9

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Axial Loading Injury Tolerance for the Small Female Wrist

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ABSTRACT

Previous experiments with human cadavers and side airbags revealed the potential for wrist injuries as a result of the hand becoming entrapped in the handgrip. These injuries included osteochondral fractures of the carpal bones and comminuted distal radius and ulna fractures. The purpose of this paper is to develop an injury tolerance for the small female wrist that may be used in the design phase of side airbags in order to reduce the risk of wrist injuries resulting from side air bag deployment. Small female cadaver upper extremities were used to develop the wrist tolerance as a conservative estimate of the most vulnerable section of the driving population. In order to simulate the interaction between the wrist and handgrip in a controlled test environment, a pneumatic impactor was configured with simulated handgrip. The applied axial pulse from the impactor was controlled to match the force onset rate, impulse, and peak force in order to simulate the load profile of a deploying side airbag. A total of 17 ($n=17$) axial impact experiments were performed on the wrists of small female cadavers. Instrumentation included an impactor load cell to measure axial load, accelerometers on the impactor mount for mass compensation, and magnetohydrodynamic angular rate sensors on the hand and forearm to measure wrist flexion. Bone mineral density was determined for each subject, and injury analysis was performed using radiographs and detailed necropsy. Post-test necropsy revealed that 9 of the 17 tests resulted in wrist injuries. The injury patterns were identical to those observed from cadaver tests with side airbags and included fractures of the scaphoid (AIS 2), lunate (AIS 1), distal radius (AIS 3), and distal ulna (AIS 2). The peak impactor force ranged from 344 N to 3616 N. Using the injury outcome as the binary variable, a logistic regression analysis was performed. When mass scaled to the 5^{th} female, the analysis produced an injury risk function that predicts a 50% risk of injury at a wrist load of 1700 N ($p = 0.0037$). Risk of injury was found not to be dependent of subject bone mineral density ($p = 0.49$), age ($p = 0.99$), mass ($p = 0.31$), and stature ($p = 0.69$). Based on the similarities in impact load profile and observed injury patterns between the impactor tests and the side airbag tests, it is suggested that the injury risk function will accurately predict the risk of wrist injuries in the automobile crash environment.

INTRODUCTION

One of the most common injuries observed by orthopedic surgeons is the distal radius fracture (Lidstrom *et al.*, 1959). The typical injury mechanism is a fall on the outstretched hand, thereby producing an axial compressive force across the wrist on the distal forearm. There are many names for fractures of the distal radius, such as Colles, Smith, Barton, and Pouteau (Jupiter *et al.*, 1997). Unfortunately, there is limited published data on the dynamic loading required to cause these injuries. However, there are several studies that offer insight into the long term changes associated with distal forearm injuries. Short *et al.* (1987) used cadaveric specimens to examine disabilities arising from radius fractures. He inserted load cells into the proximal radius and ulna and used pressure-sensitive film in the radioulnar carpal joint. He showed that if the radius was not restored to its pre-fracture length, the ulna load could increase from 21 % to 67 %. Werner *et al.* (1992) used 58 cadaveric specimens to examine radial shortening and also found that any change in the ulna or radius length will dramatically alter the force transmission. While the applied loads in these studies were quasi-static and below injury level, they do provide insight into the serious nature and possible long term complications associated with distal forearm injuries.

Although dislocations and fractures of the carpal bones have been well documented, there are multiple load paths that may induce injuries. Several authors suggest that hyper flexion is an important mechanism (Friedenberg *et al.*, 1949, and Wilson *et al.*, 1925), but many more consider hyperextension to be the primary mechanical component that contributes to carpal injuries (Fisk *et al.*, 1970, Hill *et al.*, 1970, and Linscheid *et al.*, 1972). Hyperextension is related to the typical position of the outstretched hand as a person falls to the ground and explains why it is the predominant injury mechanism for wrist injuries. In addition, pre-existing radial or ulnar deviation can contribute to carpal injuries (Weber *et al.*, 1978). Large radial or ulnar deviation can result not only in carpal injuries, but also in radial or ulnar styloid fractures (Bonnin *et al.*, 1943). Although the incidence and severity of carpal injuries are well documented, no published studies report the dynamic force required to induce injury. Moreover, the theories on wrist injury mechanisms are often contradictory.

In addition to falls, the interaction between a deploying side airbag and the upper extremity has been shown in several experiments to result in wrist injuries. Jaffredo et al. (1998) found a wrist injury in a male cadaver test with that was the result of the hand becoming entrapped in the handgrip as the side airbag forced the upper extremity forward. Duma et al. (2001) evaluated side airbag deployment and the effect of a door mounted handgrip with six small female cadavers. The tests resulted in two wrist injuries: a transverse fracture of the distal radius, and an osteochondral fracture of the lunate carpal bone. Given the results of these cadaver tests, and the recent increase in implementation of side airbags, the purpose of this study was to develop a dynamic wrist injury criterion for pure axial loading. The present study was the first to apply pure axial loads to the palmar side of the extended wrist. In order to develop a wrist injury criterion, dynamic axial loads were applied across the wrist joint in a similar manner as the resulting wrist loading in the side air bag tests.

METHODS

The test configuration was designed to impact the palm and force the wrist into compression. Only female subjects were used with an instrumentation package configured to determine the load across the wrist joint. Injury evaluation was performed using post-test radiographs and necropsy.

Test Subjects

The left and right aspects of nine small female cadaver subjects were utilized for these experiments (Table 1). Pre-test radiographs of each upper extremity were taken to identify any pre-existing skeletal abnormalities. Specimens were not included in the study if any bone pathology was observed. The bone mineral density of each specimen was determined by using the Osteogram® technique (CompuMed Inc., Los Angeles, CA). This procedure uses radiographs of the subject's hand and forearm with a phantom for calibration. The radiograph is then scanned and the bone mineral density is determined and presented as an Osteogram® bone mineral density index, or BMD Index. The BMD Index is not the actual bone mineral density, but rather an index number relative to other Osteogram® scans. The more useful output from this technique is the bone mineral density T-score, or BMD T-score, and Z-score, or BMD Z-score. The BMD T-score represents the number of standard deviations away from the average the subject's bone mineral content is compared to the average healthy individual between 25 and 50 years. In this application, each subject is compared to healthy females. The positive or negative sign denotes greater or lower bone mineral density respectively. T-scores at –1.0 or greater are considered normal, between –2.5 and –1.0 are considered osteopenia or low bone mineral content, and below –3.0 is considered osteoporotic. The Z-score is the number of standard deviations away from the average bone mineral content of females at the subject's exact age.

Table 1. CADAVER SUBJECT ANTHROPOMETRY FOR ELBOW COMPONENT TESTS.

Subject	Sex	Age (years)	Weight (kg)	Stature (mm)	BMD Index	BMD T-score	BMD Z-score
10	Female	86	66	1651	67.8	-4.0	-0.9
100	Female	42	71	1676	120.0	0.9	0.9
13	Female	80	41	1626	65.1	-4.2	-1.3
14	Female	63	65	1676	58.7	-4.8	-2.1
3	Female	72	55	1702	83.9	-2.5	0.1
2	Female	60	63	1651	99.9	-1.0	0.5
101	Female	66	52	1676	90.3	-1.9	0.2
99	Female	64	91	1626	128.2	1.6	3.3
106	Female	72	67	1778	101.5	-0.9	1.7
Average ± St. Dev.		67 ± 13	63 ± 14	1674 ± 47	91 ± 24	-1.9 ± 2.2	0.3 ± 1.56

Although the average bone mineral density of the test subjects was –1.9 ± 2.2 standard deviations below that of a young healthy female as presented by the T-scores, the average bone mineral density of the test subjects was 0.3 ± 1.56 standard deviations above the normal when adjusting for subject age as presented by the Z-scores. In other words, the subjects had very low bone mineral density relative to young females, but had nearly the average bone mineral density when compared to the test population with an average age of 67 ± 13 years. The effect of bone mineral density on injury outcome will be examined in the results section. Post-test radiographs and detailed necropsy were used to observe resulting injuries. The cadavers were obtained through the Virginia State Anatomical Board with permission of the family given to conduct biomechanics research. All test procedures were approved by the institutional review board at the University of Virginia. Screening for Hepatitis A, B, C, and HIV was conducted with each cadaver prior to acceptance into the research program.

Test Configuration

Due to the contradictory wrist injury mechanisms presented in the literature and the complex wrist loading observed in the side air bag experiments, it was necessary to choose a single wrist injury mechanism as a starting point. Pure axial compression of the wrist was selected based on an examination of the high speed video and handgrip loads from previous side air bag experiments (Duma *et al.*, 2001).

The energy source used to apply the axial loads was a pneumatic impactor. The initial tank pressure, impactor mass, and foam between the impactor and the transfer piston were adjusted to tailor the impact pulse such that the wrist force, onset rate, and momentum transfer were similar to the those calculated from the dummy upper extremity in the side air bag tests (Figure 1). The upper extremities were disarticulated at the mid-shaft humerus location and supported by two cables that held the elbow in 90° of flexion. The forearms were kept intact in order to preserve the load distribution allowed by the interosseous ligament (Pfaeffle *et al.*, 1999). A modified handgrip was constructed out of aluminum and rigidly attached to the transfer piston configuration. The purpose of this handgrip was not to simulate the automobile handgrip, but rather to provide a narrow and rigid contact surface that was used to apply the impact load directly to the wrist.

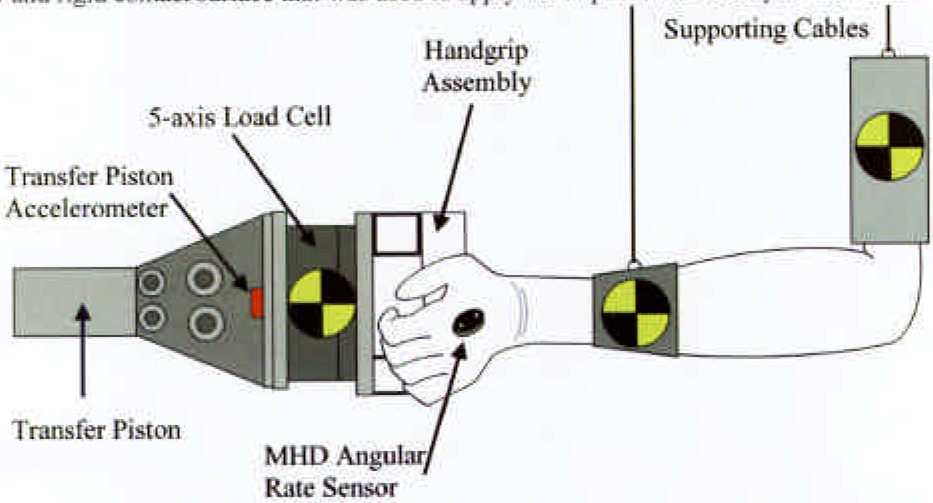

Figure 1. Lateral view of the wrist test configuration. A pneumatic impactor (not shown) strikes the transfer piston causing compression of the wrist joint.

The wrist was initially oriented on the handgrip in extension (Figure 2). The palm was positioned on the handgrip without overlap in order to minimize the applied moment due to off-axis loading. This position was established as the most vulnerable position of the wrist that was observed in the side air bag tests. The wrist was held in place using a light tape that broke upon impact. There was no pre-load applied to the wrist through the forearm, which was free to translate away from the handgrip after impact. The handgrip was covered with a course sand paper to prevent the palm from sliding off during impact.

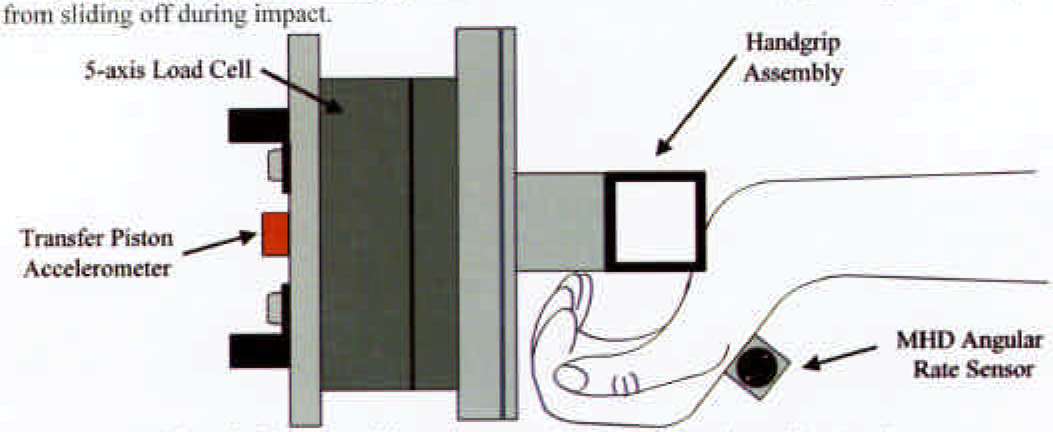

Figure 2. Top view of the wrist test configuration illustrating wrist extension and placement of the palm on the handgrip.

Instrumentation

A five axis load cell (Model 1968, Robert A. Denton Inc., Rochester Hills, MI) measured the reaction forces between the transfer piston and the handgrip. In order to determine the axial load, or load along the long axis of the forearm, applied to the wrist, an accelerometer (Model 7264a, Endevco Inc., San Juan Capistrano, CA) on the transfer piston was used to compensate inertially for the mass of the handgrip attachment and one half of the load cell. A magnetohydrodynamic angular rate sensor (Model ARS-04E, ATA Sensors Inc., Albuquerque, NM) was attached to the posterior surface of the hand (Hall *et. al.*, 1997). By measuring the initial wrist orientation, the angular rate sensor data was used to measure wrist flexion during the impact. Data acquisition was performed with a sampling rate of 10,000 Hz and all test data were filtered to CFC 180 filter class.

RESULTS

A total of 17 wrist impact tests were performed on 17 specimens. A discussion of the sensor and injury data is presented followed by an analysis the subject anthropometric parameters.

Sensor and Injury Data

The wrist peak force, onset rate, and momentum transfer values were determined from the inertially compensated impactor load cell axial force (Table 2). The impact tests were characterized by an average force onset rate of 505 ± 368 N/ms and an average momentum transfer of 19.9 ± 8.7 Ns. These values were within the range of onset rates and momentum transfer values as observed in the handgrip loads of the most aggressive side air bag test 587C and the least aggressive side airbag test 585A (Figure 3) (Duma *et al.*, 2001). Due to the complex loading of the wrist in the side air bag tests, the resultant handgrip load was used as a conservative estimate of the applied wrist loads to compare with the axial loads of the component tests. Analysis of the high speed video showed that the hand remained in contact with the handgrip and did not separate until the end of the impact loading event. This indicates that the force and momentum were transferred through the wrist in each event as desired.

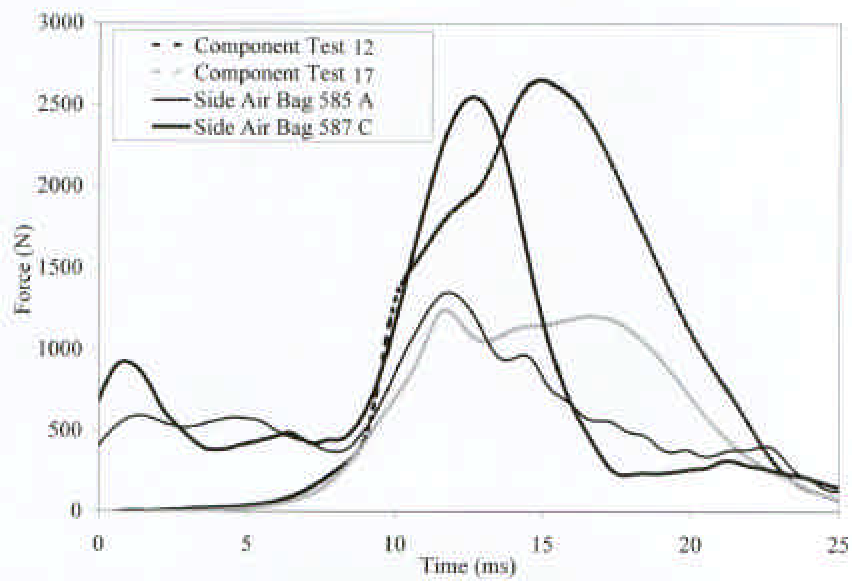

Figure 3. Wrist axial loading from component tests W14 and W19 compared to the resultant applied wrist load in side air bag tests 585 and 587 as measured from the inertially compensated resultant handgrip load

Table 2. WRIST LOAD AND INJURY DATA FOR THE WRIST COMPONENT TESTS.

Test	Specimen	Peak Force (N)	Force Onset Rate (N/ms)	Momentum Transfer (Ns)	Observed Injuries
1	13 Left	1199	174	12.6	None
2	13 Right	2001	738	17.0	Distal Radius fx (AIS 2) Ulna Styloid fx (AIS 2)
3	10 Right	1794	352	14.1	None
4	10 Left	2007	508	21.5	Scaphoid fx (AIS 1) Distal Radius fx (AIS 2) Ulna Styloid fx (AIS 2)
5	2 Left	1412	182	14.4	None
6	3 Right	890	140	9.2	None
7	14 Right	1657	435	16.1	Comminuted Distal Radius fx (AIS 3) Ulna Styloid fx (AIS 2)
8	99 Right	2708	893	26.8	Scaphoid Chondral Avulsion fx (AIS 1)
9	100 Left	2212	242	21.8	None
10	101 Right	3358	430	28.9	Scaphoid Osteochondral fx (AIS 2)
11	106 Left	2385	758	28.4	Comminuted Distal Radius fx (AIS 3)
12	3 Left	2653	395	24.0	Distal Radius fx (AIS 2) Scaphoid fx (AIS 2)
13	14 Left	418	38	5.1	None
14	99 Left	4336	1006	33.2	Chondral Abrasion Distal Radius (AIS 1)
15	100 Right	4698	1460	36.3	Comminuted Distal Radius fx (AIS 3) Ulna Styloid fx (AIS 2) Lunate Avulsion fx (AIS 2)
16	101 left	2584	571	16.2	None
17	106 Right	1242	269	13.5	None

The wrist angular rotations were tracked for each test by the angular rate sensor on the posterior wrist. The initial wrist extension angle ranged from 55° to 62°. The typical impact was characterized by additional wrist extension of approximately 10° as the impactor loaded the palm of the hand and forced the wrist rotation (Figure 4). After the peak wrist load, the wrist rotated back towards the neutral position.

Nine of the 17 tests resulted in at least one wrist injury. The observed injuries ranged from the most severe comminuted distal radius fracture (AIS 3) and ulna styloid fracture (AIS 2) of test 7 compared to the minor chondral abrasion of the radius in test 14. The observed injuries to the distal radius, ulna styloid, and carpal bones were similar to the type of wrist injuries observed in the previous side air bag tests (Figure 5). Given that the wrist force onset rates and momentum transfer values were comparable to the handgrip loads of the side air bag tests, and observed injuries of the component tests were similar to those observed in the side air bag tests, it was assumed that the dynamic loading conditions of the wrist component tests were representative of wrist loading that resulted from side air bag deployment.

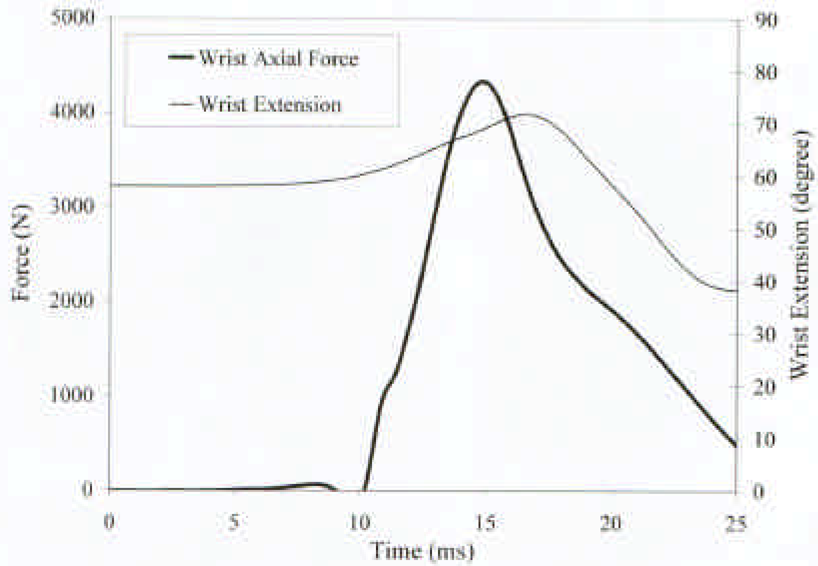

Figure 4. Measured wrist force and extension for test 14.

Effect of Subject Anthropometric Parameters

Logistic regression analysis was used to investigate the effects of subject anthropometric parameters. The analysis showed that the injury outcome was independent of specimen age ($p = 0.99$), mass ($p = 0.31$), stature ($p = 0.69$), and bone mineral density ($p = 0.49$). The lack of dependence on subject anthropometric parameters was attributed to the narrow sample size in this group of experiments, which consisted only of small female specimens.

Wrist Injury Criterion

An important factor in choosing a statistical model for injury criteria development is whether or not the measured parameter is dependent or independent of injury. In other words, it must first be determined if the peak response occurs at the same time as injury. If the time of injury is concurrent with the peak response, then that response is dependent of injury. This would be the case in the system sustained a catastrophic failure and is no longer able to support load. Otherwise, the response is independent of injury. The peak wrist force was assumed to be independent of injury. Given that the chondral abrasions and fractures were minor injuries, it seems logical that the joint was capable of transmitting the impact load after a cartilage injury occurred. In cases were the impact resulted in a comminuted fracture of the distal radius, it may be argued that the injury occurred at the time of peak load due to the catastrophic failure of the bone; however, assuming the independence of force and injury is the more conservative approach in order to account for the less serious cartilage injuries. The logistic regression technique was used to develop the wrist injury criterion given the assumed independence of injury and peak wrist force. In addition, the fact that wrist tests resulted in a binary injury outcome, tests with and without injury, further supported the use of logistic regression.

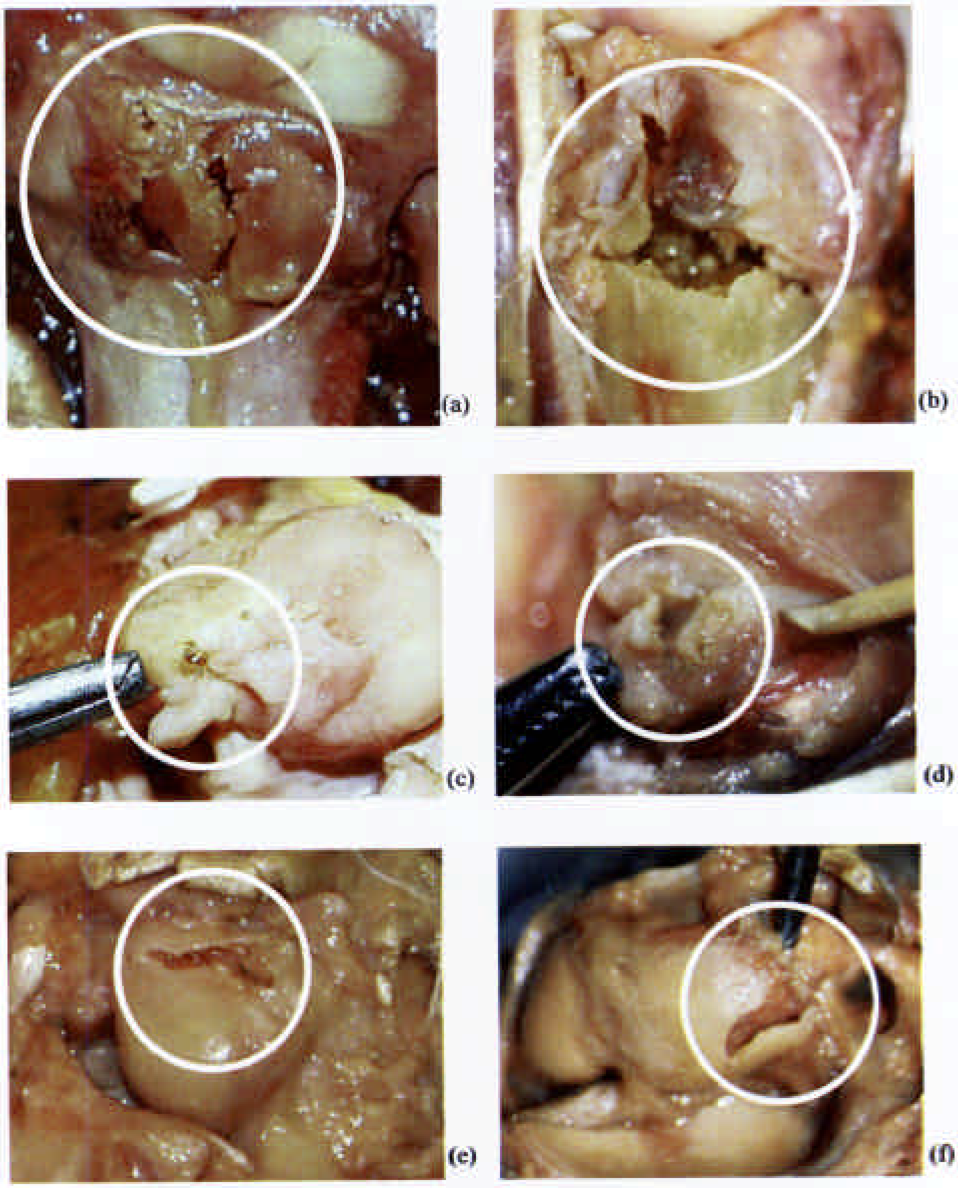

Figure 5. Comminuted distal radius fracture from component test 7 (a) compared to the transverse distal radius fracture of a side air bag test (b); ulna styloid fracture from component test 4 (c) compared to the ulna styloid fracture from a side air bag test (d); scaphoid osteochondral fracture from component test 10 (e) compared to lunate osteochondral fracture from a side air bag test (f).

In order to produce the injury criterion for the fifth percentile female, the peak wrist axial forces were scaled using the mass scaling procedures of Eppinger *et al.* (1984) (Table 3). The mass of the fifth percentile female was taken as 48 kg. The logistic regression analysis produced the wrist injury criterion based on peak wrist axial load ($p = 0.0037$) that predicted a 50 % risk of wrist injury at a wrist axial load of 1700 N (Equation 1 and Figure 6). For the logistic regression equation, the constants were established as **a** = -4.40, **b** = 0.00259, and the **x** variable was the applied load through the wrist joint.

Table 3. WRIST PEAK FORCE VALUES AFTER SCALING FOR THE FIFTH PERCENTILE FEMALE.

Test	Specimen	Subject Mass (Kg)	Scaling Coefficient	Measured Peak Wrist Axial Force (N)	Scaled 5th% Femal Wrist Axial Force (N)
1	13 Left	41	1.05	1199	1323
2	13 Right	41	1.05	2001	2207
3	10 Right	66	0.90	1794	1449
4	10 Left	66	0.90	2007	1621
5	2 Left	63	0.92	1412	1183
6	3 Right	55	0.96	890	818
7	14 Right	65	0.91	1657	1363
8	99 Right	91	0.81	2708	1777
9	100 Left	71	0.88	2212	1703
10	101 Right	52	0.97	3358	3174
11	106 Left	67	0.90	2385	1917
12	3 Left	55	0.96	2653	2438
13	14 Left	65	0.91	418	344
14	99 Left	91	0.81	4336	2845
15	100 Right	71	0.88	4698	3616
16	101 left	52	0.97	2584	2443
17	106 Right	67	0.90	1242	998

$$\text{Probability of Wrist Injury }(x) = \frac{1}{1+e^{(4.40-0.00259\cdot x)}} \quad (1)$$

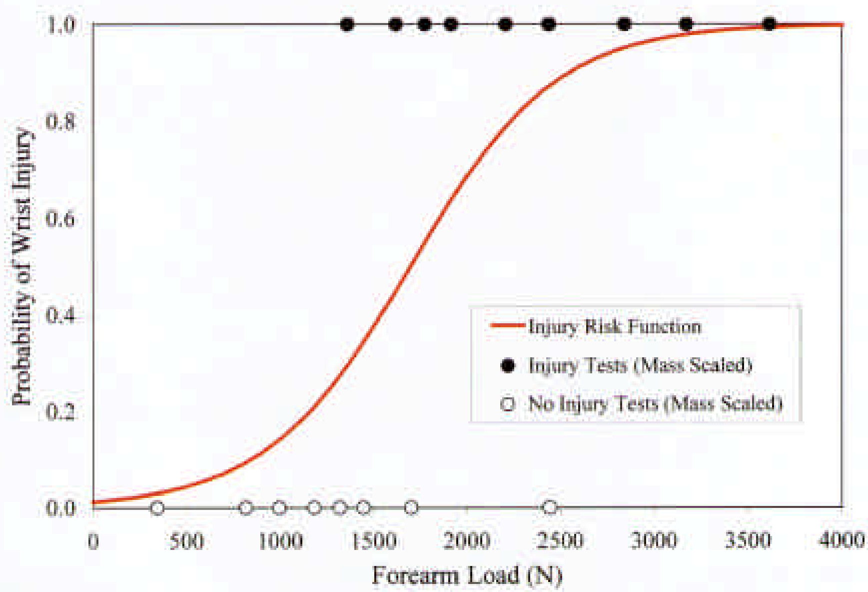

Figure 6. Wrist injury criterion based on peak axial (FZ) wrist load of the fifth percentile female.

CONCLUSIONS

The applied axial pulse from a pneumatic impactor was controlled to match the force onset rate, impulse, and peak force in order to simulate the load profile of a deploying side airbag. A total of 17 (n=17) axial impact experiments were performed on the wrists of small female cadavers. Small female cadaver upper extremities were used to develop the wrist tolerance as a conservative estimate of the most vulnerable section of the driving population. Post-test necropsy revealed that 9 of the 17 tests resulted in wrist injuries. Using the injury outcome as the binary variable, a logistic regression analysis was performed. When mass scaled to the 5^{th} female, the analysis produced an injury risk function that predicts a 50% risk of injury at a wrist load of 1700 N ($p = 0.0037$). Risk of injury was found not to be dependent of subject bone mineral density ($p = 0.49$), age ($p = 0.99$), mass ($p = 0.31$), and stature ($p = 0.69$).

Based on the similarities in impact load profile and observed injury patterns between the impactor tests and the side airbag tests, it is suggested that the injury risk function will accurately predict the risk of wrist injuries in the automobile crash environment. It is anticipated that this criterion will be used directly in the Hybrid III fifth percentile female instrumented dummy upper extremity to help reduce the risk of wrist injuries from side airbags. In the dummy upper extremity the wrist load is calculated using the mid-shaft forearm load cell and the forearm acceleration. The z-axis load (FZ) should be inertially compensated for the mass of the distal forearm between the load cell and hand, 0.48 kg, by using the z-axis forearm accelerometer. This calculated wrist load should be used directly in the wrist criterion to predict risk of wrist injury. Since axial load is not the only mechanism producing wrist injuries with side airbags, additional research is needed to elucidate the effects of an applied moment on the risk of wrist injury. Combined loading must be addressed before a comprehensive predictive function can be determined for wrist fractures. This data may be useful to understand the combined loading mechanisms by aiding computational studies that address the limitations and lack of biofidelity of the dummy upper extremity.

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DISCUSSION

PAPER: **Axial Loading Injury Tolerance for the Small Female Wrist**

PRESENTER: ***Jeff Crandall, Automobile Safety laboratory, University of Virginia***

QUESTION: *Mike Schlick, Medical College of Wisconsin*

We also used the instrumented arm and I saw a lot of rotation on that elbow. And you say, though, it's axial loading that causes the injury to the wrist. Is that rotation, which I imagine happened in the cadaver also, was that later and can that be isolated using your auditory pickups?

ANSWER: We've done a one-to-one comparison in the side airbag test of the instrumented arm to the cadaver test and there are a couple things we've observed. Number one is position sensitivity. We found we had to be within a few millimeters of repeatable positioning conditions in order to get the same behavior from the airbag, very specific. We had to use a FARO arm for all our initial positioning.

The second thing is in terms of general behavior between cadavers and the instrumented arm; the largest difference we saw had to do with the shoulder complex. The initial humerus interaction is fairly similar, but once you start to push the humerus forward, the Hybrid III dummy, for example, which we used, does not have shoulder compliance. Therefore, after a certain amount of forward motion of the humerus you start to lose bio-fidelity.

The third thing I would mention is in terms of stiffness. What we developed here is a forced criteria based only on the human response. The instrumented arm in terms of axial compression is probably not very bio-fidelic. While we've done matched comparisons in bending modes, we have not done it in axial compressions. Thanks for the question.

Q: *Guy Nusholtz, Daimler Chrysler*

Did you do anything with the angular velocity sensor that you had on the hand because nothing showed up?

A: Yes, I only had 10 or 15 minutes to talk so I had to skip it. Let me tell you what we did do with that. We started the initial position at about 45 to 60 degrees of extension. And this was where we saw the position in which we saw the maximum forces applied in the side airbag test. Then what we wanted to do is we wanted to look at, during the duration of axial load, how much flexion or extension occurred during that time because we really wanted to minimize it, we wanted it purely to be an axial load. We saw between 8 and 10 degrees I believe. That is to say, if it started 45 to 60 degrees of extension, we saw 8 to 10 degrees of flexion during the event.

Q: In general, how censored was your data? Was it always say 80 percent up, was it close 80 percent near the peak? Your fracture occurred before the peak force.

A: In terms of the acoustic sensored data?

Q: Yes, in terms of the acoustic sensor, did you get some halfway up and some 80 percent?

A: No. In fact, the acoustic sensor is still under development and we didn't use it for all cases, otherwise we could have picked the exact forces and used that in the analysis. It's always within 4 or 5 milliseconds, probably something of that vicinity.

Q: But what percentage of the peak, is it like 80 percent of peak?

A: More than 80, more like 90, 95 percent but this could be loading condition specific.

Q: *Srini Sundarajan, Ford Motor Company*

Jeff, what are the incidents of these types of fractures in the field, and are you comparing the result of the side airbag to an axial loading on the hand? What would happen if you didn't have a side airbag, what type of injuries would the person suffer versus the injuries suffered here?

A: Yes, that's a good question. In fact, there was a nice study, but I can't remember whom the first author was from the Birmingham group, presented at IRCOBI a number of years ago where they looked at the differences in terms of upper extremity injuries. These do exist in the non-airbag environment. You would have these types of injuries from the hand, wrist, and forearm coming forward into a door for example. So they do exist.

In terms of these side airbag injuries, there are almost no documented cases because we just don't have that many field cases. But if they exist, we want to protect against them, we can produce them in the laboratory and we just want to protect against them in the field.

Q: My concern is, from the automotive point of view, if you're trying to offer protection to one mode of injury and now we have another regulation if you have another criteria for another mode we may throw the baby with the bath water so we just want to be careful about that.

A: From my standpoint, the goal of this study was to identify the biomechanical tolerance of this region. If countermeasures can be developed to prevent these injuries, then that is great but I agree that there has to be a complementary analysis on frequency, risk, side-effects, and cost-benefit. I appreciate your concerns.