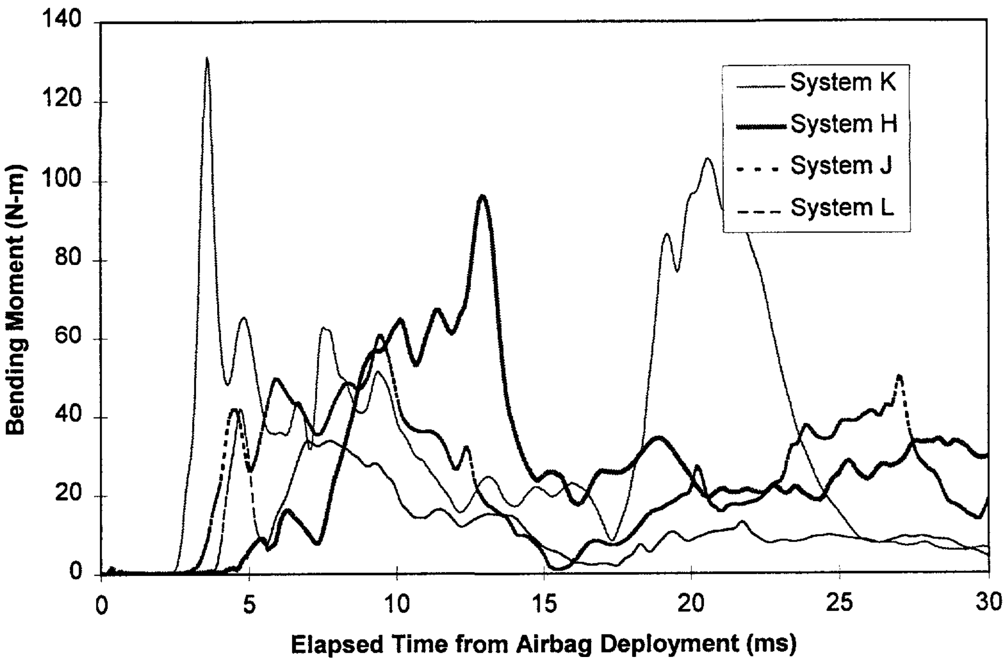


Figure 11. SAE Arm Midshaft Forearm Resultant Bending Moment (All Signals Filtered to SAE CFC-600).

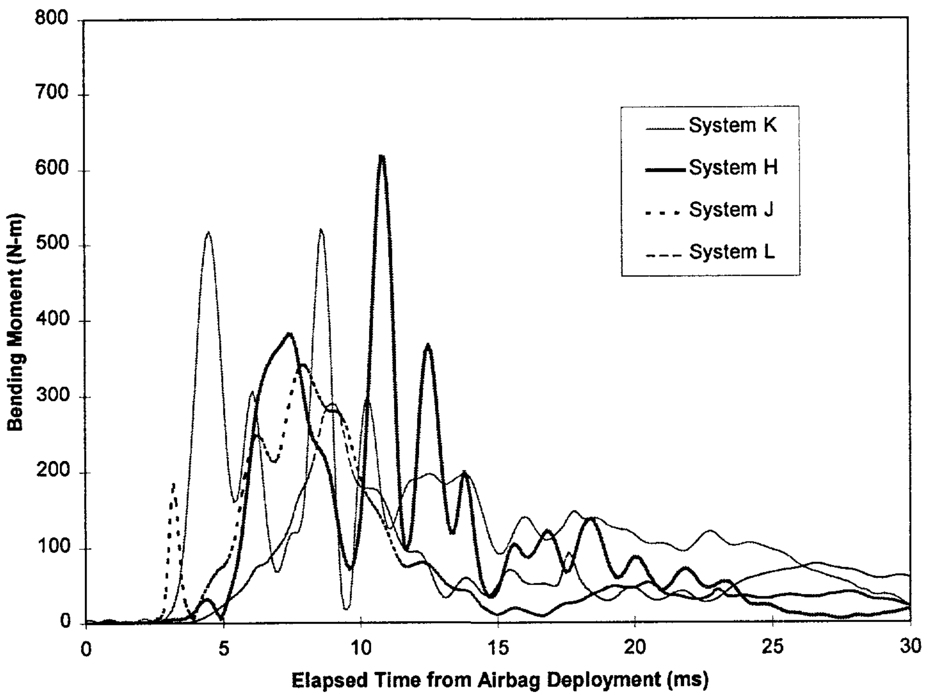


Figure 12. RAID Midshaft Forearm Resultant Bending Moment (All Signals Filtered to SAE CFC-600).

SAE elbow flexion is shown in *Figure 13* under System H air bag deployment for typical dummy and cadaver tests. The two tests see peak flexion angles of approximately 50^0 with similar timing. The minimal effect of the soft joint stop is seen in the System H tests. The SAE arm enters the joint stop region of 40^0 flexion at approximately 18 ms and reaches the limits of travel at approximately 23 ms. In contrast, the effect of the joint stop on the SAE arm is seen clearly in the System K air bag deployment. The slope of the flexion is substantially larger than that seen in the System H dummy tests, so the arm attains larger velocities and hence larger forearm bending moments entering the joint stop. On the other hand, all dummy tests see the elbow reach the joint stop later than 20 ms from the time of air bag deployment. As this time is much later than the time of primary contact injury as determined in the cadaveric tests, the behavior in the joint stop is not relevant for research into primary contact injuries. So, although flexion to simulate a human elbow is not expected to be biofidelic with the RAID, these results suggest that the lack of a biofidelic humerus may not detract from use of the RAID as a diagnostic device for investigation of primary contact air bag injuries.

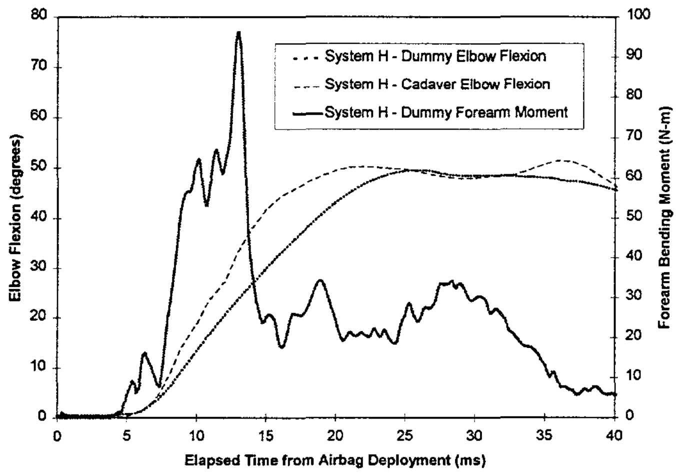


***Figure 13.** SAE Arm - System H - Dummy vs. Cadaver Elbow Flexion and Dummy Forearm Moment (Moment Filtered to SAE CFC-600, Flexion Angles Filtered to SAE CFC-1000).*

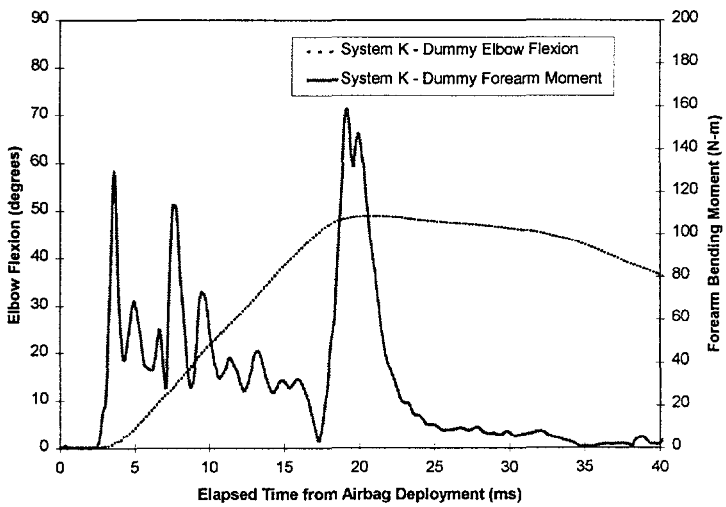


***Figure 14.** SAE Arm - System K - Dummy Elbow Flexion and Dummy Forearm Moment (Moment Filtered to SAE CFC-600, Flexion Angles Filtered to SAE CFC-1000).*

Figure 15 shows the similarity of the responses of System L air bag deployments into the SAE arm with the arm on the dummy compared to the arm mounted to the universal joint fixture. This suggests that such fixture tests are appropriate for simulation of primary contact arm/air bag interactions. The three tests show resultant forearm peak moments that are within 14%, and the timings of the initial peaks are within 2 ms. Similar repeatability in the resultant forearm moments is seen in the System H tests. These results provide additional evidence that the use of the fixed test fixture with the SAE arm is appropriate for investigation of primary contact arm/air bag interactions.

In addition, with the System L air bag, we can separate the effects of arm/flap and arm/air bag interaction. The first peaks in bending moment are the result of flap deployment into the arm, ending at approximately 7 ms as identified from high-speed video analysis. The second peaks, however, are solely the result of arm/air bag interaction. These second peaks rival the first in magnitude for each of the tests and have substantially greater impulse.

For the RAID, the results of two repeated tests using System K air bags are shown in *Figure 16*. The initial peak in resultant moment in repeated tests using the System K air bag shows only 3% difference in value and 0.1 ms difference in peak timing. In addition, the second peak shows less than 10% difference in value with a 0.5 ms difference in peak timing. Additional repeated tests with the System J air bag showed good repeatability in both peak values and timing. So, both the RAID and the SAE arm showed good overall repeatibility.

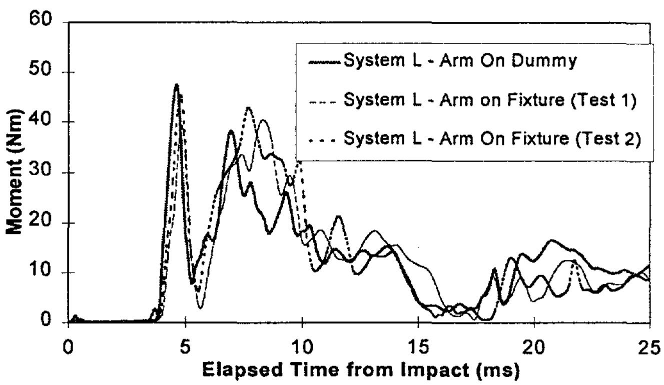


Figure 15. *SAE Arm - Forearm Resultant Moment - Arm on Dummy vs. Arm on Fixture (Signals Filtered to SAE CFC-600).*

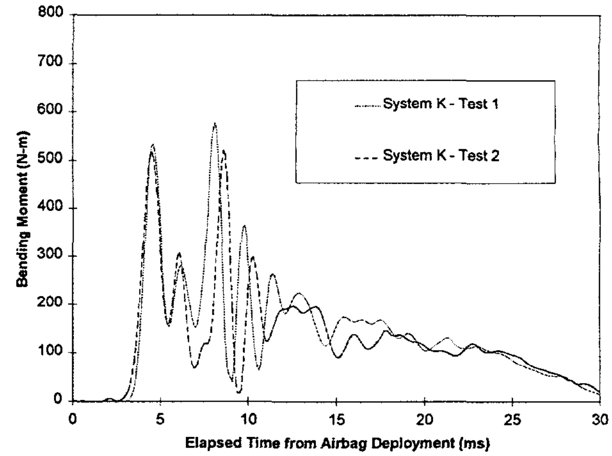


Figure 16. *RAID Repeatability – Forearm Moment Response for System J Air Bag Deployment.*

INJURY RISK FUNCTIONS

Since the currently reported arm/air bag tests were performed in nominally the same condition as previous cadaveric tests **[Bass 1997]**, we can correlate the injury results from the cadaveric testing with the average peak forearm bending moments resulting from air bag deployment into the SAE arm and the RAID. This is further justified by the strong correlation between the peak forearm moment response of the SAE arm and the RAID. For the forearm bending moments, we use all the tests with the System K, System H, System J, and System L air bags as seen in ***Error! Reference source not found.***. For the cadaveric forearms, we limit the injury sample to the 11 small female cadaveric subjects tested with the same air bags on the SAE arm test fixture. For a model of fracture/no fracture, the cadaveric response shows complete separation at an average SAE arm peak forearm moment value of 61 N-m. If, however, we assume a polytomous process where the level of fracture in the cadaveric tests is associated with the average bending moment for the repeated tests with a given air bag, we obtain the logistic regression for the probability of either an ulna or an ulna/radius fracture for the SAE arm as shown in *Figure 17*. The result is statistically significant to $p=0.02$. The regression suggests a 50% risk of at least one fracture at 67 N-m (+/- 13 N-m Standard Deviation) forearm moment in the SAE arm under the same test conditions. The risk of both radius and ulna fracture using the same model for the SAE arm is shown in *Figure 18*. This curve suggests that there is a 50% risk of both radius and ulna fracture at 91 N-m (+/- 14 N-m Standard Deviation) peak forearm bending moment in the SAE arm. For both logistic risk curves, the one standard deviation confidence intervals are plotted.

Injury risk functions for the RAID using peak moment values correlated with cadaver injury data are shown in *Figure 19* and *Figure 20*. These risk functions indicate that for 50% risk of ulna or ulna/radius fractures, the RAID peak forearm moment is 373 N-m (+/- 83 N-m standard deviation). For 50% risk of fracture of both the ulna and the radius, the RAID peak forearm moment is 473 N-m (+/- 60 N-m standard deviation). The results are statistically significant to p = 0.06, and the one standard deviation confidence intervals are plotted.

Table 5. *Test Device Peak Bending Moments vs. Cadaveric Injuries [Bass 1997].*

Air Bag	Average Peak Forearm Bending Moment (N-m) SAE Arm	RAID	Cadaver Tests	Ulna/Radius Fractures
System K	131	522	2	2/2
System H	111	617	3	3/2
System J	61.0	350	4	2/1
System L	42.5	290	2	0/0

These risk functions for forearm fracture can be analyzed using the available quasistatic ultimate bending moments for isolated arm bones reported above. Grouping all the available tests, we obtain a weighted average value of 39 N-m for ulna ultimate strength. Carter and Hayes **[Carter 1976]** suggest dynamic dependence on strain rate of the form $F \propto \varepsilon^{0.06}$ where F is a compressive ultimate load and ε is the dynamic strain rate. For our typical dynamic strain rates of 5 per second, the Carter and Hayes strain rate dependence results in 53% increase in ultimate strength for dynamic bending as compared with UVa quasistatic ultimate strength for the ulna. This is consistent with the suggestion of Melvin and Evans **[Melvin 1985]** who suggest an increase of 50% for dynamic ultimate strength over quasistatic ultimate strength. Further, Schreiber *et al* **[Schreiber 1997]** report a 68% increase in the dynamic bending strength of the tibia over quasistatic tests at strain rates of 5 per second.

So, if we assume 50% increase in the ultimate strength of the isolated ulna, the dynamic bending strength of the isolated ulna is approximately 59 N-m. If we assume that the radius provides some support under dynamic bending in the region of the distal third of the forearm, the 50% risk of fracture at SAE dummy forearm moments of 67 N-m seems quite consistent with the quasistatic data. In addition, for a pronated subject arm, we expect a forearm ultimate strength to be less than the sum of the ultimate strengths of the radius and the ulna. Dynamic drop tests presented above suggest that there may be a 30% decrease in dynamic ultimate strength from impact into a pronated arm as compared with a supinated arm. If we assume that the ultimate strength of a supinated forearm is approximately the sum of the ultimate strength of the radius and ulna, we obtain a weighted average ultimate forearm bending moment of 73 N-m under quasistatic conditions. Further, if we compensate this value as above for the increase in dynamic ultimate strength and for the decrease in ultimate strength owing to arm pronation, we obtain an ultimate dynamic bending strength of approximately 76 N-m for the forearm in a pronated position. This compares well with the 50% risk SAE arm moment value of 91 N-m for forearm ulna and radius fractures. Given the nature of the approximations above, there is a rough correspondence between quasistatic bending results and the derived risk functions for the SAE forearm.

Using this simple order of magnitude analysis, it is clear that the moments measured in the RAID are far larger than expected in small female human forearms under primary contact from a deploying air bag. However, the RAID was designed as a research tool to investigate air bag aggressivity and primary contact air bag injuries. As shown above, measurements taken using the RAID under air bag deployment can be successfully correlated with both cadaver injury and more biofidelic test devices.

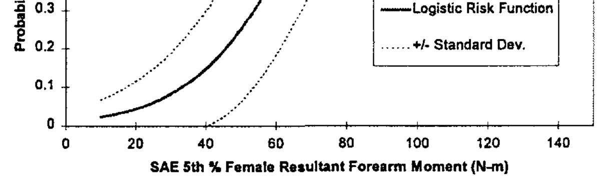


Figure 17. *SAE Arm - Risk of Ulna or Radius/Ulna Fracture.*

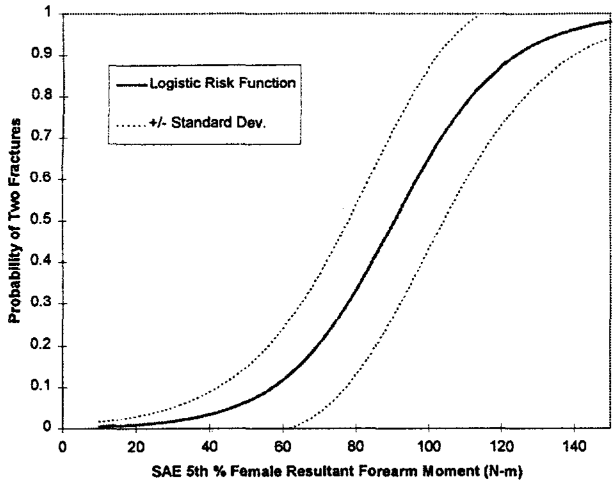


Figure 18. SAE Arm - Risk of Radius and Ulna Fracture.

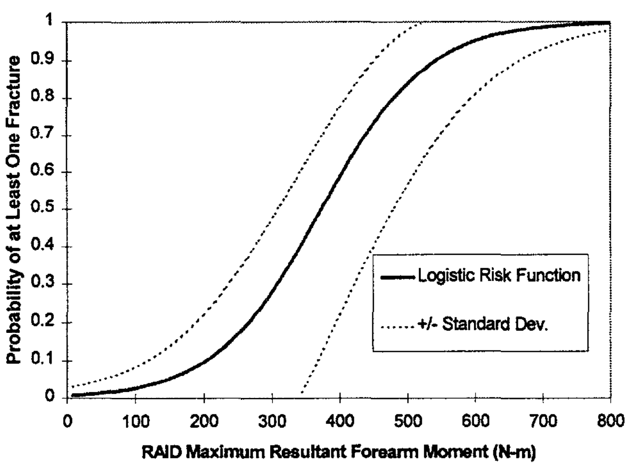


Figure 19. RAID - Risk of Ulna or Radius/Ulna Fracture.

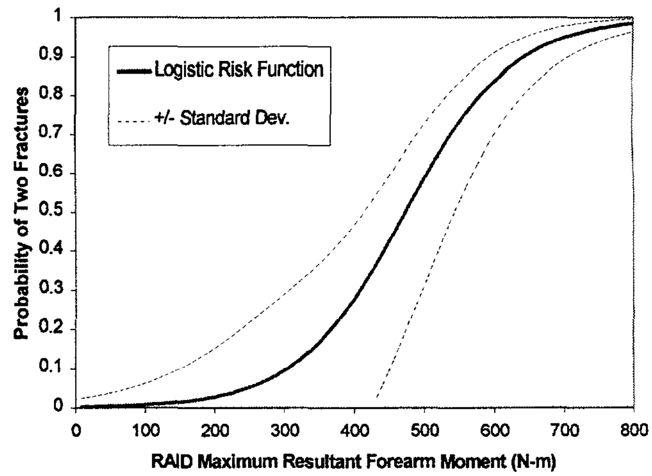


Figure 20. RAID - Risk of Radius and Ulna Fracture.

CONCLUSIONS

This study investigated the primary contact phase of air bag deployment into dummy upper extremities using four OEM air bags representative of a range of air bag aggressivities in the current automobile fleet. This aggressivity may be quantified using forearm moment response of a dummy surrogate in an appropriate worst-case position. Using this measure for primary contact injuries, this study found the System K air bag and the System H air bag to be relatively more aggressive, the System J air bag to be moderately aggressive, and the System L air bag to be less aggressive.

Maximum moments and accelerations for both test devices under air bag primary contact occur early during air bag deployment. However, peak forearm moments obtained using a System H air bag with the SAE arm occurred after the time of significant module cover/arm interaction. So, module cover interaction may not be necessary for injury with current OEM air bags.

Both the RAID and the SAE arm were found to be appropriate for examination of air bag aggressivity under primary air bag contact. Results from previous cadaveric tests suggest that primary contact injuries occur very early, before significant elbow flexion occurs. This is confirmed with moment and acceleration results from both the SAE arm and the RAID. This suggests that both devices can be successfully correlated with cadaver primary contact injury data.

There is, however, one significant potential caveat with the use of the RAID for primary contact injuries into small female occupants. As the result of a large mass and lateral moment of inertia, the kinematic response of the RAID is dramatically different from both a human forearm and the more biofidelic SAE arm. This is seen clearly in the distal acceleration response of the RAID. For all air bag systems, the acceleration was substantially smaller than that seen with either the human forearm or the SAE arm. So, the RAID will not generally be suitable for the investigation of forearm moment response of primary contact phenomena that depend on details in timing of the arm/module cover/air bag interaction. In addition, for the investigation of later phases of air bag deployment, flinging, or occupant contact, the SAE arm is more appropriate since its allowed motions are approximately anthropomorphic.

A comparison of tests using the SAE arm mounted to a Hybrid III dummy and the SAE arm mounted to a universal joint test fixture show that the use of a translationally fixed fixture has minimal effect on forearm response. So, either the SAE arm or the RAID may be used in a fixed test fixture for experimental convenience without significant effect on primary contact response.

The dummy forearm moment obtained under air bag deployment into the SAE arm and RAID correlates well with injury levels observed in cadaveric testing with the same upper extremity orientation. A logistic injury risk function was developed for small females in the 'worst-case' position using the cadaveric injuries and the dummy forearm moments. This risk function predicts a 50% risk of ulna fracture at a SAE forearm moment of 67 N-m (+/- 13 N-m standard deviation) or a RAID moment of 373 N-m (+/- 83 N-m standard deviation). The SAE arm value is consistent with an extrapolation of quasistatic ultimate bending strength of the ulna to dynamic conditions. As the result of differences in mass and moments of inertia, the moment value in the RAID is not expected to be similar to those found using a more biofidelic small female arm. In addition, we find a 50% risk of radius and ulna fracture at a SAE forearm moment value of 91 N-m (+/- 14 N-m standard deviation) that is consistent with the combined bending strength of the radius and ulna in a pronated position. A similar risk of two forearm fractures is seen with a RAID peak forearm moment of 473 N-m (+/- 60 N-m standard deviation).

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