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Airbag Interaction with Cadaveric Upper Extremities

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ABSTRACT

Although the use of airbag systems as supplemental restraints has significantly decreased the risk of fatality in automobile collisions, airbag deployments bring the potential for an increased incidence of non-fatal injuries. Epidemiological evidence suggests that the airbag system itself increases the risk of certain injuries including burns, abrasions, and eye injuries. Additionally, case studies suggest that upper extremity injuries, including severe fractures, may be caused by airbag deployment. This paper reports the results of static airbag deployments into cadaveric upper extremities: Phase I testing using four arms attached to cadaveric subjects, and Phase II testing using four arms amputated at the shoulder. For Phase II testing a four degree-of-freedom mounting system was developed to hold the arms via a three-part shoulder joint. Three driver-side airbag types were used in the study that are representative of a wide range of airbag deployment aggressivities in the current automobile fleet. High-speed video and film were used to record the deployments. The arms were instrumented with strain gauge rosettes on the radius and ulna and with magnetohydrodynamic angular rate sensors at the proximal ends of both the ulna and the humerus. Accelerometers were placed in the distal regions of the radius, proximal end of the ulna, and at the proximal end of the humerus. Deployment-induced injuries were diagnosed using pre-test and post-test x-rays as well as post-test tissue necropsy.

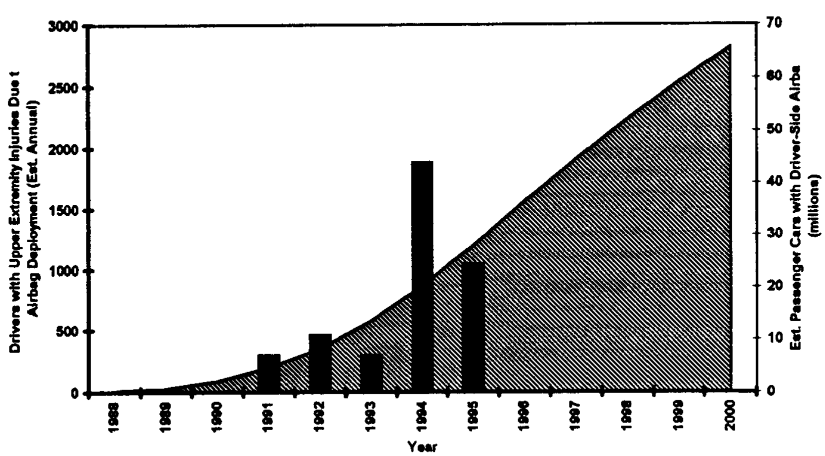
Injury results suggest that primary airbag contact can result in severe forearm injury with a sufficiently aggressive airbag deployment. In contrast, the lack of injury for a large male subject during Phase II testing with a relatively aggressive airbag deployment suggests that, even for an 'aggressive' airbag in the current fleet, arms with greater bone strength have a low risk of injury from primary airbag contact. A comparison of Phase I and Phase II results suggests that significant occupant parameters in arm/airbag interactions are upper arm position/forearm orientation and arm/bone anthropometry.

INTRODUCTION

Although the use of airbag 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 airbag deployment. In addition, case studies suggest that upper extremity injuries, including severe fractures, may be caused by airbag deployment [c.f. **Marco 1996, Freedman 1995, Huelke 1995, Kirchoff 1995**, and **Roth 1993**]. Kuppa *et al* analyzed several accident databases to determine incidence of upper extremity injury for accidents with and without a driver-side airbag 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 airbag.

Two modes of injury have been suggested to explain this increased incidence of upper extremity injuries with airbag deployment. The first type is a flinging type of injury in which the airbag propels an arm into an object in the vehicle (e.g. b-pillar, roof, occupant's head). The second type is primary contact with the airbag or airbag 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 these primary contact injuries that are the subject of the current study.

A demonstration of the current and future magnitude of the upper extremity/airbag injury problem is shown in *Figure 1*. For AIS-2 or AIS-3 upper extremity injuries, including fractures of varying severity but excluding all skin injuries, the incidence of injury rose from essentially zero in 1988 to nearly three per day in 1995. In 1996, an estimated 30% of passenger cars in the automobile fleet had driver side airbags. By the year 2000 under current regulations, this percentage climbs to 50% and increases in succeeding years. So, with no change in airbag characteristics, the incidence of upper extremity injury owing to airbag deployment is expected to increase correspondingly.



Source: NASS 1988-1995

Figure 1: Upper Extremity Injuries Owing to Driver-Side Airbag Deployment

Existing case studies of upper extremity/airbag injuries provide a strong indication of the etiology of such injuries, specifically primary contact injuries. The rather horrendous archetypal case for the current study was reported by Huelke *et al* [**Huelke 1995**]. In this case, a 75 year old woman driving a Mercury Grand Marquis executing a left turn with her right hand was involved in a collision and subsequent airbag deployment. She suffered multiple segmental fractures of the right radius and ulna, including the proximal and distal ulna. In addition, the driver also sustained a circumferential degloving laceration involving both skin and subcutaneous tissue. Blood spatter patterns in the interior of the vehicle suggested that the hard tissue injuries occurred as the result of primary contact with the airbag.

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/airbag module, 2) Women experience an age-related loss of bone mineral density, and 3) Women have generally smaller bones and, hence, lower bone strength.

To investigate the upper extremity/airbag interactions causing these injuries, Saul *et al* used an instrumented 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 airbag's inflator properties, flap, and steering wheel orientation.

In addition, the Research Arm Injury Device (RAID) was developed by Conrad Technologies Inc. and NHTSA to investigate the interaction between a deploying airbag and an upper extremity in close proximity to the airbag [**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 airbag module and the separation distance between the two. Maximum moments were recorded when the forearm was positioned perpendicular to the airbag module. This situation occurs, for example, when making a left turn with the right hand. In this situation the sides of the airbag are at the 1 and 7 o'clock position, 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 airbag and the forearm was increased from 1.3 cm to 7.6 cm.

It is likely that a specific airbag 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 airbags of different deployment properties (e.g. leading edge speeds) between vehicle models. Three airbag types were used in this study; these airbags were identified using RAID testing as representing a wide range of aggressivities in the current passenger car fleet. Using a National Highway Traffic Safety Administration (NHTSA) coding scheme, these systems are termed System H, System K, and System L airbags. The System H and System K airbags produce relatively more aggressive airbag deployments, and the System L relatively less aggressive deployments. In addition, the System H airbag has been identified in case studies as producing primary contact upper extremity injuries under certain circumstances.

The principal goal of this study is to determine if these primary contact injuries can occur and if they can be reproduced in a laboratory setting. This study has two phases of testing denoted Phase I and Phase II. In Phase I testing, airbags were deployed into arms attached to female cadaveric subjects. The rest of the body was used for concurrent studies. These Phase I tests were run with limited instrumentation. In Phase II testing, excised cadaveric arms attached to a four degree-of -freedom universal joint acting as a shoulder are used. The Phase II tests were more extensively instrumented using load cells, strain gauges, accelerometers and magnetohydrodynamic angular rate sensors. This paper begins with a short upper extremity anatomy review, proceeds to descriptions of the Phase I testing, Phase II testing and follows with conclusions and observations regarding the research.

ANATOMY REVIEW

The foundation of the upper extremities is the skeletal structure consisting of 64 bones. All of these are long bones with the exception of the shoulder girdle. *Figure 2* details the major bones of the upper extremity. The skeleton of the shoulder girdle consists of the clavicle and the scapula. The shoulder cavity holds the primary joint of the structure. This joint is a ball-and-socket joint which consists of the rounded head of the upper arm and a curved plate within the shoulder cavity. The curved plate does not surround the head of this joint which allows the upper arm the greatest mobility. Motion of the shoulder joint transpires in three planes known as the sagittal, frontal, and horizontal planes. The shoulder joint, acting as the origin of these three planes, allows the extended arm three degrees of freedom. The general motion is portrayed in *Figure 3*.

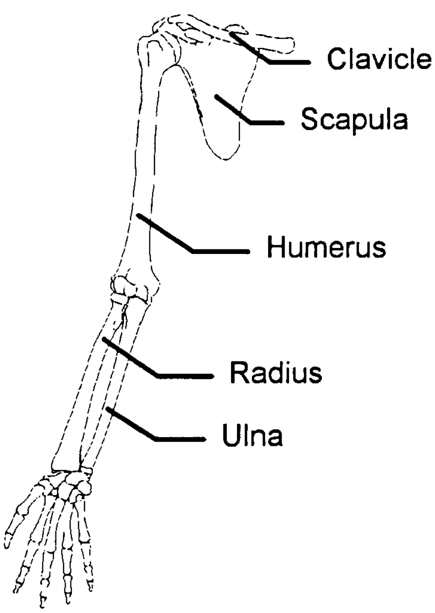


Figure 2: Anterior Aspect of Bones in the Right Upper Extremity

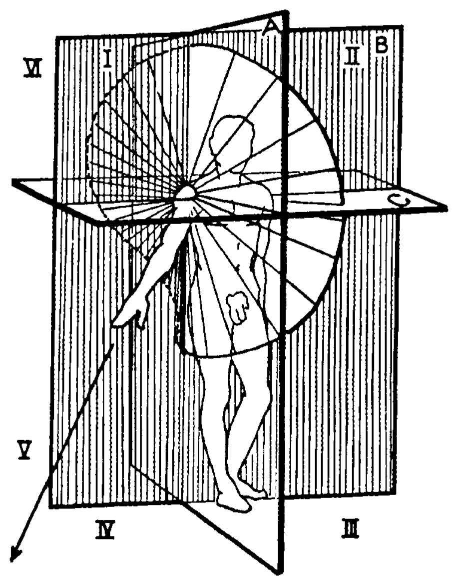


Figure 3: Range of Motion for the Upper Extremities in Three Planes: A=Sagittal, B=Frontal, C=Horizontal

The rounded head of the shoulder joint is the head of the upper arm known as the humerus. This long bone is cylindrical at its top and triangular at its distal head. Grooves and ridges in its structure enable muscles to attach to its surface. Two components of the elbow joint lie within the distal humerus head. The first of these is the trochlea which acts as a pulley. It is shaped as a cylindrical bar whose edges slope inward to a central groove. The second component is a hemispherical ball known as the capitulum which is attached to the lateral edge of the trochlea. The forearm consists of two bones known as the radius and the ulna. The trochlea connects with the larger head of the ulna while the capitulum joins with the radial head. The elbow joint has less freedom of motion than the shoulder joint. Single planar movement of the elbow joint allows for two principal ranges of motion which include flexion and extension. The maximum angle of motion during active flexion is approximately 145°. Relaxation of the muscles during passive flexion increases the range of motion to approximately 160°.

The base of the forearm connects to the wrist, the most complex jointed region in the upper extremities. The wrist contains several joints in order to increase the mobility of the hand. Its eight carpal bones, connected by fifteen muscles, allow for a range of motions including flexion, extension, adduction, abduction, and circumduction.

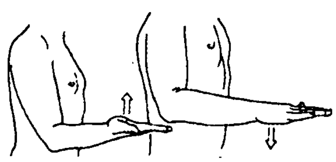


Figure 4: Supination (left) and Pronation (right)

Both the jointed regions at the wrist and at the elbow enable two major functions known as pronation and supination as shown in *Figure 4*. These regions, the superior radio-ulnar joint at the elbow and the inferior radio-ulnar joint at the wrist, allow for rotation along the longitudinal axis of the forearm. The radius and ulna are complementary given that the proximal ulna head and distal radius head are both nearly equal in size, but at opposite ends of the forearm. This geometry aids the rotation of the forearm. The reference position for pronation and supination is similar to that of lateral and medial rotation with the forearm perpendicular to the upper arm with a neutral positioning of the hand. Pronation and supination can then be measured from the movement of the palm from its neutral point.

PHASE I TESTING

Phase I testing investigated the static deployment of an airbag into an arm resting across the airbag perpendicular to the tear seam. The Phase I test matrix is shown in *Table 1*. A total of four arms from two female embalmed subjects were used: one a fifth percentile female subject. The subjects were embalmed with a fluid designed to maximize subject biofidelity [**Crandall 1994**]. In addition, two airbag types were used in the Phase I testing, the System H airbag and the System L airbag. The System H airbag is representative of a more aggressive airbag in the current fleet; the System L airbag is representative of a less aggressive airbag. The airbags were mounted in original equipment steering wheels appropriate to the airbag tested. As the investigation of injury behavior was the primary goal of Phase I testing, the tests used limited instrumentation consisting of a single five-axis steering column load cell.

The occupant was seated beside the steering wheel as shown in *Figure 5*. For each test, the fingers were taped to the rim of the steering wheel to place the airbag flap seam in the distal third of the forearm. The distance from the steering wheel to the proximal forearm, 0.6 cm, was determined using RAID test results to obtain maximum forearm moments. Each forearm was oriented in approximately a neutral position. Padding was used in the back of the test fixture and along the face and torso of the subject to minimize the risk of subsequent contact injuries.

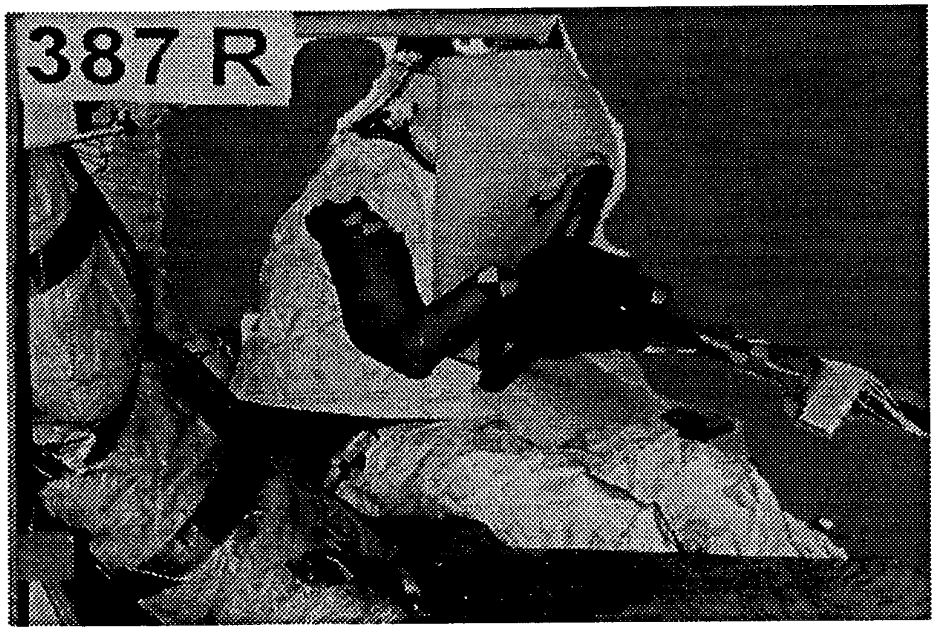


Figure 5: Phase I Test Setup

Phase I steering column loads seen in a System H airbag deployment are compared with an System L airbag deployment in *Figure 6*. As expected, the axial column reaction seen below the steering wheel is far larger for the System H airbag than for the System L airbag. Typical peak values of the axial load are approximately 5500 N for the System H airbag and 3200 N for the System L airbag. In addition, the System H airbag produces axial force peaks that are more than 6 ms sooner than for the System L airbag, and the System H airbag has a significantly greater onset rate. This strengthens the assertion that the 1991 System H airbag has a more aggressive deployment than the System L airbag.

The System H airbag shows much greater extremes of steering column Y bending moment (bending about a horizontal line affixed to the face of the airbag) than the System L airbag. A typical maximum value for Y bending moment in a System H airbag deployment is over 200 N-m while the System L airbag deployment produces less than 100 N-m. As seen in the steering column axial load, the peak bending moment occurs later for the static deployment of the System L airbag than for the System H airbag. As with the axial force, the steering column bending moment seen in the System L airbag deployment has a slower onset rate than that seen in the System H airbag deployment.

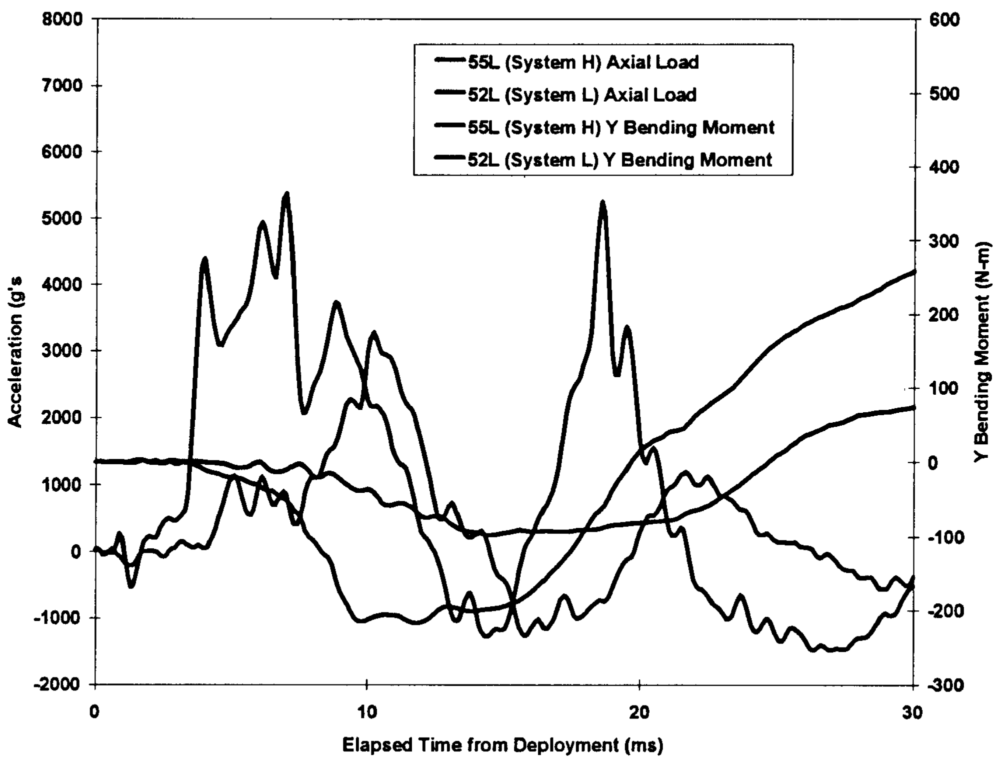


Figure 6: Phase I Steering Column Load Cell Data - System H Airbag and System L Airbag

As seen in *Table 1*, no forearm injuries were produced in Phase I testing that were attributable to primary airbag contact even though there were significant differences in steering column load between the two airbags tested. The sole hard tissue injury was a torsional humerus fracture that was attributed to moments resulting from the airbag flinging the forearm rearward.

An analysis of the Phase I results suggests two possibilities. The first possibility is that neither of the airbags tested are sufficiently aggressive to produce forearm injuries in a normal driving posture. This appears unlikely owing to the identification of the System H airbag in several case studies of upper extremity primary contact injuries. The second possibility is that there are differences between the Phase I test conditions and the driving configurations seen in case studies with hard tissue injuries. Owing to the need to protect the subject thoraxes for concurrent testing, there are two significant differences between the Phase I test setup and that appropriate to a 'natural' driving condition. The first is the position of humerus, oriented to the side of the steering wheel. In a natural driving posture, the humerus is oriented approximately normal to the plane of the steering wheel. The second is that in the Phase I tests, the forearm is oriented neutrally or slightly supinated. In an ordinary crossover maneuver during a turn, the arm used would be substantially pronated. In Phase II testing with excised arms, these differences are eliminated.

Subject	Airbag	Sex	Age	Injury
59L	System L	Female	45	No FX
59R	System H	Female	45	No FX
58L	System H	Female	61	Spiral Humerus FX
58R	System L	Female	61	No FX

Table 1: Injury Results from Phase I Testing

PHASE II TESTING - EXCISED ARMS

To obtain a more 'natural' driving position for the arm than that in Phase I testing, Phase II testing used excised cadaveric arms to avoid potential interactions with the subject torso. Subject arms were attached to a four degree-of-freedom universal joint, as diagrammed in *Figure 7* and *Figure 8*, providing essentially anthropomorphic joint motion. In the test position, a 'natural' driving posture in a one-hand turn crossover maneuver, the forearm was pronated with the humerus normal to the plane of the steering wheel. The fingers were taped on the steering wheel rim with the arm positioned so that the distal third of the forearm was over the airbag tear seam. The case studies referenced above suggest that most fractures occur at the distal third in both the humerus and radius. Four arms, three male and one female, were used from three cadaveric subjects that were embalmed with a fluid designed to maximize subject biofidelity [**Crandall 1994**].

One possible objection to the Phase II 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 9*. 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 airbag. 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 Phase II experimental setup.

The instrumentation locations for Phase II experimentation are shown in *Figure 10*. Two strain gauge rosettes are located on the ulna and two strain gauge rosettes on the radius to measure local strain on the bone surface. These gauges may be used to derive strain rates and local moments during the test. A humerus load cell was mounted at the interface between the excised arm and the four degree-of-freedom universal joint at the shoulder. Magnetohydrodynamic (MHD) angular rate sensors were used to measure the local rotation of the upper extremity at several locations. Uniaxial MHD angular rate sensors were mounted on the distal radius and the proximal ulna; the distal radius angular rate sensor was used to determine the relative position of radius and ulna, and the proximal ulna sensor was used to measure the rotation of the forearm relative to the humerus. A triaxial MHD angular rate sensor was mounted to the humerus load cell to measure the rotation of the humerus relative to the fixed laboratory reference frame. In addition, a triaxial accelerometer was mounted to each MHD angular rate sensor.

Each Phase II subject arm was strain gauged on the midshaft medial and anterior ulna bone surfaces and the midshaft posterior and lateral radius bone surfaces using standard surface preparation and mounting techniques. Incisions were made to minimize tissue injury and interference with the interosseous membrane. Nondestructive three point static bending tests were performed on each arm to provide physical properties for later use in deriving dynamic moment data under airbag deployment. Multiple strain gauge failures that occurred during dynamic testing and handling prevented the derivation of moments from the strain gauge data for Phase II tests.

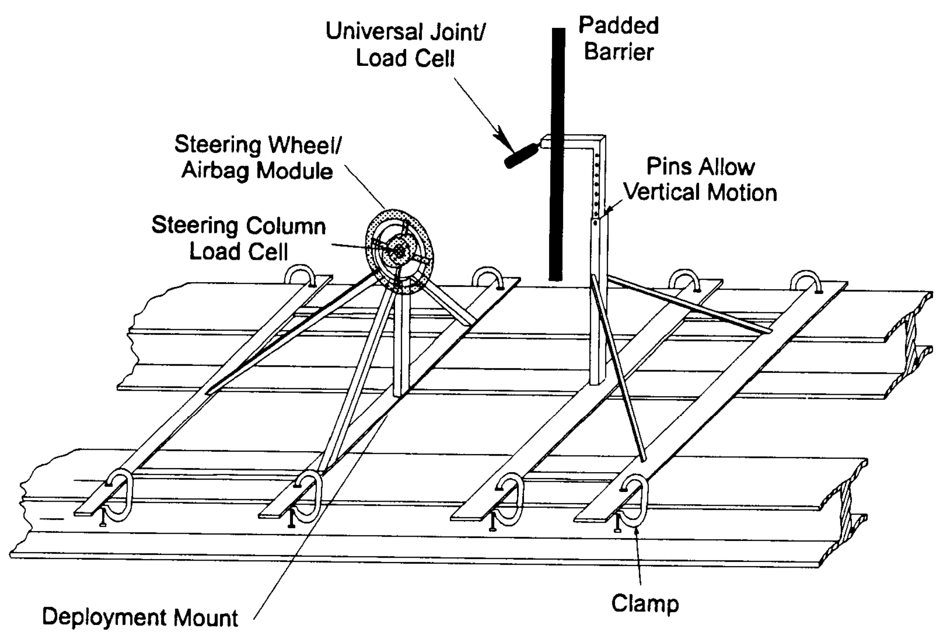


Figure 7: Phase II Test Schematic Diagram

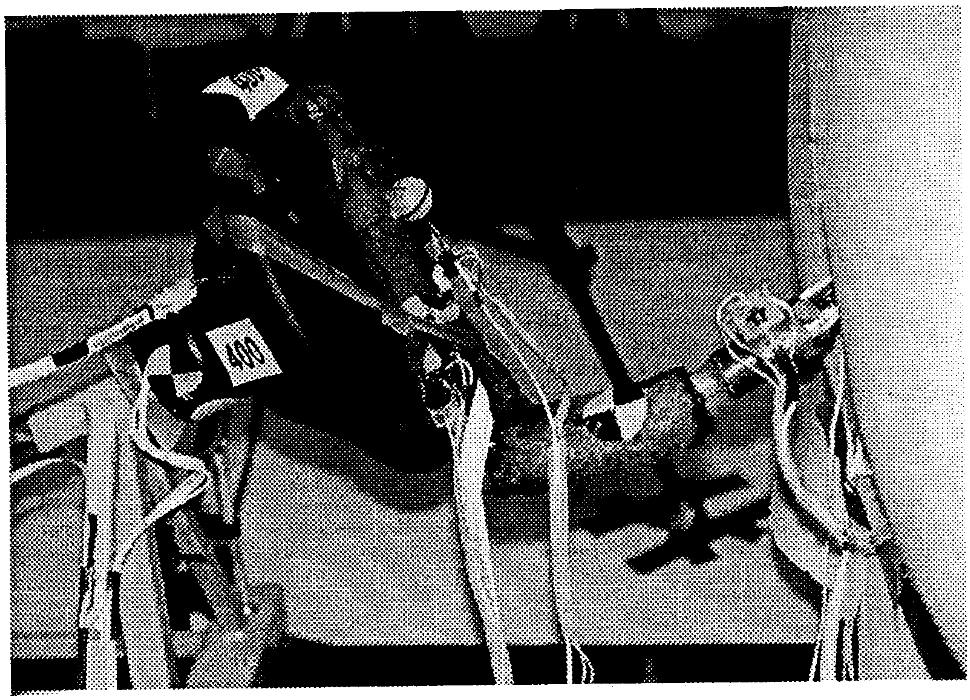


Figure 8: Phase II Test Fixture

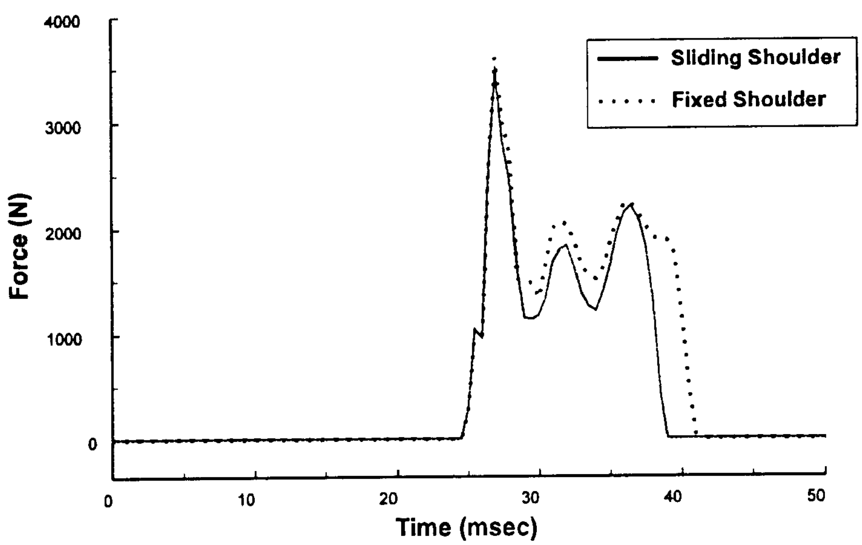


Figure 9: ATB Simulation of Fixed vs. Sliding Shoulder

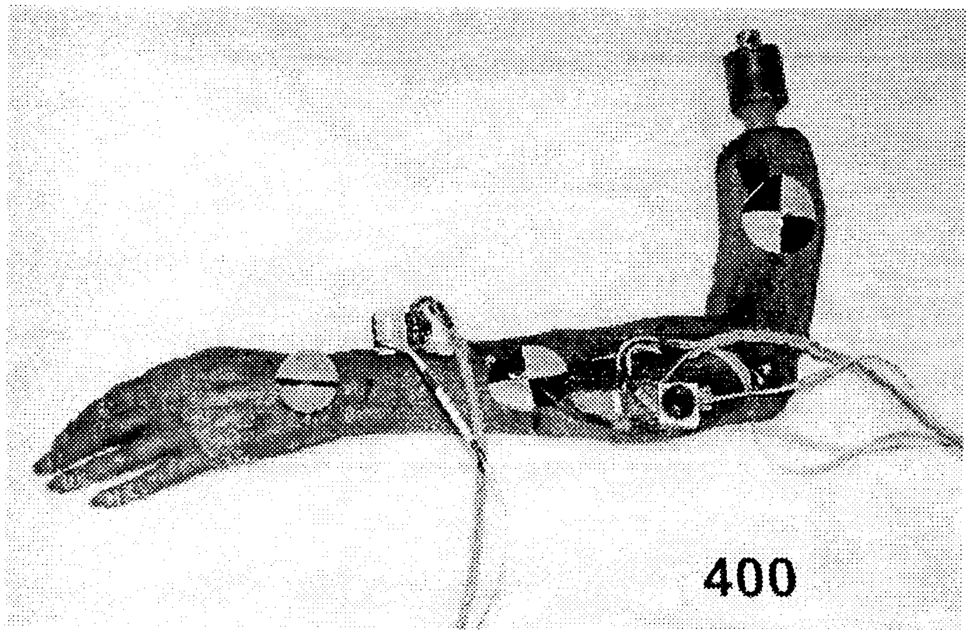


Figure 10: Instrumentation Locations for Phase II Testing

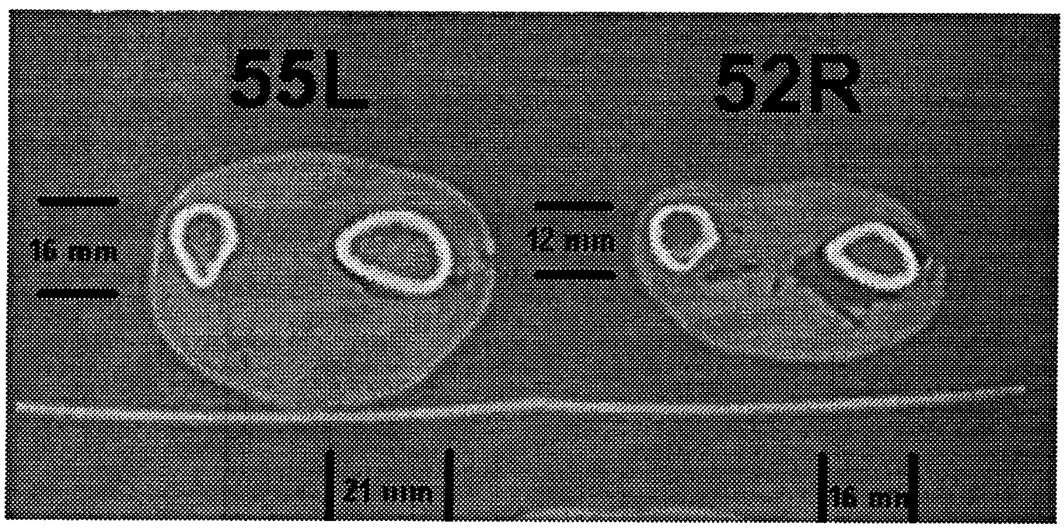


Figure 11: Comparison of Bone Cross Sections - Subjects 55L and 52R

Injuries sustained by the excised arms for Phase II dynamic tests are shown in *Table 2*. These injuries were identified using post-test necropsy and radiological examination. Injuries owing to primary contact with the airbag included comminuted radius and ulna fractures for the relatively aggressive System K airbag with a female subject and a comminuted ulna 'nightstick' fracture for the relatively aggressive System H airbag with the small male subject 52R. The metacarpal and humerus fractures suffered by subject 63L were artifacts of the arm striking the test apparatus padded barrier after interacting with the airbag. As expected, the less aggressive System L airbag produced no fracture in subject 52L. Interestingly, the System H airbag deployment produced no injury in subject 55L, a large male subject, though injuries were seen in the System H airbag test with subject 52R, a small male subject. This suggests bone geometry and bone strength plays an essential role in the risk of upper extremity injury from a deploying airbag.

A comparison of the distal third bone cross-sections of subjects 55L and 52R is shown in *Figure 11* that was derived from a slice of a Computerized Tomography (CT) scan of each arm. The slice for 52R has been mirrored for ease of comparison. Though the bone thickness and bone density are comparable for the two subjects, the moments of inertia of the radius and ulna for subject 55L are much larger than those for subject 52R. This suggests that a portion of the population with sufficiently strong and large bones is at relatively low risk of primary contact airbag-induced upper extremity injury even for a relatively aggressive airbag deployment.

The injuries sustained by subjects 63L and 52R occurred soon after airbag deployment. Fractures were identified using strain gauge data and high speed film. All primary contact injuries occurred within 5 ms of seam tear, during the period of interaction with the airbag flap. This suggests that the airbag flap may play an important role in the production of primary contact injuries.

Subject	Airbag	Sex	Age	Injury
63L	System K	Female	61	Comminuted Ulna FX Comminuted Radius FX 2nd Metacarpal FX Humerus FX
52R	System H	Male	64	Comminuted Ulna FX
52L	System L	Male	64	No FX
55L	System H	Male	48	No FX

Table 2: Injury Results from Phase II Testing

The maximum strain rate derived from strain gauge data from Phase II testing is shown in *Table 3*. The System K and System H strain rates for tests 63L and 52R respectively were higher than in the subsequent two tests, capturing the yield region immediately prior to fracture. The strain rate results suggest a microfracture of the radius of subject 52R even though no fracture was seen in the post test analysis. Subject 52L (System L airbag) saw very low strain rates as expected. As subject 55L did not suffer any fractures, the test produced low strain rates compared to the previous System H deployment into subject 52R.

Test	Airbag	Max Strain Rate (1/s)	Gauge
63L	System K	285	Radius Lateral 2
52R	System H	115	Radius Lateral 3
52L	System L	2.4	Ulna Posterior 2
55L	System H	4.3	Ulna Anterior 1

Table 3: Phase II Strain Rates

Peak acceleration at the distal radius for subject 52L (System L airbag) was 316 g's at 44.5 ms. For subject 55L (System H airbag), the peak acceleration was 451 g's at 6.7 ms. It is interesting to note that, as with the column force reported for Phase I testing, the System H airbag showed a nearly 50% larger peak acceleration than the System L airbag, and the peak occurred much earlier with the System H airbag. Gauge failures prevented determination of accelerations from other Phase II tests.

The peak humerus axial load is shown in *Table 4*. Relatively consistent axial force peaks are seen among subjects 63L, 52R, and 55L (System H and System K airbags) with early force peaks at approximately 11-15 ms. In contrast, the relatively less aggressive System L airbag used for subject 52L showed less than half the peak humerus axial force at time 49.5 ms, significantly later than the more aggressive airbags. This trend is also seen in the time history of humerus axial load compared for subject 52L (System L airbag) and subject 55L (System H airbag) in *Figure 12*. The humerus axial load for subject 55L peaks substantially sooner and with significantly greater onset rate than the humerus axial load for subject 52L.

Both MHD angular rate sensors showed in excess of 66 radians per second on the forearm. These rates, much higher than expected, clipped both sensors for all tests.

Subject	Airbag	Max. Humerus Axial Load (N)	Max. Time (ms)
63L	System K	2030	15.5
52R	System H	2060	11.8
52L	System L	880	49.5
55L	System H	1530	11.6

Table 4: Phase II Humerus Axial Load

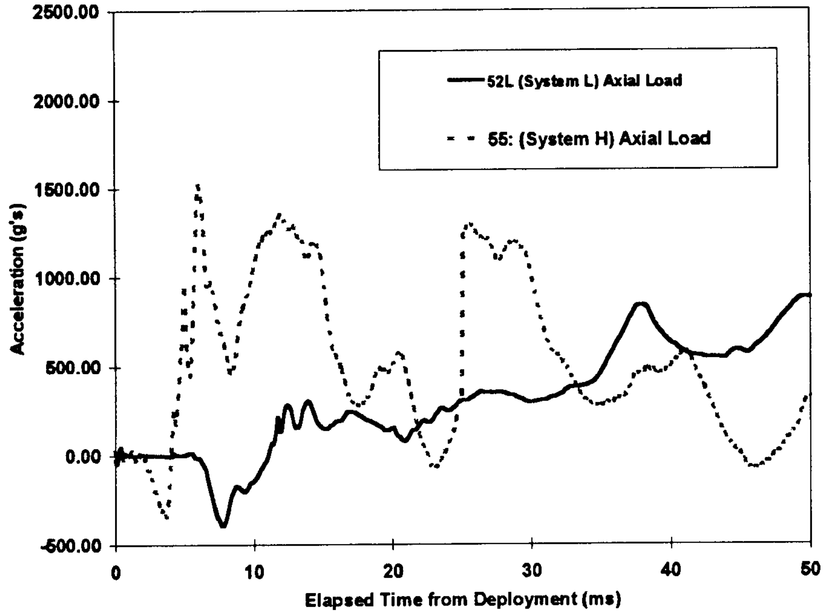


Figure 12: Phase II Humerus Axial Load

CONCLUSION

This study investigated the interaction of upper extremities with deploying airbags in two configurations. In Phase I, airbags representative of a wide range of deployment 'aggressivity' were deployed into cadaveric upper extremities attached to cadaveric subjects. In Phase II testing, excised cadaveric upper extremities were mounted to a test fixture that models the motion of the human shoulder. No injuries were produced in Phase I testing, even for a relatively aggressive airbag deployment. This result is attributed to differences in arm orientation between the Phase I test setup and a 'natural' driving posture. These differences were eliminated in Phase II testing, and forearm injuries were sustained very early during arm/airbag interaction for both female and small male subjects. This result suggests that primary airbag contact can result in severe forearm injury for a sufficiently aggressive airbag deployment. In addition, the testing suggests that airbag deployment aggressivity can contribute to occupant upper extremity injury. In contrast, the lack of injury for a large male subject during Phase II with a relatively aggressive airbag deployment suggests that, even for an 'aggressive' airbag in the current fleet, arms with greater bone strength have a low risk of injury from primary airbag contact. A comparison of Phase I and Phase II results suggests that significant occupant parameters in arm/airbag interactions are upper arm position/forearm orientation and arm/bone anthropometry.

Future work will be performed to elaborate on these results. Drop testing will be performed to determine the effect of forearm pronation vs. supination on the peak load to failure to separate effects of forearm orientation from the effects of humerus position on the characteristics of upper extremity/airbag interactions. Additional airbag deployments into excised subject upper extremities will be done for a test matrix of airbags to explore a wider range of airbag aggressivities and to develop a risk function for upper extremity primary contact airbag injuries.

ACKNOWLEDGEMENTS

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REFERENCES

[Crandall 1994] J.R. Crandall, *Preservation of Human Surrogates for Biomechanical Studies*, PhD Dissertation, University of Virginia, Charlottesville, Virginia, January, 1994.

[Freedman 1995] E.L. Freedman, M.R. Safran, and R.A. Meals, Automotive Airbag-Related Upper Extremity Injuries: A Report of Three Cases, *Journal of Trauma*, 38:577, 1995.

[Huelke 1995] D.F. Huelke, J.L. Moore, T.W. Compton, J. Samuels, and R. Levine, Upper Extremity Injuries Related to Airbag Deployments, *Journal of Trauma*, 38:482, 1995.

[Kirchoff 1995] R. Kirchhoff, and S.W. Rasmussen, Forearm Fracture Due to the Release of an Automobile Airbag, *Acta Orthopaedica Scandinavica*, 66:483, 1995.

[Kuppa 1997] S.M. Kuppa, C.W. Yeiser, M.B. Oslon, L. Taylor, R. Morgan, and R. Eppinger, RAID - An Investigation Tool to Study Airbag/Upper Extremity Interactions. SAE Paper (to be published), SAE International Congress and Exposition, Detroit, MI, 1997.

[Marco 1996] F. Marco, A. Garcia-Lopez, C. Leon, and L. Lopez-Duran, Bilateral Smith Fracture of the Radius Caused by Airbag Deployment, *Journal of Trauma*, 40:663, 1996.

[Roth 1993] T. Roth, and P. Meredith, Hand Injuries From Inflation of an Airbag Security System, *Journal of Hand Surgery*, 18B:520, 1993

[Smock 1995] W.S. Smock, and G.R. Nichols, Airbag Module Cover Injuries, *Journal of Trauma*, 38:489, 1995.

[Saul 1996] R.A. Saul, S.H. Backaitis, M.S. Beebe, and L. Ore, *Hybrid III Dummy Instrumentation and Assessment of Arm Injuries During Air Bag Deployment*, SAE Paper 962417, 40th Stapp Car Crash Conference, Albuquerque, New Mexico, 1996.

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DISCUSSION

PAPER: **Air Bag Interaction with Cadaveric Upper Extremities**

PRESENTER: Dale Bass, University of Virginia

QUESTION: John States, University of Rochester

Did you measure the ash content or do something to measure the extent of osteoporosis in your cadaver specimens? This makes a difference of maybe a factor of five in respect to strength.

ANSWER: Right. We are doing the CT studies, and will extract the bone mineral content from the CT studies. We haven't done that yet.

Q: That's great. Thank you.

Q: Guy Nusholtz, Chrysler Corporation

When I looked at your curves that you kept on talking about force, all I saw in your axis was acceleration.

A: Yes. I apologize. That was force not acceleration.

Q: What was that in?

A: Newtons.

Q: Newtons. So you were getting around 5,000 Newtons for the system K and somewhere around 1,000 Newtons for the system L. Is that correct?

A: It was a couple of thousand Newtons for the system H and maybe 800 for the system L.

Q: That's a load cell?

A: That's right. It's a load cell. We've been previously using these as tibia load cells but now we just put them in between the U joint and the humerus. So it's just a load cell.

Q: How do you sort out the forces that are going to come from the airbag deploying?

A: Sort out the forces on the forearm as opposed to forces on the humerus.

Q: Well, you instrument a steering column with a load cell so as the airbag deploys it is going to generate a force which is going to drive it down anywhere from 100 to 500 lbs. or anywhere from 500 Newtons to 2,000 Newtons.

A: Right.

Q: And then you've got a force on the arm so if the load cell is in the steering wheel, how do you sort those different forces out?

A: Sort those forces out as regards to their effect on the forearm? We haven't yet. Our idea is that what we will do is get an idea of the forces in the forearm from our strain gaging procedure rather than from the steering wheel load cell. It is a horrible thing to do. From just our system L tests, the variation in the steering column axial load, for instance, it seems very difficult to sort out the dynamics on the other side from the dynamics underneath the steering wheel. So, hopefully what we will be doing is measuring them directly on the other side. Well, as directly as we can through strain gages on bone.

Q: Have you compared the pathology and the geometry and the positions of the injuries that you are getting in those tests with the ones that are occurring in the field?

A: No, we haven't done that.

Q: Another question. Have you given any consideration to the effect of the tendons and the muscle structure as to affecting the forces that are going to operate on the arm?

A: Yes. That is the reason we are using the strain gages. That's the reason we are trying to be as minimally invasive as possible. On the other hand, we think the fractures are occurring very early. There may be very little time for other musculature and the tendon structure to appear. For instance, for it to affect what is happening in the forearm.

Q: Reaction is one thing and it is very quick. You're over in about four or five milliseconds. But what about pre-tensioning or pre-conditioning?

A: The only pre-tensioning study we've done is the simulation that I mentioned, grip strength. The sensitivity of the contact force to grip strength and we found it wasn't very sensitive.

Q: OK. Thank you.

Q: Larry Schneider, UMTRI

I have two questions. One has to do with this issue of the strain gages which you indicated using, but I don't think you talked about any results and my question would be how would you get quantitative useful results out of those strain gages?

A: We're working on a calibration procedure. We've had difficulty with the strain gages. In the beginning we were losing nearly 100% out of them before the end of the test. In this testing that I'm showing you, we were losing about 50% of them before we got to the end of the test. Currently, we hope we're losing (testing is actually going on right now) and we hope we're losing far fewer. We've had wires ripped out. Heating has been a very big problem for us.

Q: How do you intend to calibrate them?

A: What we've been doing is putting them in a force testing device prior to the test and forcing them in different orthogonal directions and then we are planning to back out the numbers from that kind of calibration. On the other hand, we have very little dynamic data now to try that with. So, we're not sure if that's going to work well or not. We think it will.

Q: Another question. I may have known the answer to this at one time but I don't remember. You went from the intact arm to excised arm; was this something you had to do because you didn't get fractures?

A: Well, getting full cadavers is difficult and when we get full cadavers, we do certain kinds of tests with them. Getting excised arms is difficult but it is less difficult.

Q: OK. So, it wasn't because you weren't getting the injuries in the intact condition.

A: It was that we couldn't put the full cadavers into the appropriate driving position because we need to use the thorax for other research.

Q: Jeff Pike, Ford Motor Company

Another aspect of the older driver is that they tend to be over-represented in having crashes while they are making left hand turns. So, it looks like you might have several factors all adding to each other.

A: Yes. I think that's probably true.

Q: Thank you, Dale.

A: Thank you.