ASSESSMENT OF FOREARM INJURY DUE TO A DEPLOYING DRIVER-SIDE AIR BAG

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INTRODUCTION

Since their introduction, air bags have been shown to save lives and reduce the risk of injury. [1] Early reports stated that injuries related to deployment of an air bag were minor and infrequent. [2] As the number of cars equipped with air bags increases, more data is becoming available and more reports of injuries caused by air bag deployment are appearing in the literature. Air bags have been designed to fully inflate before the occupant contacts it, typically within 50 milliseconds of vehicle impact.[2] Occupants who are either out of position or in close proximity to the air bag module at the time of deployment have sustained severe injuries. Because of growing concern with regard to the position of drivers in relation to the air bag module, it is important that the mechanisms of injury associated with air bag deployment be assessed.

A Canadian study of crashes in which an air bag deployed found that the upper limb is the most frequent site of injury to drivers. [3] While seldom life threatening, upper limb injuries can cause significant disability. Injury to the median nerve, tendon rupture, vascular injuries and other soft tissue damage may occur as a complication of fracture or as a separate event. Long term complications can include arthritis and joint instability. [4] Friedman et al., Huelke et al., and Smock et al. [5, 6, 7] describe upper limb injuries, including abrasions lacerations, contusions and burns, which they attribute to deployment of a driver's side air bag. They also report fractures of the humerus, radius, ulna, and metacarpals.

Taylor et al. conducted a search of the National Automotive Sampling System (NASS) data base. They reviewed 65 cases in which an air bag deployed and an upper limb injury of greater than or equal to level 2 on the Abbreviated Injury Scale (AIS 2) was reported. [8] In studying this data, they found that drivers restrained by a three point belt and an air bag had four times the incidence of upper limb injuries as did drivers using a three point belt alone. In addition, they identified three mechanisms by which an air bag can cause an upper limb injury. All of these mechanisms require that the hand or forearm be in close proximity to the air bag module when it deploys. One is a result of direct interaction between the forearm and the deploying air bag. The other two involve the forearm being flung upward or laterally and contacting the interior of the vehicle. These injuries were found in crashes where there was a change in velocity of less than 15 mph, as well as in high speed collisions.

Taylor et al. used the Research Arm Injury Device (RAID) to measure forces and moments generated by a deploying air bag impacting a driver's forearm. [8] The RAID is an aluminum tube with a joint at one end allowing rotation about two axes. At the other end is a concentrated mass representing the hand. The tube is instrumented with strain gauges and accelerometers. Using this device, they have measured moments far in excess of previously reported values of human tolerance. Although the RAID measurements were high, Bass, et.al., have shown that the measurements correlate to cadaver injuries under similar test conditions.[9]

Researchers at the National Highway Traffic Safety Administration (NHTSA) used an instrumented dummy to quantify the forces generated by a deploying air bag impacting a driver's forearm.[10,11] The forearm was positioned over different air bag module systems in various configurations and the loading during static deployment was measured. This data demonstrated the same trends as the RAID data, but higher accelerations and lower moments and wrist velocities were measured with the dummy. Both the RAID and the dummy were able to differentiate between aggressive and less aggressive air bags.

The question of how much force the forearm can withstand was addressed by Pintar and Yoganandan [12] using dynamic three-point bending tests. Fresh cadaver forearms were placed on simple supports resting on load cells with the supports contacting both the radius and the ulna. Each forearm was tested once with an impactor speed of either 3.3 m/s or 7.6 m/s. Fractures were produced on all

specimens. The higher speed impacts produced more comminuted and more distal fractures. They suggested that a person with a lower forearm mass may have a decreased failure bending moment.

Researchers at the University of Virginia have addressed the forces generated by a deploying air bag impacting a driver's arm. They conducted three series of tests using human cadaver upper limbs disarticulated at the shoulder, some frozen and some embalmed. [13] In the first series, forces and bending moment in the steering wheel and humerus were the only data gathered. Forearm acceleration was added in the second series of tests. In the third series, strain gauges were added to assess forearm bending moments. They concluded that severe injury may result from contact with a deploying air bag but that increased bone strength may decrease the risk of injury.

At the University of Michigan Transportation Research Institute, researchers used unembalmed cadavers to investigate forearm interaction with a deploying drivers side air bag.[14] They varied the spacing between the forearm and the air bag module and found that increased spacing between the two greatly reduced the incidence of fracture. They also determined values for peak forearm velocity and bone mineral content that separated incidents of fracture and no fracture.

This study has examined forearm injury patterns produced by static deployment of a driver's side air bag. Attention was focused on the loadings that occur during the punchout phase of deployment because it has already been determined that the highest loadings occur during punchout[15] Loading patterns and the correlation of loadings and injuries were examined with the expectation that fractures would occur during punchout and that bending would be the mechanism of failure.

METHODS

The test fixture used was a rigid seat padded with stiff foam. The seat and seat back angles were fixed, but the seat could be moved forward or backward. The steering column was rigid but the height and angle could be adjusted. The supporting framework behind the seat was also padded with foam to protect the forearm from further injury following loading by the air bag. Two different air bags were chosen in conjunction with researchers in other laboratories. One is considered an aggressive air bag and the other is considered less aggressive. The designations "aggressive" and "less aggressive" were based on the tank pressures and the incidence of injury from field data. [8] The aggressive system is coded H-91 and has an inflater output of 350 kPa x 22 kPa/msec for a 28L tank. The less aggressive system is coded L-92 and has an inflater output of 319 x 12.

Paired tests were conducted with each subject so that an H-91 system was used on one forearm and an L-92 system was used on the other. Prior to the first test, a matrix was prepared which alternated whether each air bag was used on the right or left limb. Subjects were placed in the matrix in the order that they became available.

Four adult cadavers, 3 females and 1 male were used. They ranged in age from 68 to 89 years. Although the subject ages tended to be somewhat advanced, it was determined that these subjects were suitable for assessing arm injury due to air bag deployment since the Bone Mineral Density was typical for post-menopausal women. All cadavers were unembalmed and all instrumentation and testing was done within 60 hours of death. Prior to instrumentation and testing, the upper limbs were palpated and exercised through their range of motion to eliminate rigor mortis and to check for evidence of previous injury. Both forearms of each cadaver were placed in supination and the pre-test X-rays were taken. Anthropometric measurements of the upper limbs for each subject were recorded. The radius and ulna lengths were measured from the x-rays. Post-test x-rays of both forearms for each subject were taken immediately after testing. A summary of these measurements is recorded in Table 1.

Planar strain gauge rosettes were applied to the radius and ulna in order to qualify the strain patterns at or near any distal fracture sites. Since the majority of forearm fractures occur in the distal third of the radius, the site selected was just proximal to the distal metaphysis. This site maximized the possibility of measuring strains at or near the fracture site and minimized interference with other instrumentation. The posterior aspect of the forearm was chosen to minimize the possibility of destroying or dislodging the gauges on impact. A three gauge rectangular rosette, Micro-Measurement CEA-06-062UR-350, was chosen to define the principal strains.

The distal radius and ulna were exposed and a site on the posterior aspect of the bone was selected for gauge application as shown in Figure 1. Although the preferred site was the posterior aspect, a more medial or lateral site was necessary to provide a relatively flat area large enough to mount the gauge on some of the smaller bones. The periosteum was removed from the bone at the selected site. The bone was abraded with fine sandpaper, then degreased and dried with isopropyl alcohol. A catalyst was applied to the back of the gauge and it was bonded with cyanoacrylate. (M-bond 200) A few additional drops of cyanoacrylate were placed at the wiring solder joints for mechanical protection. A protective coating of microcrystaline wax

Table 1.
Subject Anthropometry

Subject Number	Sex	Age	Stature (cm)	Weight (kg)	Forearm Subcutaneous Tissue (mm)		Ulna Length (cm)		Radius Length (cm)	
					R	L	R	L	R	L
1	F	89	161	45	<1	<1	24.50	24.29	22.61	23.22
2	M	80	175	49.5	2	2	27.70	28.08	26.37	26.77
3	F	87	162	64.5	6	6	24.81	25.04	23.01	23.44
4	F	68	168	65.9	8	5	27.80	26.86	25.91	25.32

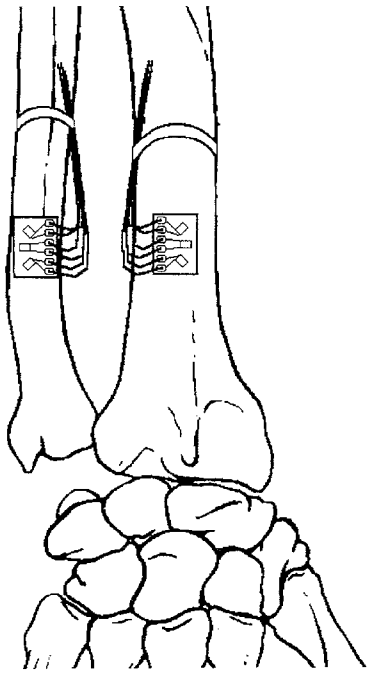

Figure 1. Strain gauge placement - gauges are on the posterior aspect of the radius and ulna.

(M-Coat-W1) was applied to the gauge and surrounding area, and allowed to set. A nylon cable tie was used to secure the wires to the bone with a loop left free for strain relief. The skin was closed with cotton mortuary sutures.

Three Endevco 7264-2000 accelerometers were positioned on a mounting block with their axes in an orthogonal arrangement as shown in Figure 2. The local coordinate system is designated as the x axis in the anterior - posterior direction with posterior positive, the y axis in the medial - lateral direction with lateral positive, and the z axis in the superior - inferior direction with inferior positive. This arrangement was positioned over the posterior wrist at the level of the ulnar styloid and another was placed at the antecubital fossa. Since it was felt that screwing the mounts directly into the bone would compromise the integrity of the bone, special mounts were designed that could be secured to the forearm with nylon cable ties. A six axis load cell in the steering wheel measured triaxial force and moment.

Subjects were positioned in a manner similar to that used in tests with anthropomorphic dummies. [10] The position simulates a driver in the process of turning the vehicle since this is an action that may bring the forearm very close to the air bag module. The subject was seated in the fixture shown in Figure 3 and centered behind the steering wheel, with the hips against the back of the seat. The forearm was positioned with the hand at the 10 o'clock position and the elbow at the 4 o'clock position for the right forearm and the 2 o'clock - 8 o'clock for the left. The steering wheel was rotated so that the seam of the air bag module was perpendicular to the forearm and the forearm

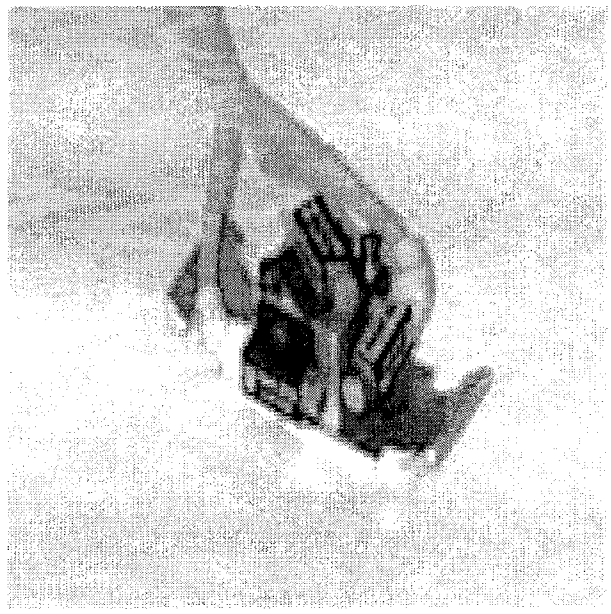

Figure 2. Forearm accelerometer mount - the accelerometers are in a triaxial arrangement.

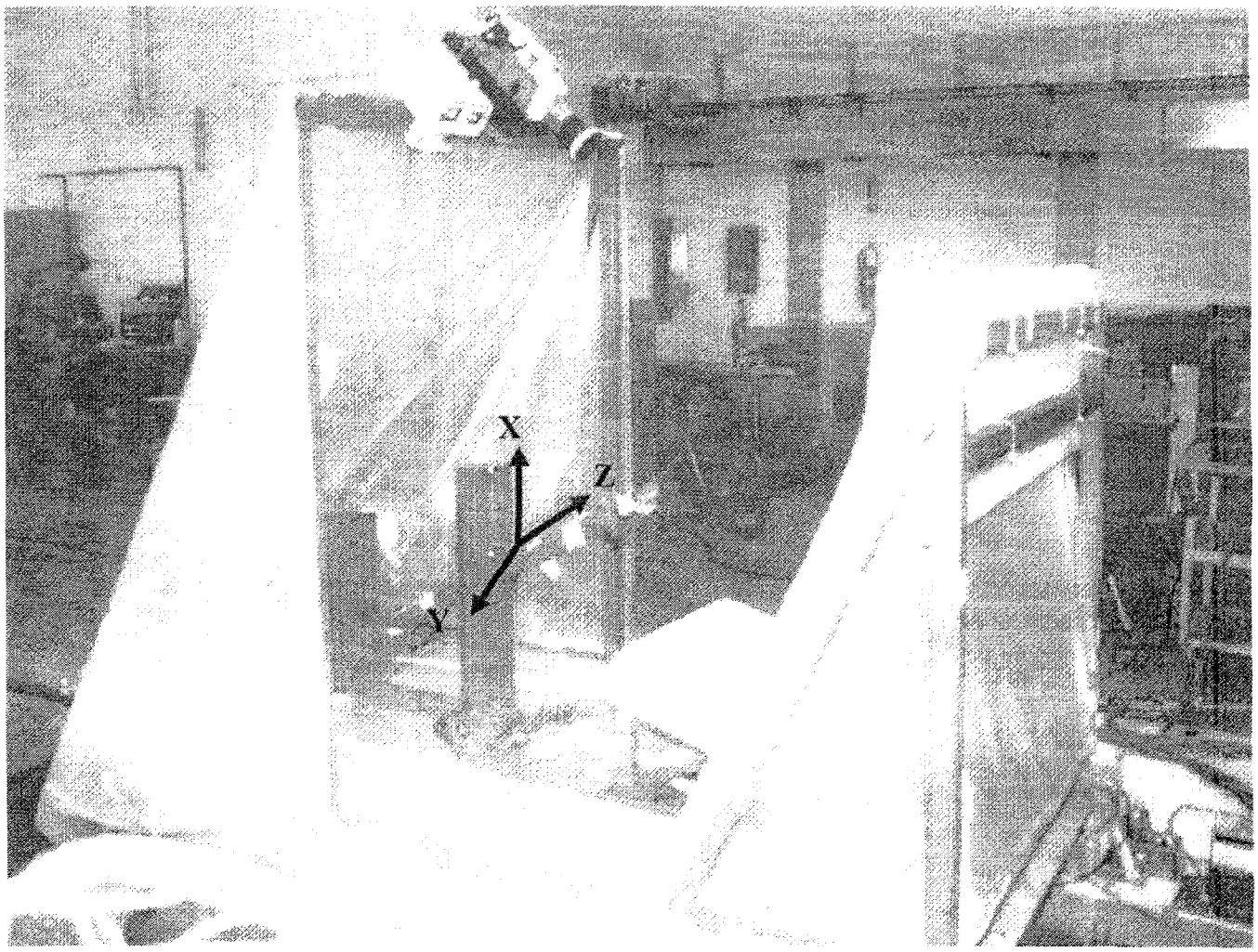

Figure 3. Test fixture. The orthogonal coordinate system for steering column loading is shown. The positive X-axis is up, the positive Y-axis is left and the positive Z-axis is toward the rear.

was placed against the air bag module. The hand was secured to the steering wheel rim with a single piece of masking tape around the fingers and steering wheel rim. The torso was positioned to allow the forearm to contact the center of the steering wheel module. This was accomplished with an additional piece of firm padding placed behind the back or shoulders of the subject.

Data was collected using a 96 channel data acquisition system. Signals from each transducer were transmitted via umbilical cable to a central data acquisition system. Analog-to-digital conversion was performed with a sampling rate of 12,500 Hz. Data was filtered to SAE Class 1000. Each test was recorded with high speed photography from overhead and from the side at 1000 frames per second.

The data from the strain gauge rosettes was reduced to give strain along the long and transverse axes, and shear strain.

Bone samples were collected at autopsy. The distal radius and ulna were disarticulated at the wrist and the distal 10 cm from each bone was removed for bone mineral and geometric property determination. The samples were stored in 10% formalin solution prior to scanning. The samples were scanned for bone mineral density and bone mineral content using a Lunar® DPX-L dual energy x-ray bone densiometer. Accuracy was assured by daily calibration against a phantom of known density. The images were analyzed for bone mineral content and bone mineral density in four 15 mm sections using custom software provided by the manufacturer.

After scanning, each sample was evaluated for cross sectional and inertial properties. Bones that had fractured were cut perpendicular to the long axis of the bone, 1 mm distal to the fracture site. The ends of each sample were smoothed and the periosteum was removed from around the cut end. Images were made of the ends and a Pascal version of the Slice program described by Nagurka and Hayes was used to calculate moment of inertia about the y axis and the cross sectional area. [16]

A logistic regression was performed to identify factors associated with bone fracture using SAS software. Bone mineral density, moment of inertia, steering column force and thickness of subcutaneous tissue and modulus of elasticity were assessed for ability to predict fracture.

RESULTS

The results of the testing is summarized in Table 2. Peak forearm acceleration and steering wheel force are listed as well as the mineral and geometric properties of the individual bone samples.

The external accelerometer mounts used for this study were designed to eliminate an invasive or destructive application. However, this presented several other problems. Although the wire ties used to secure the mounts were pulled as tightly as possible, the data was very noisy, suggesting a high signal to noise ratio or vibration in the mounts. To help compensate for this noise, the data was further filtered to SAE Class 180 for interpretation using a 2 pole Butterworth filter with a 300 Hz cut off frequency. In addition, the wire ties on the upper forearm mounts broke on every test so the upper forearm acceleration data was not considered for analysis. Other data acquisition problems resulted in no accelerometer data being collected for testing of the right forearm of the first subject and the left forearm of the second subject. Typical curves are shown in Figure 4. The H-91 bag showed consistently earlier and higher peak accelerations in the x direction than the L-92 bag. In

Table 2.
Test results

Sample	Air Bag	Fx	BMD (g/cm^2)	BMC (g)	Area (mm^2)	Iy (mm^4)	SWFZ (N)	LAXG (g)
1LR	H-91	Y	.369	.710	80.49	783.44	-1856.2	319.03
1LU	H-91	Y	.318	.554	88.02	860.92	-1856.2	319.03
1RR	L-92	Y	.387	.755	96.70	959.93	-1347.7	***
1RU	L-92	Y	.361	.647	67.80	580.72	-1347.7	***
2LR	L-92	N	.668	1.751	184.28	2987.55	***	***
2LU	L-92	N	.813	1.449	159.28	2618.48	***	***
2RR	H-91	N	.714	1.909	161.52	2195.46	-1680.0	437.72
2RU	H-91	N	.768	1.578	173.07	3038.42	-1680.0	437.72
3LR	H-91	N	.363	.781	71.48	567.45	-2073.2	402.00
3LU	H-91	N	.283	.484	58.67	394.37	-2073.2	402.00
3RR	L-92	N	.389	.884	79.77	648.85	-1340.7	376.86
3RUp	L-92	N	.265	.488	99.66	874.51	-1340.7	376.86
3RUd	L-92	Y	.187	.306	41.17	429.18	-1340.7	376.86
4LR	L-92	N	.368	.848	67.59	709.65	-1871.9	296.79
4LU	L-92	N	.482	.783	70.30	568.65	-1871.9	296.79
4RR	H-91	N	.400	1.033	89.29	1029.18	-2250.9	347.68
4RU	H-91	N	.386	.605	85.35	814.32	-2250.9	347.68

Sample = subject number- left/right- radius/ulna, Fx = fracture, BMD = Bone Mineral Density, BMC = bone mineral content, Area = Cross-sectional area, Iy = Moment of Inertia about the y axis, SWFZ = Peak steering wheel force in the z direction, LAXG = Peak forearm acceleration in the x direction, time = Time of peak value, *** = Lost data

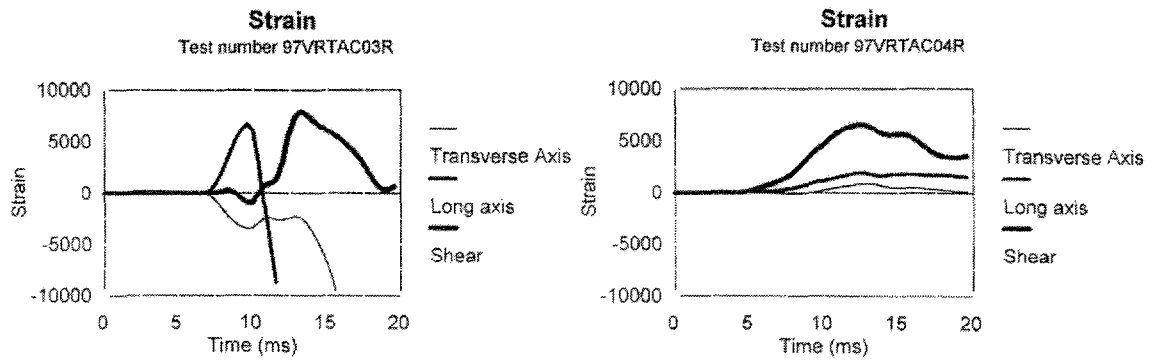

Figure 6. Typical strain time history. Air bag L-92 is on the left and shows the long axis is in tension and the transverse axis is in compression, suggesting the possibility of a bending mechanism. Air bag H-91 is on the right showing both the long and transverse axes are in tension suggesting that the loading caused an explosive injury.

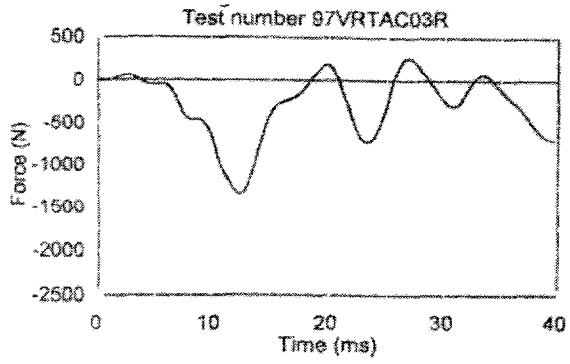

Figure 7. Typical steering wheel Z-axis force time history. Air bag L-92 is on the bottom with a maximum value of -1341 N. Air bag H-91 is on the top with a maximum value of -2073 N. The peaks fall within 1-2 ms of the peaks in forearm acceleration.

third region of right ulna #3 represents an average of the BMD values for the section with the bone ends together and the two broken ends. This procedure gave results consistent with the pattern seen for the other samples. The values reported in Table 2 are from the proximal section with the exception of 3RUd. This sample represents the distal end of the fracture site and the value for the third section is reported.

Cross sectional area and moment of inertia are reported in Table 2 on page 5. The values reported are from the diaphysis of the bone with the exception of sample 3RUd. This value represents the distal end of the fracture in that sample. Cross-sectional area was determined using both NIH Imager software and Slice software. Because the two methods gave very similar results, only the data from the Slice software is shown. Moment of inertia is calculated about the medial-lateral axis passing through the neutral axis of the bone. This is the axis of interest when considering bending in the sagittal plane.

DISCUSSION AND CONCLUSION

Whether or not a material fails is dependent on a number of factors, including the strength of the material and the magnitude of the force applied to it. Both of these factors were addressed in this study. The strength of a material is dependent on its geometric properties and its material properties. Assuming that bending is the mechanism of failure, a large moment of inertia will produce less stress for a given bending moment. Bone strength will also increase, up to a certain point, with an increase in the mineral content. A decrease in the force transmitted to the bone will also decrease the likelihood of fracture.

Logistic analysis is suitable to situations in which the dependent variable is characterized by discrete outcomes. Although some researchers [17,18] have argued that the logistic model is not a good model for assessment of injury risk unless the test conditions to cause or avoid injury are well defined and have a large number of reasonably well controlled test points, others [19] have used the logistic model where the outcome is either death or survival. In this study, since the predictor variables are continuous and the response, fracture or no fracture, is binary, a logistic model was used to assess fracture risk. Fit was assessed using a Wald Chi square test with significance defined as $p < 0.05$. Odds ratio was also calculated. It should be noted that this analysis is based upon a very limited sample size, and consequently, the results should be used cautiously.

Bone mineral density was examined first. In this study, subject 2 had larger, denser bones than the other three subjects and no fractures were produced. The other three subjects had similar BMD and moment of inertia, but

there were no fractures in subject 4 and only one in subject 3. This is in contrast to the results obtained by other researchers who were able to identify a value of bone minereal content, above which no fractures were produced and below which fractures were usually produced. The results of a logistic regression shown in Table 3 suggest that there is no significant relationship between the occurrence of fracture and bone mineral density in this study. Moment of inertia is not a significant predictor either.

Table 3.
Results of statistical analysis*

Factor	P- value for Wald Chi square test	Odds Ratio
BMD	0.1665	0.000
Iy	0.2764	0.998
SQ	0.0661	0.546
SWFZ	0.1237	1.003
E	0.1446	0.000
EIy (strength)	0.2981	0.996
SWFZ / SQ (attenuated force)	0.1081	0.995
Attenuated force / strength	0.0490	8.043

*Note that analysis is based upon a very limited sample size

BMD = bone mineral density, Iy = moment of inertia, SQ = thickness of subcutaneous tissue, SWFZ = steering wheel z-axis force, E = modulus of elasticity

Modulus of elasticity was then addressed. Strain in the longitudinal direction was differentiated to give strain rate. There was noise in this data so the graphs were examined over the interval of 5 to 10 ms, the time during which punchout occurs. The H-91 air bag showed a strain rate of about 1/sec and the L-92 was about 0.3/sec.

Apparent density was calculated by dividing the BMC by the volume of the scanned region. This segment was modeled as a cylinder and the volume was calculated by multiplying the cross sectional area by the length of the scanned segment. Modulus of elasticity was calculated using the equation

$$E = 3790\, \dot{\varepsilon}^{\,0.06}\, \rho^{3}$$

derived experimentally by Carter et al. [19] Modulus of elasticity was not found to be significant either.

Moment of inertia and modulus of elasticity were multiplied to give an overall strength factor but this did not seem to be a significant predictor of fracture either.

Steering wheel force along the z-axis was not consistent for all subjects but it was consistently higher for the H-91 air bag than the L-92. Logistic regression demonstrates that steering wheel force is not a significant indicator of fracture. Moreover, the steering wheel force may not be an accurate indicator of force transmitted to the bone. The force is not applied in full to each bone and it is not shared equally. The magnitude of the force actually transmitted to the bones is also in question. The first two subjects had very thin subcutaneous tissue in contrast to the thicker tissue of the other two. It is possible that the thicker subcutaneous tissue attenuated the force of the deploying air bag and less energy was transmitted to the bone. This provides an additional factor that could explain the fracture pattern observed. The steering wheel force was divided by the thickness of the subcutaneous tissue and this attenuated force was assessed. However, it was not a significant predictor of fracture either. The attenuated force and the bone strength were then related by dividing attenuated force by bone strength and a force to strength ratio was obtained. This was significant by logistic regression and the odds ratio was 8.034. Fracture vs. force to strength ratio was plotted and fit with a logistic curve with the equation:

$$\pi = \frac{e^{\beta_o + \beta_1 x}}{1 + e^{\beta_o + \beta_1 x}}$$

with $\beta_0 = -3.14$ and $\beta_1 = 2.08$. The probability of fracture is π and the fracture to strength ratio is x. This graph is shown in Figure 8.

This suggests a possible explanation for the observed fracture pattern. Subject 2 had very strong bones that were able to withstand the force of the deploying air bag. Subjects 3 and 4 had weak bones but their thicker subcutaneous tissue may have attenuated the force of the deploying air bag and enabled their bones to resist fracture. The first subject had weak bones and thin subcutaneous tissue and suffered fractures of all 4 forearm bones.

The fracture seen on the right ulna of the third subject due to the L-92 system would not be expected from the above explanation and deserves separate consideration. This was an oblique fracture with minimal displacement. Isolated ulna fractures are often caused by a direct blow to the ulna and are commonly called "nightstick fractures". These fractures are generally transverse or oblique. The ulna is susceptible to this type of injury because there is very little soft tissue on its medial aspect. For this test, the radial styloid was positioned about 76 mm from the center of the air bag module. The fracture was about 58 mm from

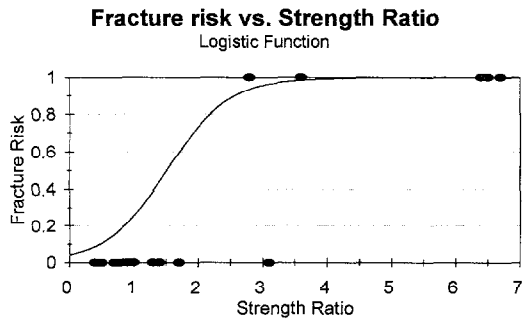

Figure 8. Graph of fracture risk vs. force to strength ratio. The logistic function is fit to data. Note that this fit is based upon a very limited sample size.

the radial styloid. Both of these measurements are subject to error. Accurate location of the radial styloid and the center of the air bag module was difficult due to instrumentation and clothing and could only be estimated. The measurement from the post-test x-ray is subject to error due to positioning for the x-ray and the displacement of the bone ends. With these considerations it is possible that the fracture site was very close to the center of the air bag module. If the forearm was not in full pronation at the time of deployment, the ulna could have taken a direct hit from the edge of the module cover as it was opening. This cannot be confirmed from the high speed film due to the unsuitable camera angle.

The strain gauges were problematic. Only 4 of the 16 gauges survived to autopsy with the wiring intact. No data was collected for two of these due to data acquisition problems. Four others generated useful data prior to failure. Only two were on bones that fractured.

Three of the surviving gauges were on tests using the less aggressive air bag. These all show a similar pattern of tension along the dorsal long axis of the bone and compression on the transverse axis. This pattern suggests a bending load. The tests using the more aggressive air bag showed a different pattern, one of biaxial compressive stress. For tests of the right forearm of the second subject and the left forearm of the third subject, there was compression in both the transverse and longitudinal axes. There was biaxial tension for the test of the right forearm of the fourth subject. This could happen for one of two reasons. If the gauge was mounted on the side of the bone instead of the posterior aspect, it could generate this pattern of stress. The post-test x-rays were examined for strain gauge placement. The radial strain gauge was positioned slightly toward the lateral aspect for the test of the right forearm of the fourth subject, but the other two were well centered over the posterior aspect of the radius. The other possible explanation is that the loading was not bending but a high energy explosive type of mechanism.

This study was limited by the small sample size and this probably accounts for the fact that there was no significance shown for any individual factor by chi-squared analysis. Statistical analysis was complicated by the fact that only 4 subjects were used and it is difficult to obtain reliable conclusions from such a small sample. Viewing each bone as a separate observation was felt to be reasonable since each bone had a unique BMD and moment of inertia, and since the alternative of testing additional subjects is not feasible at this time. The other limitation is the age of the cadavers. Only one was less than 80 years old and this is not representative of the average driver. The reduction in bone mineral density and muscle mass that occurs with aging probably increased the incidence of fracture. [21, 22, 23]

The estimation of the force attenuation was based on the assumption that the response had a linear relationship to the thickness of the subcutaneous tissue. Research by Robinovitch et al. supports this but it also assumes that the force attenuation is proportional to the force. [24] This may not be the case, instead, a given thickness of tissue may attenuate a constant amount of force despite the magnitude of the force delivered. Since Robinovitch used only one force level in his research, this relationship remains unclear.

If padding due to soft tissue is playing a significant role in force attenuation, it is possible that padding the air bag module cover could reduce the risk of injury. Experimentation with different padding materials and different thickness using dummies and the Research Arm Injury Device would be appropriate to address this question.

The strain gauge data suggests the possibility of a fracture mechanism other than bending at high energy levels. Examination of data from other laboratories for evidence of similar discrepancies and further research with isolated limbs could provide further information.

This study looked at forearm fractures as a result of the punchout phase of driver's side air bag deployment. It was determined that bone strength alone is not a good predictor of fracture risk and that the force generated by the bulging air bag module may be attenuated by the soft tissue of the forearm. The combination of bone strength and attenuated force provided a better predictor of fracture risk. Further research in the area of fracture mechanism, and force attenuation by padding is needed.

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