CHARACTERIZATION OF LEG INJURIES FROM MOTOR VEHICLE IMPACTS

Tyler A. Kress, Ph.D.
Engineering Institute for Trauma & Injury Prevention
The University of Tennessee, U.S.A.
David J. Porta, Ph.D.
Bellarmine College, U.S.A.
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ABSTRACT

The objective of this investigation was to understand relationships among loading characteristics as they affect the kinematics and injury of a pedestrian's lower extremity. Real-life pedestrian and motor vehicle collision scenarios were modeled by impacting 604 human cadaver intact legs and long bones with a cart/guide rail impacting system designed to simulate the front end of an automobile. A parametric study was conducted that varied the boundary conditions on the foot as well as test parameters such as loading direction, impact velocity, and impactor geometry. The series of tests can be categorized as follows: **(1)** Fracture Characterization, **(2)** Threshold Velocity, **(3)** Friction versus Inertial Constraint, **(4)** Anterior and Lateral Thigh Impacts, and **(5)** Embalmed vs. Unembalmed. Documented data for various specimens include, but are not limited to, specimen anthropometrics, fracture patterns, failure force levels, and calculated bending moments. Representative values include averages as follows: Failure forces for the tibia ranged from 1.19 to 7.07 kN. Failure forces for the femur ranged 1.31 to 8.37 kN. Bending moments averaged from approximately 100 to 500 Nm. These values varied depending on the speed of impact, impactor geometry, direction of impact, gender of specimen, etc. The results and observations may be helpful as an aid for evaluating the effectiveness of any protective or mitigative devices or strategies.

INTRODUCTION

Pedestrians account for approximately one out of seven of all traffic fatalities. The number of pedestrian injuries every year from motor vehicles is at least a factor of ten greater than the number of deaths. Statistics indicate that head injury is the leading cause of fatalities among pedestrians. The most common injury is to the leg. The United States National Traffic Highway Safety Administration recently stated that the development and evaluation of test procedures which assess the injury potential of vehicle structures is needed.

The injury potential in a motor vehicle collision with a pedestrian is high because of the great difference between the mass of a vehicle and that of a pedestrian. Since the mass of a vehicle is so much greater than that of a pedestrian, the velocity of can be very low and still have very significant injury-causing potential. A vehicle does not need to achieve high velocity in order to severely damage pedestrians. Approximately 80% of pedestrian impacts occur with vehicles moving less than 35 mph (NHTSA 1993a, NHTSA 1994b). A report issued by NHTSA in late 1999 (NHTSA 1999) titled "Literature Review on Vehicle Travel Speeds and Pedestrian Injuries" states that approximately 5% of pedestrians die when struck by a vehicle traveling at 20 miles per hour or less. This is compared to fatality rates of 40, 80, and nearly 100 percent for impacting speeds of 30, 40, and 50 (or more) miles per hour respectively.

The kinematics associated with vehicle-pedestrian impacts are often governed more by the shape of the impacting vehicle than its mass or velocity. Buses, trucks, and vans tend to have broad, flat leading edges that enable them to strike a larger portion of a pedestrian including at a height at their center of gravity. When this is the case, the body is generally thrown forward. There tends to be less sever injuries under these circumstances if blows to the head are avoided. With most cars, the hood and bumper design are lower and more pointed for reasons of aerodynamics and style. Therefore, cars tend to strike below the center of gravity of a pedestrian such that the pedestrian is undercut and lands on the hood or roof of the vehicle (Aldman 1984a).

The first car was manufactured by Henry Ford in 1893 (Grun 1982). Soon thereafter pedestrian-automobile impacts were so common that specific injuries were attributed to them. The term "bumper fracture" was used in the medical literature at least as early as 1929 by Cotton and Berg to identify fractures of the proximal tibia (Hendrix

1972; Levine 1994). This type of injury is indicative of the normal sequence of events when a pedestrian is struck by an automobile. The episode usually begins with initial contact at the lower limb by the bumper (Manoli 1986). This may result in a variety of injuries to the lower extremity including fractures, lacerations, avulsions, and occasionally, hip dislocations (McClelland et al. 1987). Often, a bumper (approximately 17-23" above the ground can impact the leg crushing the common fibular nerve. This causes a condition known as "foot drop" in which the extensor muscles of the foot do not function properly (Huelke 1986). In a study of pedestrian victims, the incidence of fracture was observed to steadily increase with age such that the over-65 group had the highest incidence (3.5) of fractures per patient. (O'Malley et al. 1985).

Breaking strength and fracture patterns of long bones have been studied quite extensively with good documentation dating as far back as the 19th century. Messerer (1880) tested 500 bones from 90 cadavers of both sexes and various ages. He found that the cracking or tearing of the bone generally occurred on the convex (tension) side of the bone. For bones exhibiting significant bend there was crushing on the concave (compression) side, at the point of application of the load, before a tearing or tension fracture occurred. The significance of tensile stresses as the cause for bone failure was further emphasized by Evans and Lissner (1948) through stresscoat studies. Mechanical property studies over the years have shown that bone is weaker in tension than in compression. Rauber (1876) was one of the first researchers to discover that when a bone is subjected to increasing amounts of equal tensile and compressive forces it fails in tension first. Kress and Porta (1993) have found that the human femur seems to be approximately 1.5 times stronger in compression than tension, even during dynamic loading conditions.

Nyquist et al. (1985) and Kress et al. (1990) have probably provided the most thorough studies but there is still room for improvement in the analyses of anthropometry and injury, especially with respect to fracture patterns. Although Kallieris and Schmidt (1988) presented a considerable amount of anatomic data, theirs is the exception to the rule. Most previous publications fail to even mention age, sex, causes of death, or specific sites of impact and the resultant injuries to the cadavers. Therefore, one emphasis of the work reported on in this paper is to provide detailed anatomic data alongside basic engineering findings. Attempts are made to investigate any correlation between the two.

PURPOSE

Reduction or prevention of impact injury through design of protective devices/safer environments requires certain biomechanical information. This information includes a characterization of how the body region of interest responds to impact forces in terms of mechanical parameters such as force-time histories of impact, accelerations/decelerations, and deformations in the tissue structures. Also, mechanisms by which the tissues fail, mechanical parameters by which they respond, and the values of the injury criteria are important results of impact biomechanics research. These "biomechanical behaviors" and "injury characterizations" for a pedestrian's lower extremities are the essence of the efforts discussed in this paper. In order to keep this paper to a reasonable length, not all data could be included. A list of several observations near the end of the paper is an attempt to provide succinct information regarding some of the results.

GENERAL METHODOLOGY

Injuries to the human leg and fractures of the diaphyseal portion of long bones were investigated by conducting dynamic impact tests of 604 geriatric human cadaver lower extremity components. The majority of the tests involved the use of a horizontal pneumatic-based accelerator which propelled a 50-kg cart along a railway at 7.5 m/s. The cart was headed by varying striking surfaces (pipe, plate, bumper sections, and padded plate) instrumented with force transducers. Impactor speed was slower for many tests, and some comparative studies were made between embalmed and unembalmed bare bones and intact legs. Impactor leading edge geometry was also varied during some of the tests. Intact lower extremities and bare long bones were impacted under various conditions, including anterior and lateral impacts at approximately 7.5 m/s; as well as anterior, posterior, lateral and medial midshaft impacts at approximately 5.0 m/s. Specifically, the series of tests can be categorized as follows (number of specimens dedicated to each test series is also specified): **(1)** Fracture Characterization – 558 long bones and intact legs, **(2)** Threshold Velocity – 8 embalmed legs, **(3)** Friction versus Inertial Constraint – 8 embalmed legs (4 matched pairs), **(4)** Anterior and Lateral Thigh Impacts – 12 embalmed legs (6 matched pairs), and **(5)** Embalmed vs. Unembalmed – 12 embalmed and unembalmed legs, (6 matched pairs).

SPECIFIC METHODOLOGY, DATA SOURCES, & RESULTS

Documented data for various specimens include, but are not limited to, specimen anthropometrics, fracture patterns, failure force levels, and calculated bending moments. Representative values include averages as follows: Failure forces for the tibia ranged from 1.19 to 7.07 kN. Failure forces for the femur ranged 1.31 to 8.37 kN. Bending moments averaged from approximately 100 to 500 Nm. These values varied depending on the speed of impact, impactor geometry, direction of impact, gender of specimen, etc. All of the documented data cannot be included in this short paper, therefore some representative results are presented, and observations and conclusions are summarized. For an extremely thorough and comprehensive breakdown and discussion of the data, refer to the Ph.D. dissertation titled "The Anatomy and Biomechanics of Experimentally Traumatized Human Cadaver Lower Extremity Components" by Porta (1997).

Test Series 1 – Fracture Characterization

A total of 558 bone fracture tests were performed to gain knowledge with regard to pedestrian leg fracture characteristics and behavior. All bare bones were tested in a pin-pin setup and the intact leg tests were mostly pin-inertial (foot hanging freely) or pin-friction (shoed foot on concrete block). The pin-pin setup supported the bare bones at their ends (epiphyseal aspects) which were impacted at midshaft.

The testing apparatus consists of a pneumatic-based accelerator which propels a wheeled cart toward the mounted specimen. The accelerator consists of a piston assembly inside of a pneumatic chamber that is pressurized in order to achieve target velocities. For most tests the pressure was 0.34 MPa (50 psi) yielding a cart velocity of approximately 7.5 m/s. A ram connected to the piston pushed an aluminum and steel impact cart (50 kgs) throughout its stroke of approximately 1.5 meters. Then the cart separated from the ram and traveled along a railway for less than a meter before striking the specimen. In that stretch, it was timed by a photovoltaic cell/timer apparatus permitting the calculation of the velocity before impact.

Heading the cart for most of the tests was an instrumented 10-cm long steel impactor pipe with an outside diameter of 4.13 cm. It was mounted to the front of the cart via slide pins. When contacting a specimen, the pipe was freely able to impinge on a piezoelectric quartz force transducer (PCB Piezotronics model 208A03), thereby producing a measured force equal to that which is delivered to the specimen. The transducer signal was recorded on a Hewlett Packard 3562A signal analyzer allowing storage of a force vs. time plot for each impact.

Test Series 2 – Threshold Velocity

As part of an attempt to validate a fuzzy logic computer model developed at The University of Tennessee Engineering Institute for Trauma and Injury Prevention and to further understand threshold velocity at which legs fracture, different test scenarios were utilized for each of eight embalmed cadaver leg impacts. Some of tests of individual specimens were rather interesting. As can be seen in Table 1, all but one of the eight specimens were impacted multiple times. The protocol called for the first impact of a specimen to occur below 5.0 m/s and subsequent impacts were made at gradually higher velocities until fracture. A skin flap was cut in each specimen in order to aid with the inspection of the bone between tests. As might be expected, the most resilient bone was that of the youngest male (55 years old). The leg was impacted 19 times with fracture finally achieved at a velocity of 8.0 m/s and a peak force of 6.24 kN. Surprisingly, the second strongest specimen was that of a 92 year-old female. Her limb failed on the ninth impact which was made at 7.9 m/s with a peak force of 3.29 kN recorded. Figure 1 shows the second and last impacts for this specimen. Notice that the leg maintains its structural integrity in the top photo (and each of the first eight impacts) but the last impact photo shows the limb wrapping around the impacting plate as the bone fails. At the top of the picture a special rigging is visible, a portion of the weight system utilized to apply a 20 kg downward force vector through a rod that was passed transversely through a drilled hole in the femoral condyles.

Table 1.
Leg Tests for Threshold Velocity Analysis

Leg #	Age & Sex	Site of Impact	# of Impacts	Velocity @ Fracture (m/s)	Peak Force (kN)	Test Conditions (all legs had a 20 kg weight applied downward)
755-L	56 f	Anterior	8	6.3	1.49	V P S A F
991-L	92 f	Anterior	9	7.9	3.29	V P S A
009-L	55 m	Anterior	19	8.0	6.24	V q S
070-L	81 f	Anterior	3	5.2	1.14	V P
949-L	84 f	Anterior	2	4.9	1.67	V P S
070-R	81 f	Posterio r	3	2.8	1.14	V P A
961-R	80 f	Posterio r	1	5.1	1.34	V P S A
949-R	84 f	Posterio r	5	6.3	2.06	V P S A

* Legend for Test Conditions: V- 20 kg mass exerting an downward force vector on the specimen; P- Impact surface was a 10 cm wide steel plate; q- Impact surface was a 4 cm wide plate; S- Shoed foot placed on concrete block; A- Air springs positioned between cart and impact plate; F- Foam placed on face of plate

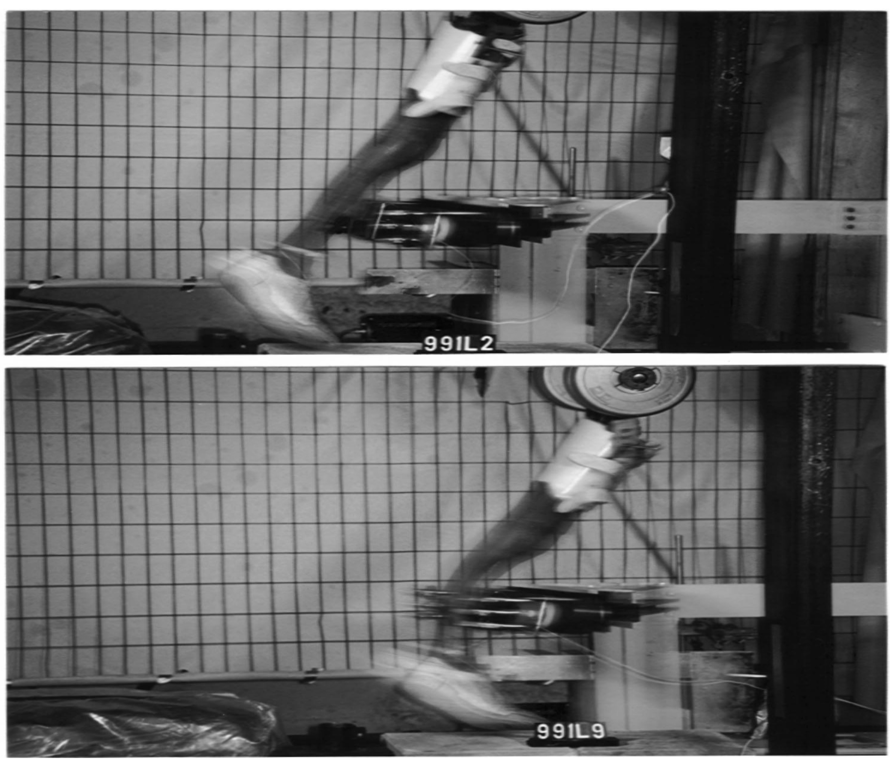

Figure 1. Impacts of a leg resulting in no fracture and fracture.

All of the limbs were thoroughly dissected and, as expected, there was little soft tissue damage in these embalmed specimens. Nerves were typically observed to be grossly unaffected. It was noted that posterior impacts resulted in more distal fractures than the anterior strikes.

Test Series 3 – Friction vs. Inertia

The application of a frictional component was one aspect of the variety of conditions employed in the “threshold velocity” tests that appeared particularly amenable to this next series of tests. According to Aldman (1984a), during walking, the leg supports between 80 and 120% of the static body weight. He noted that a bumper force of 4 kN may fracture an unloaded tibia, but with body weight it may be only 1 kN. A simple controlled study was designed in an effort to examine the effects of simulated body weight and frictional forces on peak impact forces. Matched pairs of embalmed legs from four cadavers were collected for this study. The right legs were each suspended by a rod passed through a drilled hole in the femoral condyles. The foot did not contact the ground. The inertial mass effects of the foot and ankle were the only distal constraints on these limbs. When the left legs were tested, they were positioned either precisely like those of the previous study (Figure 2) or in a vertically oriented rigging.

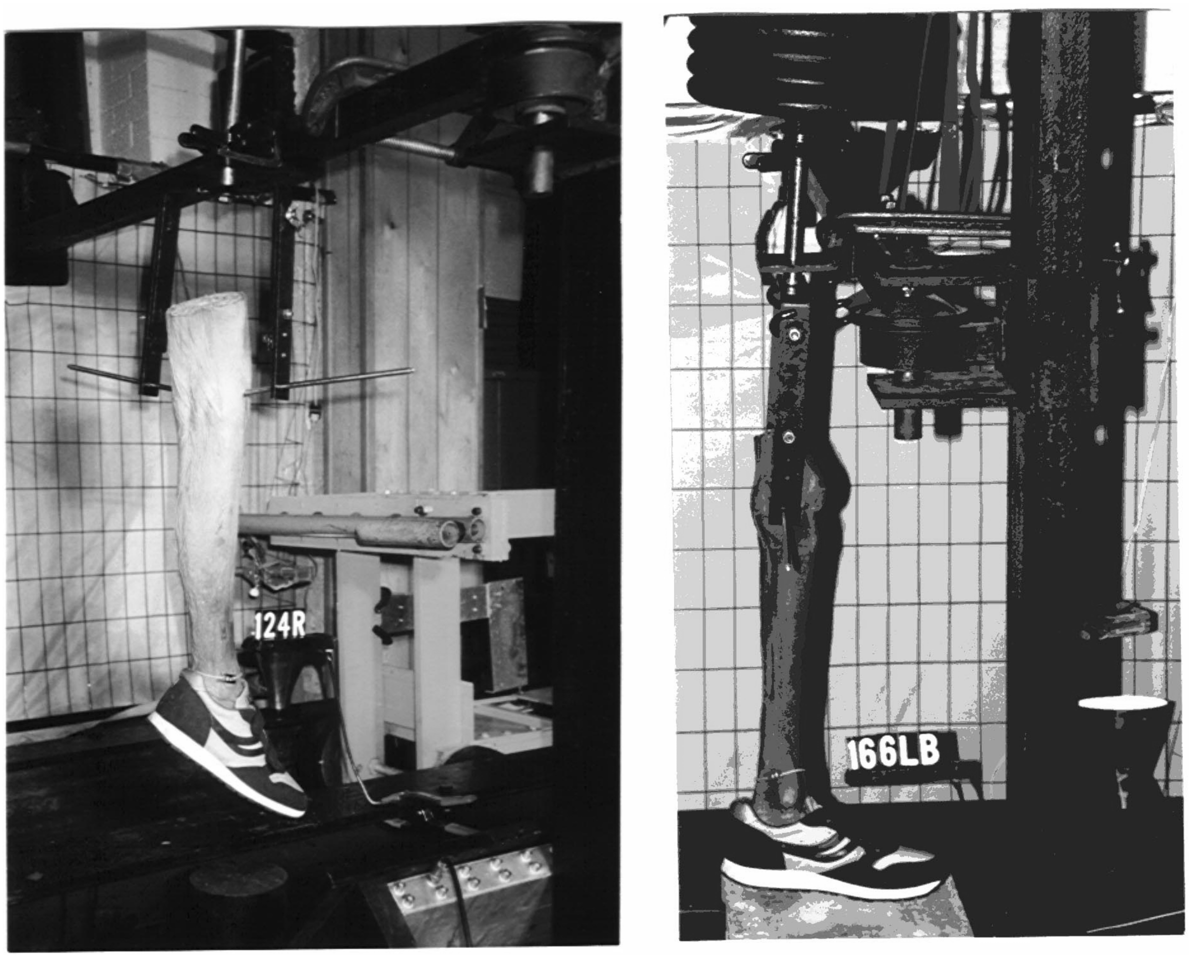

Figure 2. Set-up for tests for inertial and frictional constraints.

All specimens were impacted on the anterior surface of the mid-leg. Test results are presented in Table 2. It appears that the results of this small study conflict with the statement by Aldman (1984a) regarding differences in forces for loaded versus unloaded legs since 3 of the 4 peak forces recorded for the limbs in the frictional set-up were higher than their inertially constrained counterparts. *Student's T-test* was performed in order to identify the significance level for the force and constraint data. The average peak force for the friction study was 7.20 kN (standard deviation = σ = 1.97 kN) while the inertially supported limbs were
subjected to an average peak force of 5.94 kN (σ = 0.70 kN). There was not a statistically significant difference between the two groups when a 5% confidence level is desired. It should be noted that there was no significant difference in cortex thicknesses between the groups (average thickness of bones in friction set = 5.51 mm (σ = 0.86 mm); average thickness of bones in inertial set = 5.21 mm (σ = 0.68 mm). Nor were there any significant differences ($p = 0.9100$) between the mean velocity of impact for the friction (7.89 m/s with σ = 1.01 m/s) versus the inertial constraint (7.97 m/s with σ = 0.90 m/s). Although this is a small study, there appears to be a trend towards slightly greater damage in the limbs with the frictional constraint. Major muscular damage and severed blood vessels were observed in 3 of the 4 friction limbs and only 1 of the 4 inertially constrained limbs. In addition, one of the only impacts to result in a severed nerve occurred to a limb in the friction set up.

Table 2.
Results of Friction vs. Inertia Series

#	Age & Sex	Set-up[1]	Velocity (m/s)	Peak Force (kN)	Fracture Pattern	Avg. Cortex (mm)
110	71 m	L - Frxn	7.51	8.23	Segmental	4.77
		R- Inert	7.49	6.27	Segmental	5.14
124[2]	74 f	L- Frxn	7.26	7.55	Comminuted	6.36
		R- Inert	7.42	6.73	Transverse	5.98
166[3]	64 m	L- Frxn	9.40	8.93	Transverse	6.13
		R- Inert	9.31	5.63	Compression Wedge	5.37
051	85 m	R- Frxn	7.37	4.33	Comminuted	4.78
		L- Inert	7.64	5.14	Comminuted	4.35

[1] Limb side: L-Left; R-Right; Frxn- Friction constraint; Inert- Inertial constraint
[2] A prosthetic knee was found in Leg 124-R.
[3] The left & right legs had to be impacted 5 and 2 times respectively until fracture.

Test Series 4 – Anterior and Lateral Thigh Impacts

Additional tests were performed in the investigation of injury from perpendicular impacts to the mid-thigh. Matched pairs of thighs from six embalmed cadavers were studied. In order to use the same cart impactor the specimens were inverted and suspended against a large steel elbow plate in the test zone by a rod passing through the tibial plateau such that impact occurred on the anterior or lateral surface. Testing results are shown in Table 3.

Table 3.
Results of Anterior vs. Lateral Thigh Impacts at 7.3 m/s
(all thighs were inverted and simply supported)

#	Age & Sex	Impact[1]	Peak Force (kN)	Fracture Pattern	Avg. Cortex (mm)
919	79 m	L – Ant	na	Oblique	7.17
		R – Lat	4.61	Jagged Transverse	7.61
901	86 m	L- Ant	8.23	Jagged Transverse	6.92
		R- Lat	7.36	Jagged Oblique	6.76
879[3]	79 m	L- Ant	na	Segmental	7.76
		R- Lat	8.96	Oblique	7.32
860	95 f	R- Ant	5.29	Oblique[4]	6.90
		L- Lat	4.85	Comminuted Oblique	5.43
846	88 f	R[2]- Ant	5.75	Comminuted Tension Wedge	5.61
		L- Lat	4.52	Oblique[4]	6.38
810	78 f	R- Ant	3.96	Oblique	7.14
		L- Lat	6.74	Comminuted Segment	5.68

[1] Limb side and Impact Site: L-Left; R-Right; Ant- Anterior; Lat- Lateral
[2] A prosthetic plate was found in Thigh 846-R.
[3] The left and right legs had to be impacted 5 and 2 times respectively until fracture
[4] Additional fractures were found in the hip- due to interaction with the support system

Analysis of the dissection data from these embalmed thighs was very similar to that from the embalmed legs. There was very little superficial damage and precious few vessels and nerves were grossly injured. Muscular damage was generally observed to be greater in the compartment opposite the site of impact. A prosthetic device in specimen 846R appears to have acted as a stress riser since fractures emanated from the screw hole closest to the impact area.

Test Series 5 – Embalmed vs. Unembalmed

Intact legs from six geriatric cadavers were fractured in a controlled study aimed at documenting the effects of embalming on both the soft and hard tissues of cadaver specimens subjected to biomechanical impact research. Upon bequeathal, one leg was removed and frozen while the other remained with the cadaver for embalming. The embalmed legs were amputated later and pre-test radiographs were made. For testing, a rod was inserted in the upright leg such that simulated upper body mass could be applied. A 50 kg cart propelled by a pneumatic accelerator to 7.5 m/s struck the anterior leg midway between the knee and ankle. The cart was headed by an instrumented steel pipe (4.75 cm dia.) coupled to a transducer which relayed impact force data to a Hewlett Packard 3562A signal analyzer. Testing was captured on standard VHS video (30 frames/s) and 16 mm Color High Speed Film (1,000 frames/s). Post-test analyses included radiographs and thorough dissection. Peak forces were comparable for matched pairs.

Figure 3 shows impact tests of two legs, one embalmed and the other unembalmed.. Note the increased wrapping of the unembalmed leg around the impacting pipe after fracture has occurred. The foot is still in its original orientation with respect to the vertical plane. A portion of the tibia can be seen protruding from the posterior aspect of the leg.

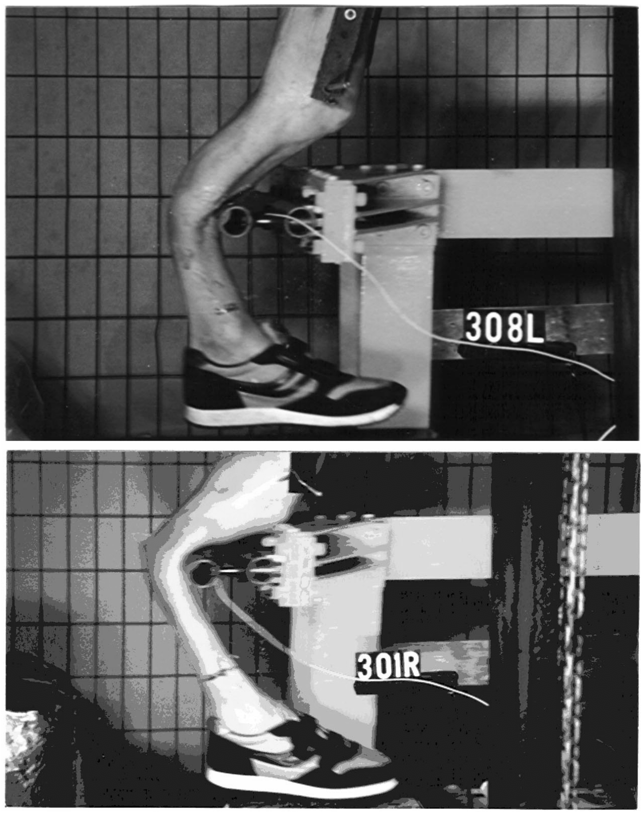

Figure 3. Comparison of embalmed leg impact (top) to that of an unembalmed leg impact (bottom)..

The matched pair of legs from a cadaver were subjected to identical test conditions with the only difference being the fact that one leg was embalmed. Therefore, age, sex and overall physical condition could essentially be "factored out" allowing for a more meaningful comparison of the collected fracture and soft tissue damage.

Every effort was made to make the cadaver specimens and test scenario as "life-like" as possible in hopes that the dynamic response would be similar to that of a pedestrian struck in the leg by a vehicle. Testing conditions attempted to account for: 1) the constraints of the upper body mass, 2) friction between the foot and the pavement, and 3) pressurization of the vasculature.

The frozen specimens were allowed to thaw for at least twenty-four hours. Immediately prior to testing, the specimens were removed from their plastic bags and a hole was drilled from side-to-side in the distal femur. A rod was passed through the hole and the leg was placed upright in the impact zone of the test machine. A weight of over 50 kgs was applied to the rod in an effort to simulate the upper body mass. The foot of the specimen was placed in an athletic shoe and set on a concrete surface. Additionally, for most tests, an attempt was made to pressurize the vasculature by using a crude embalming machine to infuse the vessels with a sugar water solution via cannulation of the popliteal or femoral artery. The machine registered a pressure of 2 to 3 psi (Resting systolic blood pressure of 120 mm Hg equals roughly 16 KPa or 2.3 psi.). Two of the legs could not be pressurized due to the presence of fixed blood in the vessels (3L) and an abnormal branching pattern of the femoral artery that resulted in vessels too small to accept the cannula (5L).

Data collected regarding cadaver demographics, specimen embalming, mid-shaft tibial cortex thickness, cart velocity and peak force are listed for each test in Table 4.

Table 4.
Test Data for Embalmed vs. Unembalmed Series

Cadaver Age, Sex & Cause of Death*	Leg &	Embalmed?	Avg. / Smallest Cortex Thickness (mm)	Cart Velocity (m/s)	Peak Force (kN)
1. 74 year-old Female, Lung Cancer	Right	yes	4.33 / 1.97	7.08	No Trigger
	Left	no	4.93 / 2.10	7.94	5.95
2. 92 year-old Male Cardiac Arrest, Diabetes	Right	yes	6.74 / 4.33	7.15	6.80
	Left	no	6.56 / 3.53	7.62	7.80
3. 94 year-old Female, Pneumonia	Left	yes	6.24 / 3.36	7.30	4.78
	Right	no	4.13 / 2.90	7.87	4.18
4. 75 year-old Male, Lung Cancer	Left		y7.79 / 4.61	7.30	8.46
	Right		n7.49 / 4.81	7.84	6.21
5. 79 year-old Male, Myocardial Infarction	Left	yes	7.89 / 4.25	7.71	8.46
	Right	no	7.74 / 4.79	7.69	7.43
6. 91 year-old Female, Urosepsis	Right	yes	4.15 / 2.48	7.48	5.03
	Left	no	5.34 / 3.05	7.84	3.75

* Cadaver 2 was African-American, all others were Caucasian.

Dissection results indicating damage to skin, muscles, vessels and bone are summarized in Table 5. The fractured unembalmed specimens showed considerably more soft tissue damage than their fractured embalmed matches. Lacerations to the skin and superficial fascia were judged to be greater in five of the six pairs. Muscle damage was greater for the unembalmed leg in all six cases and vessel damage was greater in four of the six. Oddly enough, the nervous system appeared to escape serious injury as there was virtually no gross damage to any of the nerves. It is important to note that no microscopic analysis was performed. Since nerve components are often injured by "stretching" or "pinching," it is quite probable that damage was present but went undetected. The comparison of the osteologic data is more complex. The damage was similar in half of the matched pairs, but the other half appeared to show greater comminution of the embalmed legs. Further review of the post-test radiographs may lead to a more clear picture regarding bone damage.

Table 5.
Damage Summary for Embalmed vs. Unembalmed Series

Leg[1]	Laceration[2]		Muscles Damaged[3]	Vessels[4]	Bone Fractures[5]
1Re	4	20.5	5% TA	Fib V	Bad Comm >15 pieces
1Lu	2	34.0	60% Gas & Sol, 100% EHL & TP, 50% FDL, 50% FHL, and 10% FibB	None	Bad Comm >6 Pieces with Protrusion
2Re	1	1.5	10% FDL	None	Mild Comm Transverse
2Lu	1	7.5	50% FDL, 20% TA, 20% Gas, 30% Sol, 50% FHL, and 5% FibB	P. Tib A & Vs	Mild Comm Oblique
3Le	6	13.5	50% Gas, 40% Sol, 33% FHL, and 50% FDL	Part of Saph V	Bad Comm >15 pieces with Protrusion
3Ru	2	19.5	50% Gas, 50% Sol, 30% FHL, and 90% TP	Fib A & Vs	Mild Comm Transverse
4Le	0	0	<5% TA	A. Tib. A	Mild Comm Transverse
4Ru	1	1.5	10% TA, 10% Gas,, 10% FHL, and 5% TP	Fib Vs	Mild Comm Transverse
5Le	2	3.5	2 cm vertical tear in Gas	None	Mild Comm Transverse
5Ru	1	1.5	10% Gas & Sol, 30% FHL, 50% FDL, and 5% FiB	P. Tib A	Mild Comm Segmental
6Re	1	13.0	10% FDL, 10% TP, 75% FHL, and 5% TA	None	Comm >6 Pieces with Protrusion
6Lu	2	13.5	30% FDL, 10% TP, 75% FHL, 30% Gas and 30% Sol	Fib A & Vs	Mild Comm Oblique with Protrusion

[1] The specimen test number is listed followed by a designation for left (L) or right (R) and embalmed (e) or unembalmed (u).
[2] The number of skin lacerations is listed, followed by the total linear distance those cuts travel (cm).
[3] The percent values represent an estimate of the horizontal tear length as it relates to total width of the particular muscle. Muscle key: Gas= Gastrocnemius, Sol= Soleus, T= Tibialis, Fib= Fibularis, A= Anterior, P= Posterior, F= Flexor, E= Extensor, D= Digitorum, H=Hallucis, L=Longus, B= Brevis.
[4] Key for name of lacerated artery (A) or vein (V): P.= Posterior, A.= Anterior, Tib= Tibial, Fib= Fibular, Saph= Saphenous
[5] Comm = Comminuted.

Dissection data clearly indicate that soft tissue damage to fractured embalmed legs was much less than that seen in fractured unembalmed legs. Specifically, damage was greater to the skin, the superficial fascia, muscles and blood vessels; however, the nerves were an exception. In some cases, blood vessels were punctured and large muscle masses were torn for several centimeters; however, to the naked eye, nerves remained intact. The immediate question is whether this accurately models the live human response to anterior leg trauma. This question is addressed in the following two paragraphs.

Perhaps live nerves are rarely transected in mid-leg anterior impacts and the lack of damage seen in this study is appropriate. If so, then the resistance to laceration may be explained by several mechanisms: a) The anatomy of the lower limb may afford nerves a tremendous amount of protection from anterior impacts to the mid-leg. Most of the large nerves are situated posterior to the bones of the leg; therefore, fractures would absorb much of the energy of impact prior to involvement of the nerves. b) Transection may not be the most common mechanism of injury. Stretching is often cited as the cause of central nervous system injuries such as diffuse axonal injury. Compression of the brain is the primary cause of concussions. Maybe peripheral nerves of the leg are most often injured in similar manners without being grossly torn.

If nerve transection is commonly seen after "real-world" anterior mid-leg impacts then there may be factors which were not, or could not be accounted for: a) Live nerves may simply be more fragile than those of a cadaver. b) Perhaps when all of the components of the leg have their normal turgor, the nerves are put in a more precarious position. c) Nerve transection may occur secondary to the impact. This would include violent motion of the fractured limb immediately after impact or improper splinting/transport, etc. It may also include the human body's post-traumatic responses. Transection may occur during contraction of the musculature immediately after impact. This could result in laceration of the nerves as they are pinched between sharp bone fragments.

OBSERVATIONS AND CONCLUSIONS

1) Resultant fracture types for perpendicular loading (anterior-to-posterior, posterior-to-anterior, lateral-to-medial, and medial-to-lateral) do not differ with respect to impact direction. 2) It is reasonable to assume that transverse, oblique, segmental and tension wedge fractures are all just manifestations of tensile failure. 3) The most common fracture pattern is tension butterfly wedge and is followed closely by the oblique fracture. 4) The tension wedge fracture pattern can definitively be used as an indicator of the direction of impact. 5) The fracture patterns at low speed impacts (1.2 m/s) are very similar to those of high speed (7.5 m/s). This is somewhat of a unique observation because it has been commonly thought that the butterfly wedge results only from high speed impacts. 6) Spiral fractures only appear when bones are subjected to torsional loads. 7) Embalmed intact leg fractures exhibit greater comminution than unembalmed. The embalment process causes significant increase in stiffness of the soft tissue containment. 8) Although the femur is stronger and has a different cross-sectional geometric shape, its fracture patterns as a result of transverse loading are generally the same as those for the tibia. 9) Transverse and oblique fractures generally have jagged edges. 10) Spiral fractures have the "smoothest" break edge, perhaps indicating that it follows some pre-existing engineering structural line. Wedge fracture lines tend to follow curved paths similar to the spiral fracture path. 11) Tensile wedge fractures clearly originate at a location directly opposite of the point of impact and the wedge segment radiates back through the bone initially forming a 90^{o} vertex angle (propagates 45^{o} from the horizontal both superiorly and inferiorly) indicating possible transition along the lines of principal stress (transition from purely tensile to shear). 12) The only bare bones with a lot of comminution were those that were extremely osteoporotic or loaded axially at high speeds (e.g. a knee impact or those that were crushed between two hard surfaces). 13) Many oblique fractures also have tensile wedge patterns that are not detected by x-ray. 14) Fractures resulting from 7.5 m/s impacts can be quite serious, that is they cause significant injury. This conjecture is also supported by research pertaining to pedestrian injury and vehicle design by Pritz and Hassler (1975). 15) Pritz and Hassler also reported no noticeable differences in injury severity associated with cylindrical impactor radius changes from 1-inch to 4-inches. This is consistent with the findings in this study. 16) Comminuted fractures can occur without entrapment (crushing injury). For 7.5 m/s impacts of intact legs, the inertial restraint of the tibia from the upper thigh and foot is sufficient enough to result in comminuted fractures without any additional support. 17) Age changes in bone can exist, although these changes do not seem to significantly affect fracture patterns (except when compared to babies or small infants). Such changes can include changes to mineral mass, volume, density, and mechanical properties. During dynamic loading situations when ultimate strength is exceeded, bone basically fails as a brittle material (young or old). So, the fractured patterns do not vary too much, unless severe osteoporitic changes have occurred. Such osteoporosis can increase the incidence of high comminution (shatter). 18) For impact loading of the long bone shaft, arthritic changes did not seem to affect the resultant fracture pattern of the entire bone. In other words, a fair supposition would be that arthritis only affects failure patterns when they involve joints. 19) The use of embalmed tissue in the study of the biomechanics of trauma will likely lead to a reasonable determination of impact forces; however, soft tissue damage may be understated and fracture patterns may show greater fragmentation than unembalmed specimens.

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