THE EFFECT OF ACTIVE MUSCLE TENSION ON THE AXIAL INJURY TOLERANCE OF THE HUMAN FOOT/ANKLE COMPLEX

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ABSTRACT

Axial loading of the foot/ankle complex is an important injury mechanism in vehicular trauma that is responsible for severe injuries such as calcaneal and tibia pilon fractures. Axial loading may be applied to the leg externally, by the toepan and/or pedals, as well as internally, by active muscle tension applied through the Achilles tendon during pre-impact bracing. In order to evaluate the effect of active muscle tension on the injury tolerance of the foot/ankle complex, blunt axial impact tests were performed on 44 isolated lower legs with and without experimentally simulated Achilles tension. The primary fracture mode was calcaneal fracture in both groups, but tibia pilon fractures occurred more frequently with the addition of Achilles tension. Acoustic emission demonstrated that fracture initiated at the time of peak local axial force. A survival analysis was performed on the injury data set using a Weibull regression model with specimen age, gender, body mass, and peak Achilles tension as predictor variables. A closed-form solution was developed to predict the risk of fracture to the foot/ankle complex in terms of axial tibia force. Several potential injury assessment reference values are presented for different ages, body types, and injury risks.

INTRODUCTION

Lower extremity injuries sustained by survivors of automobile crashes are common and disabling (Crandall et al., 1996a). After the head, the lower extremity is the most commonly injured region of the body, comprising approximately 30% of all moderate to severe injuries resulting from frontal collisions (Morgan et al., 1991; Pattimore et al., 1991). Injury to the foot/ankle complex accounts for 30-40% of all lower extremity injuries (Morris et al., 1997; Morgan et al., 1991; Pattimore et al., 1991), and up to 10% of all reported injuries in automobile crashes (Crandall et al., 1996a). Furthermore, the most severe injuries suffered in a crash are often to the lower extremity (Pattimore et al., 1991), and these injuries may be the most common cause of long-term impairment and disability (States, 1986).

Several investigators have published detailed breakdowns of the frequency of specific below-knee fractures sustained in motor vehicle crashes (Figure 1) (Pattimore et al., 1991; Taylor et al., 1997; Sherwood et al., 1999; Parenteau et al., 1995; Crandall et al., 1995; Dischinger et al., 1994). These studies generally agree that malleolar, midfoot and forefoot fractures are the most common below-knee fractures. Fractures of the calcaneus and tibia plafond (or pilon) are also relatively common and are particularly disabling because they usually involve disruption of the articular surface of a weight-bearing joint. Because joint cartilage is not well vascularized, healing is often poor, resulting in long-term complications such as infection, malunion, and osteoarthritis. For that reason, calcaneal and tibia pilon fractures are recognized as among the most important below-knee fractures sustained in car crashes (Morris et al., 1997).

Many investigators have attempted to identify the injury mechanisms responsible for real-world lower extremity injuries by investigating crashes using medical records and accident reconstruction analysis. From a biomechanical perspective, the injury mechanism is defined by the principal direction of force applied to the injured segment in relation to the standard anatomical axes of the body (Morris et al., 1997). Axial loading is reported to be a prevalent mechanism of injury in the real world, accounting for 41% (Crandall et al., 1995) to 82% (Fildes et al., 1995) of all lower extremity

injuries sustained in frontal crashes. Furthermore, the consensus among investigators is that the most severe injuries, such as calcaneal and tibia pilon fractures, are caused primarily or solely by axial loading (Pilkey et al., 1994; Morris et al., 1997; Sherwood et al., 1999). This claim is based on the anatomical position of the calcaneus and tibia, which are situated directly along the axial loading path of the lower extremity (Figure 1).

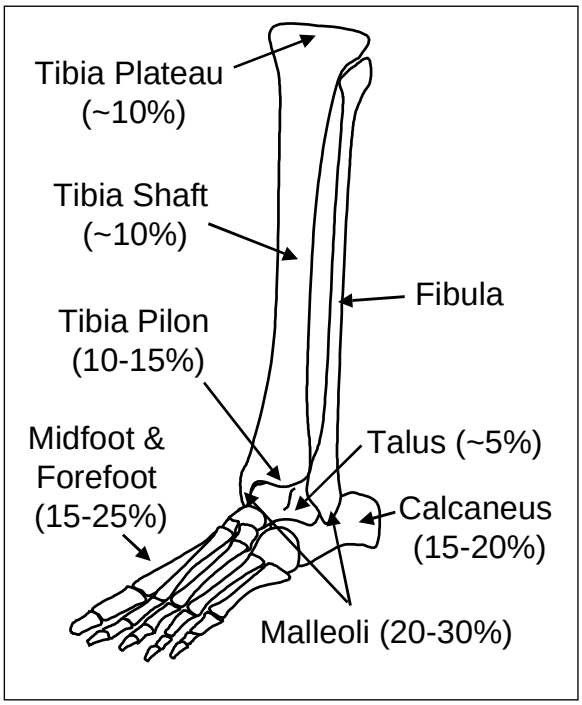

Figure 1. Frequency of below-knee fractures sustained in frontal crashes.

Because axial loading is a common injury mechanism responsible for the most severe foot and ankle fractures seen in frontal crashes, the objective of this study was to determine the tolerance of the human foot/ankle complex to blunt axial impact loading. The purpose of obtaining this biomechanical data was to formulate an injury criterion in terms of engineering parameters that could be measured by a dummy in a vehicle crash test. The injury criterion developed in this study is meant to protect against foot and ankle fractures caused by blunt axial impact loading applied through the plantar surface of the foot.

Previous Work

Several experimental studies have been performed to investigate the injury tolerance of the human lower extremity to dynamic axial loading. Yoganandan et al. (1996) reported results from lower extremity axial impact tests conducted at three institutions: the Medical College of Wisconsin (MCW), Wayne State University (WSU), and the Calspan Corporation. MCW conducted axial impact tests on lower extremities that were disarticulated at the knee. Injuries were generated in 13 out of 26 tests, and included 8 calcaneal fractures, 2 talar fractures, 3 tibia pilon fractures, and 2 unspecified tibia fractures. The average peak force measured at the proximal tibia was 7.8 kN in tests with injury.

WSU tests were conducted using a methodology similar to MCW. Injury was reported in 7 specimens, and included calcaneal and tibia pilon fractures. The average peak force in these tests was 7.6 kN measured at the mid-shaft of the leg (Begeman and Aekbote, 1996). Calspan conducted tests with the ankle positioned in 20º of dorsiflexion. Injury was generated in 9 tests and included 5 calcaneal fractures, 1 fracture of the articular surface of the tibia, 3 unspecified tibia fractures, 4 talar fractures, 4 malleolar fractures, and 5 fibula fractures (Roberts et al., 1993). The average peak force measured at the footplate was 9.8 kN.

Lower extremity axial impact tests were also performed by Klopp et al. (1997) at the University of Virginia. Limbs were sectioned at mid-femur and mounted in a position approximating typical driver geometry with a variety of different initial ankle and knee joint positions. Hard tissue injuries were produced in 11 out of 50 tests and consisted of 5 calcaneal fractures, 1 pilon fracture, 2 talar fractures, and 3 malleolar fractures. The average peak force measured in the mid-shaft of the tibia was 3.8 kN.

Muscle Tension

Based on the above studies, it is apparent that calcaneus fracture is the primary fracture mode in direct axial impact loading of the lower extremity without muscle tension. The lack of active muscle tension is a limitation of cadaver studies. In real world crashes, up to two thirds of occupants suffering lower extremity injuries may be tensing their leg muscles just prior to impact (Ore, 1992). The magnitude of muscular tensing during pre-impact bracing or braking can be quite high. Armstrong et al. (1968) showed that bracing volunteers could exert over 4 kN of axial force through their feet. Driver simulation trials on volunteers report mean peak brake pedal forces of 630 N (Owen et al., 1998), 750 N (Manning et al., 1997), and 796 N (Palmertz et al., 1998) during emergency braking. This braking force is generated by the proximal musculature of the lower extremity, and balanced by the ankle plantarflexors of the leg. Ankle plantarflexion is accomplished primarily by the triceps surae muscle group, which acts through the Achilles tendon (Ferris

et al., 1995). Assuming a 12:5 moment arm ratio between the first metatarsal head, the ankle joint center, and the insertion of the Achilles tendon, an estimated Achilles tendon force of 1.5 kN – 2 kN was required to achieve the measured brake pedal forces (Manning et al., 1997).

Active muscle tension resulting from pre-impact bracing may have a profound effect on the internal loading distribution in the foot/ankle complex. First, Achilles tension compressively loads the distal tibia, but not the calcaneus. Second, Achilles tension applies a plantarflexing moment about the ankle joint that must be balanced by an increased forefoot force and reduced heel force. By both of these means, Achilles tension will tend to protect the calcaneus, but endanger the tibia.

Although several studies have examined the effect of Achilles tension on the sub-failure biomechanics of the foot/ankle complex (Ferris et al., 1995; Manning et al., 1997), only a few have investigated the role of Achilles tension in foot and ankle injuries. Kitagawa et al. (1998) used a test setup similar to Begeman and Prasad's (1990) forced dorsiflexion tests. The Achilles tendon was gripped using a "tendon catcher" and tensioned to 2 kN. A constant force profile was chosen based on the observation that the impact duration in a car crash is much shorter than the muscle stretch reflex time in a living human, which is about 100 msec (Freedman and Herman, 1975).

The injury mechanism investigated by Kitagawa et al. (1998) was not pure axial loading, but rather a combination of axial loading and forced dorsiflexion. Impact was delivered by a pendulum, which struck the footplate 50 mm anterior to the centerline of the tibia axis. The footplate rotated during impact, which dorsiflexed the ankle and loaded the calcaneus and energy absorber in tension. Kitagawa et al. (1998) reported injury in 15 out of 16 tests, including 5 pilon fractures, 10 calcaneal fractures, 1 medial malleolar fracture, and 2 talar fractures. All calcaneal fractures were deemed primarily tension-type injuries, and none were produced in conjunction with a pilon fracture. Average peak forces measured at the tibia were 7.3 kN for pilon fractures and 8.1 kN for calcaneal fractures. Average peak forces measured at the footplate were more than 2 kN lower, suggesting that Achilles tension could endanger the overall foot/ankle complex by placing the distal tibia at risk for fracture at a lower externally applied load.

McMaster et al. (2000) delivered local axial impacts to below-knee cadaver legs using a 5 cm high impactor head. Active muscle tension was simulated by applying 1.5 – 2.5 kN of tension to the Achilles tendon using a hydraulic actuator. Local impacts were delivered parallel to the long axis of the tibia at three locations on the bottom of the foot. Injuries were generated in 16 specimens, and included 9 calcaneal fractures, 1 talar neck and 2 talar body fractures, 3 pilon fractures, 2 malleolar fractures, and 3 soft tissue injuries. Injuries varied according to impact position, with calcaneal fractures predominating when impact was nearer the heel, and pilon, malleolar, and talar fractures occurring more frequently at anterior impact locations. Average failure loads were 5.7 kN at the impactor and 6.4 kN at the proximal tibia.

Kitagawa et al. (1998) and McMaster et al. (2000) demonstrated that muscle tension plays a role in the injury mechanism for tibia pilon fractures. However, both studies were only able to generate pilon fractures by directing the impact to the midfoot instead of the calcaneus. However, compressive loading of the calcaneus is probably ubiquitous in severe frontal crashes. Volunteer studies have shown that many drivers have their heel on the floor during panic braking (Crandall et al., 1996b), and computer modeling has shown that regardless of the initial position of the heel, contact between the heel and the floorpan is inevitable during a severe crash (Pilkey et al., 1994). A more realistic loading scenario would therefore be a blunt impact to the entire plantar surface of the foot (Yoganandan et al., 1996; Begeman and Aekbote, 1996; Klopp et al., 1997). In addition, Kitagawa et al. (1998) and McMaster et al. (2000) were not able to compare injuries and failure loads from specimens subjected to impact with and without a muscular preload. Therefore, more research needs to be conducted to determine how muscular preloading affects the injury tolerance of the foot/ankle complex to blunt axial impact loading.

Time of Fracture

In a dynamic biomechanical test meant to characterize injury, it is important to determine the exact time of fracture. Often, fracture is assumed to occur at the time of peak force. However, if injury initiates before or after the time of peak force, then the peak force value will overestimate the failure strength of the bone. Accurately determining the time of fracture therefore has important implications for data analysis and injury criteria development. A common method of studying fracture is to analyze the acoustic signals emitted by a material as cracks initiate and propagate. This science, called acoustic emission (AE), is well developed for isotropic structural materials, but has been relatively unused with biological materials (Kohn, 1995).

Briefly, acoustic emission theory states that microfractures in the bone create a burst of acoustic

signal due to the release of strain energy. A major fracture will result in a rapid succession and accumulation of these stress waves, which will translate into a higher amplitude AE signal. Quasistatic AE studies on human and animal bones report low-amplitude AE during subfailure loading, followed by high-amplitude burst AE beginning near the time of fracture (Fischer et al., 1986; Kohn, 1995). The interpretation of these results is that low levels of microcracking in the bone are physiological, but beyond a certain threshold, high levels of microcracking become pathological (Kohn, 1995). Only one previous study, Allsop et al. (1988), has presented acoustic emission data from bone during dynamic impact loading. In their study on facial fracture, it was reported that fracture initiation was associated with the onset of an acoustic burst that corresponded to a discontinuity in the slope of the force-time curve. This occurred at 50-100% of the peak force level. However, Allsop et al. (1988) did not present any validation of their technique by comparing injury and non-injury tests. McMaster et al. (2000) stated that they used acoustic sensors in their study, but they presented no data.

METHODS

Human lower limb specimens were obtained from medical cadavers in accordance with ethical guidelines and research protocol approved by the Human Usage Review Panel, National Highway Traffic and Safety Administration, and a University of Virginia institutional review board. Prior to testing, all specimens were screened for HIV and hepatitis, and x-rays were checked for signs of pre-existing bone and joint pathology. All specimens were sectioned above the knee at mid-femur to preserve the functional anatomy of the knee joint and leg musculature.

A test apparatus was constructed to deliver dynamic axial impact loads to the plantar surface of the foot of a cadaver specimen (Figure 2). A compound pendulum or pneumatic impactor struck a padded transfer piston that directed the impact to pure horizontal translation of up to 16 cm. For the majority of tests, this translation was limited to 6 cm. The transfer piston was rigidly attached to a footplate via a 5-axis load cell. During impact, peak footplate velocity was approximately 5 m/s. Leg specimens were placed horizontally in the test rig with the foot neutrally positioned on the footplate and the knee flexed 90º and constrained in an adjustable block. The knee block was attached to the test rig via a 6-axis load cell. The femur was tied back to a uniaxial load bolt to prevent flexion of the knee during impact. Foam padding was placed between the foot and the footplate (19 mm thick, $E \cong 15$ MPa), and around the knee (25 mm thick, $E \cong 25$ MPa) for load distribution.

Specimens were instrumented with an implanted 5-axis tibia load cell. A mounting jig ensured that the length, rotation, and alignment of the bone ends were preserved while a 9 cm portion of the tibia diaphysis was removed and a load cell was installed in its place (Funk et al., 2000). The fibula was left intact. Two cubes with tri-axial magneto-hydrodynamic (MHD) angular rate sensors and tri-axial accelerometers were mounted to each specimen. Acoustic sensors (Pico and Nano 30, Physical Acoustics, Princeton Junction, NJ) were glued with a cyanoacrylate adhesive to the bone surface of the distal anterior tibia and/or medial calcaneus in order to detect fracture time. The acoustic sensors had an operating range of 125-750 kHz with a center frequency of approximately 140 kHz.

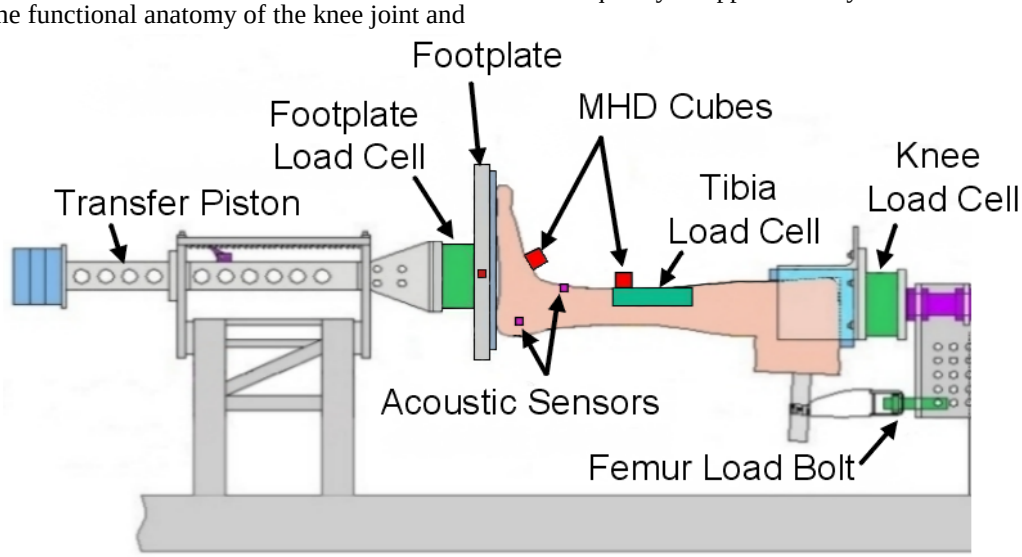

Figure 2. Schematic of the test apparatus used to deliver axial impact loads to cadaveric lower extremities.

In approximately half the tests, active triceps surae muscle tension was simulated by applying tension to the Achilles tendon. For these tests, the Achilles tendon was wrapped in gauze, which was sutured to the tendon to reduce slippage. The tendon was gripped by a device consisting of toothed cams that tightened their grip as the tensile load increased (Figure 3). The gripper assembly had a mass of 0.26 kg and could be easily attached without cutting the tendon. The tendon gripper was connected to an energy absorber by a rope looped around a pulley attached to the footplate (Figure 4). The energy absorber consisted of two aluminum strips that tore at an approximately constant force. Two different thicknesses of aluminum were used in this study to achieve Achilles tension values of either 1.7 kN or 2.6 kN. As the footplate intruded, tension was generated in the rope, causing the aluminum strips to tear. In this way, a blunt, purely axial impact generated tension in the Achilles tendon while simultaneously loading the entire plantar surface of the foot.

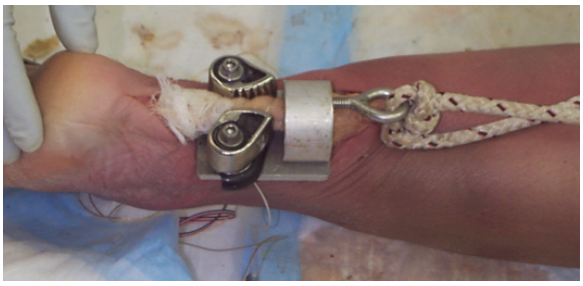

Figure 3. Picture of Achilles tendon gripping device.

The approach taken in this study was to subject the specimens to a high level of impact energy so that injury would be generated in every test. A few non-injury tests were also conducted in order to validate the acoustic sensors. In all tests, the entire leg was externally precompressed through the knee to simulate the effect of proximal lower extremity musculature and loading by the dash (Morgan et al., 1991). In tests without Achilles tension, the leg was externally precompressed to approximately half of the specimen's body weight immediately prior to impact. For tests with Achilles tension, the tendon was pretensioned as much as possible prior to impact. Typically, the Achilles tendon could be pretensioned to a static level of approximately 1 kN, which required an external precompression of approximately 1 kN to maintain heel contact with the footplate. Additional contact padding was added to the transfer piston in order to duplicate the magnitude and onset rate of the loading pulse seen in tests without Achilles tension.

All electronic data except for the acoustic sensors were sampled at 10,000 Hz using a DSP TRAQ-P data analysis system and digitally filtered to SAE J211 channel class 180. Acoustic data were sampled at 5 MHz using a digital oscilloscope, then filtered and processed to calculate counts above a threshold voltage. Video data were taken from each test using either a Kodak Ekta-Pro high speed (1000 fps) monochrome video camera or a high-speed (1000 fps) Kodak RO color imager. All data were transformed when appropriate to the local body segment coordinate frame using the SAE sign convention (positive x, y, and z axes point anterior, right, and inferior, respectively). Load cell data were debiased using offsets recorded in an unloaded state immediately prior to initial positioning. A tri-axial accelerometer array mounted within the fooplate was used to inertially compensate footplate loads to account for the mass of the accelerating footplate. For tests with Achilles tension, the Achilles load was subtracted from the axial force measured by the footplate load cell in order to obtain the applied footplate load.

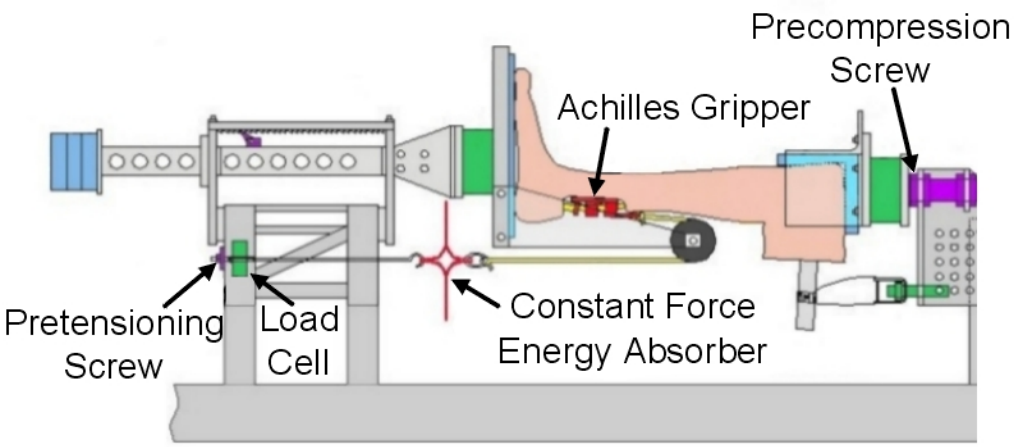

Figure 4. Schematic of the modified test apparatus used to experimentally simulate active muscle tension.

After each test, an orthopaedic surgeon evaluated post-test x-rays and performed detailed necropsies of each specimen to assess the injuries sustained during testing. Dual-energy x-ray absorbtiometry (DEXA) was used to determine the bone mineral content of the mid-diaphyseal portion of each tibia that was removed during load cell implantation. The bone mineral density of each specimen was estimated by dividing the bone mineral content by the cross-sectional area.

Data from tests in which a hard tissue injury was produced in the foot/ankle complex were statistically analyzed using the method of survival analysis with a Weibull cumulative hazard function. Survival analysis is commonly used in medical studies to predict time to death (Collett, 1994). In this study, survival analysis was used to predict force to fracture. Based on the AE results, it was determined that fracture occurred at the time of local peak axial force. Peak axial tibia forces from injury tests were therefore considered uncensored data. A number of parameters were investigated as predictor variables in the regression model, including specimen age, gender, body mass, tibia bone mineral density, and peak Achilles force. Gender was an indicator variable assigned the value of 0 for female and 1 for male. From these predictor variables, the best model was chosen based on the overall correlation coefficient of the model, the significance of each predictor variable, and the degree of cross-correlation between predictor variables.

RESULTS

Fractures of the foot/ankle complex were documented in 15 specimens tested without applied Achilles tension, and in 15 specimens tested with applied Achilles tension (Table 1). Other specimens suffered either no injury or only mid-shaft tibia fractures, which were considered artifactual because they occurred at the interface of the bone and potting material (Appendix 1). Among injured specimens, overall peak forces in the tibia ranged from 2.6 kN to 10.8 kN. A wide variety of foot and ankle injuries were produced, including 25 calcaneal fractures, 7 tibia pilon fractures, 9 talar fractures, 4 medial malleolar fractures, 8 lateral malleolar fractures, 2 distal fibula fractures, 2 navicular fractures, 2 ankle ligament tears, and one Achilles tendon rupture. In addition, injuries to more proximal structures occurred in these same specimens, including 12 tibia plateau fractures, 5 fibular neck fractures, 3 anterior cruciate ligament tears, 1 posterior cruciate ligament tear, and 2 femoral condyle fractures. X-rays of typical injuries show grossly comminuted fractures with multiple extensions into the articular joint surface (Figure 5).

The average peak Achilles force was 1.8 kN in tests with simulated muscle tension. Time histories of the axial force data show that the level of Achilles tension remained fairly constant during the time in the impact event when peak forces occurred at the footplate and tibia (Figure 6).

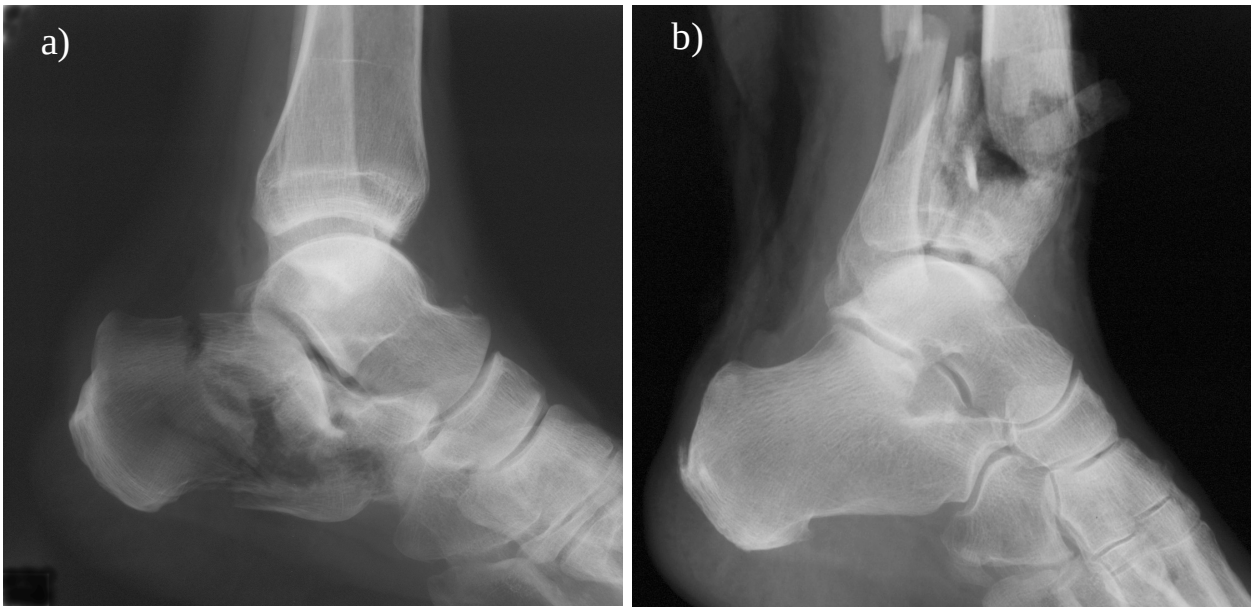

Figure 5. X-rays of typical injuries showing calcaneus fracture (a) from test 5I and tibia pilon fracture (b) from test 6F.

Table 1. Summary of data for all tests sustaining injury to the foot/ankle complex. Injuries are coded as follows: C = Calcaneus fx, P = Pilon fx, T = Talus fx, M = Medial malleolus fx, L = Lateral malleolus fx, F = Fibula fracture, Pl = Plateau fx, and A = Artifactual mid-shaft tibia fx

Test #	ID #	Age (yrs)	Sex	Height (cm)	Mass (kg)	BMD (g/cm^2)	Footplate Fz (N)	Tibia Fz (N)	Achilles Fz (N)	Hard Tissue Injuries							
										C	P	T	M	L	F	Pl	A
3B	104-L	67	F	163	63.6	1.239	8507	4106	0	X			X	X			
3D	76-L	47	M	178	52.3	1.295	5678	4463	0	X			X		X		
4B	17-L	74	F	160	60.0	0.724	4346	2574	0	X							
4C	17-R	74	F	160	60.0	0.724	5027	3827	0	X	X	X			X		X
4F	104-R	67	F	163	63.6	1.239	7785	4685	0	X		X		X			
4G	100-R	42	F	168	71.4	1.208	8546	7349	0				X				X
5A	88-L	59	M	170	47.7	1.256	7279	6854	0	X		X				X	
5C	101-R	62	F	168	52.3	1.192	6196	3221	0	X						X	
5D	88-R	59	M	170	47.7	1.256	5505	4801	0	X		X					
5F	101-L	62	F	168	52.3	1.192	7783	4644	0	X						X	
5G	93-R	67	M	191	80.5	1.270	10177	6334	0	X		X		X			
5H	93-L	67	M	191	80.5	1.270	9288	5829	0	X		X					
5I	102-R	65	M	188	84.1	1.427	7331	5506	0	X						X	
5J	102-L	65	M	188	84.1	1.427	8559	5824	0	X						X	
6C	111-R	67	M	175	70.5	1.340	10344	5185	1854		X					X	
6D	103-L	67	M	175	73.6	1.306	10439	6248	1946	X		X					
6E	105-L	57	M	173	74.5	1.383	11935	9312	1861	X							X
6F	106-R	72	F	178	66.8	1.137	4278	3912	1257		X						
6J	5-R	62	M	173	75.9	1.516	8568	8606	1903	X						X	
6K	87-L	74	F	163	78.2	1.015	7292	7486	1669	X				X			
6L	62-L	41	F	160	89.5	1.179	8703	5577	1796	X							X
6M	87-R	74	F	163	78.2	1.015	7895	5875	1730	X				X		X	
6N	5-L	62	M	173	75.9	1.516	9305	8870	1791	X							
6O	117-R	71	M	170	87.7	1.232	7950	5525	1785	X						X	
7A	108-R	65	F	157	46.8	0.944	5243	2846	1447	X	X	X	X	X			
7B	90-L	69	F	145	57.3	0.897	7321	4302	1548	X							X
7D	2-L	51	M	178	84.1	1.278	10654	10837	2688		X			X		X	
8A	108-L	65	F	157	46.8	0.944	3786	2940	0	X	X	X		X			
8B	90-R	69	F	145	57.3	0.897	6561	3835	2605	X						X	X
8D	2-R	51	M	178	84.1	1.278	11305	10550	1716		X					X	

In tests without Achilles tension, calcaneal fracture was the primary fracture mode, occurring in 14 out of the 15 injured specimens. In tests with Achilles tension, calcaneal fracture remained the dominant fracture mode, but pilon fracture was more common, occurring in 5 of the 15 injured specimens. Generally, calcaneal and pilon fractures did not occur together in the same specimen. However, three specimens did suffer pilon fractures in conjunction with calcaneal fractures. Two of these occurred in tests without Achilles tension, and one occurred in a test with Achilles tension. All three specimens sustaining combined calcaneal-pilon fractures came from osteopenic older females and were associated with very low failure loads.

The time of fracture could be precisely determined from the AE data. In tests with injury, the AE signal exhibited a sudden high-amplitude

burst (10-15 V) with an onset time corresponding to the time of the local peak axial force. When the calcaneus was fractured, the acoustic burst initiated at the time of peak footplate force (Figure 7). When tibia pilon fracture was the only injury to the foot/ankle complex, the onset of the AE burst occurred at the time of peak tibia force (Figure 8). In tests with no injury, the AE signal remained at a low amplitude (< 2 V) throughout the event (Figure 9).

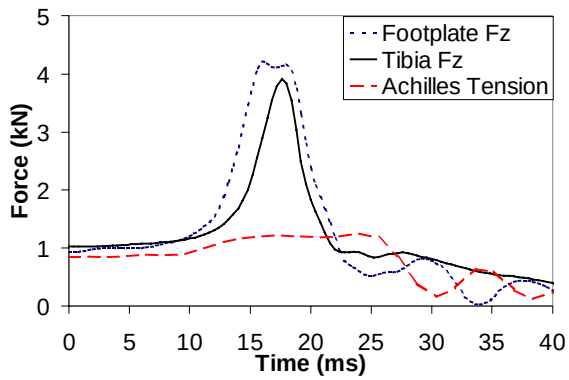

Figure 6. Representative time histories of axial forces in a test with Achilles tension (6F).

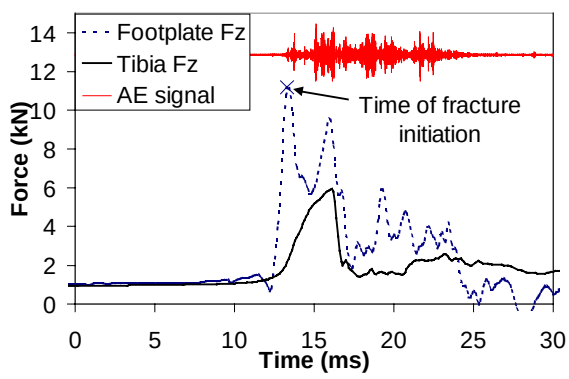

Figure 7. Representative axial load time histories and acoustic emission in a specimen sustaining a calcaneus fracture (6L).

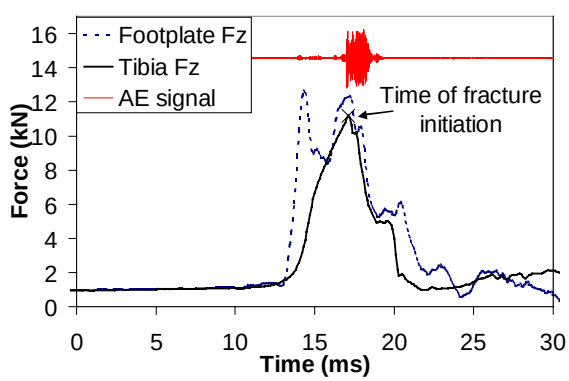

Figure 8. Representative axial load time histories and acoustic emission in a specimen sustaining a tibia pilon fracture (8D).

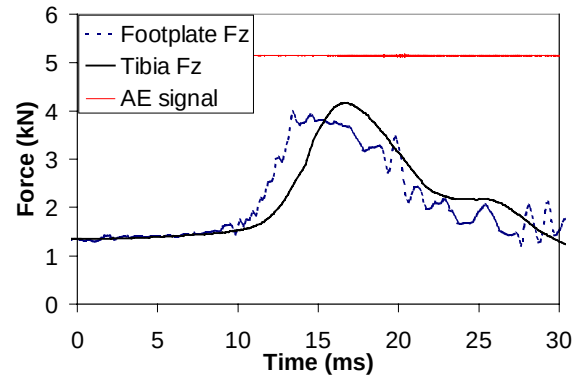

Figure 9. Representative axial load time histories and acoustic emission in a specimen sustaining no injury (6I).

A multivariate Weibull model using age, gender, body mass, and peak Achilles force as predictor variables was found to best represent the peak unscaled axial tibia force data in this study. The model proved to be an excellent fit of the data (standard R^2 = 0.90, validation R^2 = 0.81). Tibia bone mineral density was found to be an excellent predictor variable, but was strongly correlated with age, gender, and mass. Because age, gender, and mass were felt to be more useful and accessible as predictor variables, tibia bone mineral density was dropped from the model. All predictor variables were statistically significant at the $\alpha = 0.05$ level, except for Achilles tension ($p = 0.08$). The following closed-form solution for the survivor function was obtained:

$$S(f|x_i) = \exp\{-\exp[4.99*\ln(f) - 43.7 - 0.964*\text{gender} + 0.0793*\text{age (yrs)} - 0.0552*\text{mass (kg)} - 0.473*\text{Achilles tension (kN)}]\} \quad (1.)$$

where $S(f|x_i)$ is the probability that the axial failure force measured in the tibia is greater than f (Newtons), given the vector x_i of predictors. The risk of injury is therefore equal to 1-S. As previously mentioned, gender is coded 0 for female and 1 for male. In this combined injury model, failure was defined as any hard tissue injury sustained in the foot/ankle complex.

Only data from the 30 tests in which foot/ankle fracture was generated were included in the statistical analysis (Table 1). Most specimens that did not sustain foot/ankle fracture sustained an artifactual mid-shaft tibia fracture at the interface of the tibia load cell and the potting material (Appendix 1). The peak axial force at which these fractures occurred was thought to be influenced by the presence of the load cell. These non-injury data points were therefore considered informatively

censored, and were excluded from the statistical analysis.

The effects of individual predictor variables were assessed using the closed-form solution for the survivor function (eq. 1). The effects of mass and gender were examined by comparing the injury risk functions for two standardized body types: the 50th percentile male (78 kg) and the 5th percentile female (49 kg). The effect of age was as important as the effects of mass and gender. For example, the axial tibia force associated with a 50% risk of injury was as much as 2 kN greater for a 45 year-old compared to a 65 year-old (Figure 10). Likewise, active muscle tension exerted through the Achilles tendon increased the axial tibia force associated with a 50% risk of foot/ankle fracture by as much as 2 kN (Figure 11).

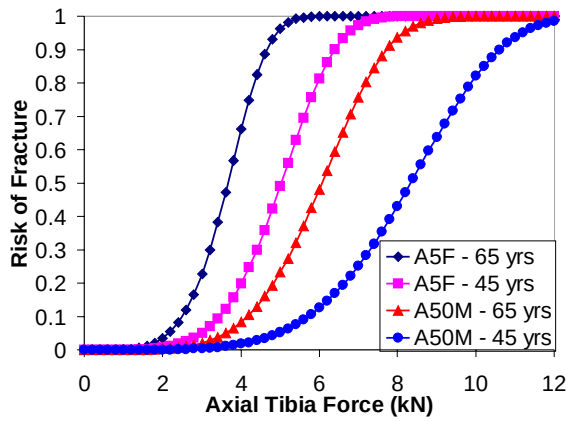

Figure 10. Injury risk functions for the American 5th percentile female (A5F) and American 50th percentile male (A50M) at two different ages assuming no Achilles tension.

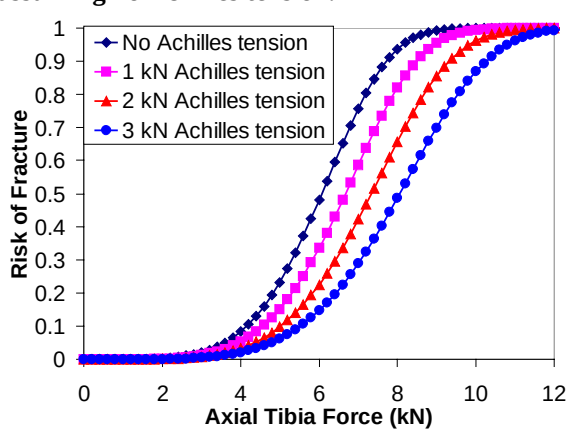

Figure 11. Injury risk functions for a 65 year-old American 50th percentile male (A50M) at varying levels of Achilles tension.

DISCUSSION

The overall purpose of this research was to determine an injury criterion for dynamic blunt axial loading injuries of the foot and ankle. A realistic injury mechanism was investigated whereby a blunt axial impact was applied to the plantar surface of the foot of cadaver specimens. The kinematics of the impact were simplified by neutrally orienting the ankle and constraining the knee in a 90º flexed position. The knee was constrained in order to develop high enough axial loads in the foot/ankle complex to consistently cause injury. The effect of active muscle tension during pre-impact bracing was experimentally simulated by applying approximately 1.8 kN of tension to the Achilles tendon. Clinically realistic axial loading injuries were produced in this study, including calcaneal, talar, and tibia pilon fractures. A survivor function was developed to calculate the risk of injury to the foot/ankle complex in terms of axial tibia force. Specimen age, gender, body mass, and level of peak Achilles tension effectively explained the variation in peak axial tibia force values among the injured specimens.

Acoustic emission data demonstrated that fracture initiated at precisely the time of local peak axial force in this study. The interpretation of this finding is that once fracture initiated, the bone was not able to support any additional axial load. This result is both intuitive and consistent with findings from previous quasistatic AE studies on bone (Fischer et al., 1986; Hasegawa et al., 1993). Based on the AE results, peak axial tibia force data from injury tests were considered uncensored predictors of foot/ankle fracture. This is significant, because many previous axial impact studies of the lower extremity have treated peak axial force data as left-censored, rather than uncensored (Yoganandan et al., 1996; Klopp et al., 1997). Treating peak axial force data from injury tests as left-censored implies that the true failure force was less than the peak force by some unknown amount. However, this study was able to demonstrate, using novel instrumentation, that peak axial force accurately represented the failure strength of the foot/ankle complex. Therefore, the method of statistical analysis used in this study is more accurate than statistical methods from similar studies in which peak axial forces from injury tests were treated as left-censored data. An additional benefit of treating peak axial forces as uncensored data is that uncensored data provides much more information than censored data, thus improving the statistical power of the model (Collett, 1994).

A further advantage of the statistical model derived in this study is that the risk of foot/ankle fracture is described in terms of parameters with known biomechanical effects. The effects of donor age, gender, and body mass on the structural strength of bone are well documented (Yamada and Evans, 1970). Age, gender, and body mass were all found to

significantly and independently affect the axial injury tolerance of the lower limb specimens in this study. No previous lower extremity axial impact study has characterized the effects of specimen age, gender, and body mass. Of course, the effects of these specimen parameters are significant even if they are not characterized. Therefore, results from these other studies must be interpreted in the context of the mean age, body mass, and gender distribution of the specimen pool. This study had the advantage of having a large sample size and diverse specimen population, which allowed the effects of these specimen parameters to be statistically modeled.

The survivor function (eq. 1) estimates injury risk with the greatest confidence when predictor variables are within the range of the data points included in this study. In this study, specimen age ranged from 41 to 74, with a mean age of 63 years. The ages chosen for comparison were 45 years old, which is approximately the average age of the driving population (Cerrelli, 1998), and 65 years old, which is near the mean age of the specimen population in this study. Specimen body masses ranged from 47 to 90 kg, and the gender distribution was equal. Injury risk curves were calculated for the 50th percentile male (78 kg) and the 5th percentile female (49 kg), because these are standard dummy types used in crash testing. The range of ages and body types in the specimen pool therefore encompassed the above populations of interest.

Peak axial tibia loads associated with foot/ankle fracture were generally lower in this study than failure loads reported elsewhere (Roberts et al., 1993; Yoganandan et al., 1996; Begeman and Aekbote, 1996; Kitagawa et al., 1998). There are two factors responsible for these differences: differences in test methodology, and differences in specimen population. In some cases, differences in test methodology are substantial. Kitagawa et al. (1998) reported considerable mechanical noise in their test data, and stated that they filtered their data using a cutoff frequency of 2500 Hz. In this study, mechanical noise was minimal, and all force data were filtered using a cutoff frequency of 300 Hz. These factors may explain why peak axial failure forces were lower in this study. Like this study, Kitagawa et al. (1998) attempted to generate injury in nearly every specimen tested. This contrasts with the approach of Klopp et al. (1997), who generated fracture in only 11 out of 50 specimens tested. Klopp et al. (1997) generally tested with a low level of impact energy, thereby selecting for weaker specimens in the fracture group. This may explain why the mean peak axial failure forces are far lower in the study by Klopp et al. (1997) than any other study.

Another important difference in test methodology between various studies is the location of the axial load measurement. For example, Roberts et al. (1993) reported peak axial loads measured at the footplate. These forces cannot be directly compared to the axial force measured in the tibia, because during a dynamic impact, the inertia of the lower leg may cause the peak footplate force to be 50% higher than the peak tibia force (Yoganandan et al., 1997). For injury criteria applications, tibia force is a more useful parameter than footplate force because it can be measured by an anthropomorphic test dummy.

Other studies that report tibia loads actually potted the tibia and fibula together and measured the combined axial load transmitted through both bones (Yoganandan et al., 1996; Begeman and Aekbote, 1996; Kitagawa et al., 1998). This measurement will naturally be higher than the compressive load measured in the tibia alone. The load-sharing contribution of the fibula has been reported to be 10% in quasistatic axial compression for a neutrally oriented ankle (Crandall et al., 1996a). The advantage of the implanted tibia load cell methodology used in this study was that no artificial boundary conditions were imposed on the leg specimen. Loading was more realistic because the functional anatomy of the knee joint was preserved and relative motion between the tibia and fibula was allowed.

Unlike differences in test methodology, differences in specimen population can be accounted for by using the survivor function developed in this study. If the distribution of failure strengths is assumed symmetric, the 50% risk of injury predicted by the survivor function should correspond to the mean peak failure force for a specimen population of a given mean age, body mass, and gender distribution. For example, the survivor function (eq. 1) predicts a 50% risk of fracture at 7.0 kN for the injured specimen population in the study by Yoganandan et al. (1996). The mean peak axial tibia plateau force reported by Yoganandan et al. (1996) for these specimens was 7.8 kN. Interestingly, this 10% difference can be entirely accounted for by fibula load sharing. Therefore, the results of this study are in good agreement with the results of Yoganandan et al. (1996).

In addition to foot and ankle fractures, a large number of other severe injuries were produced in this study. Tibial plateau fractures were observed in 40% of the specimens suffering a fracture in the foot/ankle complex. However, because tibia plateau fractures did not occur in the absence of foot or ankle injury (except in test 6B), it appears that the proximal tibia is not the “weak link” of the lower extremity in

this mode of loading. An injury criterion based on the tolerance of the foot/ankle complex should be sufficiently conservative to protect against tibia plateau fractures caused by axial loading applied to the foot. Most other studies have not been able to directly compare the tolerance of the foot/ankle complex to the tolerance of the tibia plateau, because they have not tested specimens with both the knee and ankle joints intact (Yoganandan et al., 1996; Begeman and Aekbote, 1996; Kitagawa et al., 1998; Banglmaier et al., 1999).

One goal of this study was to investigate the influence of Achilles tension on the fracture mode of the foot/ankle complex under axial loading. Injuries obtained in this study are consistent with the results from previous studies reported in the literature. In tests without muscle tension, the primary fracture mode was calcaneal fracture. In this group, pilon fractures occurred only rarely, and always together with a calcaneal fracture. These fracture patterns are consistent with previous lower limb axial impact studies conducted without Achilles tension (Yoganandan et al., 1996; Begeman and Aekbote, 1997; Klopp et al., 1997). In this study, calcaneal and pilon fractures occurred together only in osteopenic specimens from older females. This result suggests that the combined pilon-calcaneal fracture mode is a result of an impact that is very severe relative to the strength of the foot/ankle complex.

In tests with an average Achilles tension of 1.8 kN, calcaneal fracture was still the primary fracture mode. However, pilon fractures occurred more frequently in this group (5 out 15 injured specimens) and were typically not associated with calcaneal fracture. The injury distribution obtained in this study by testing with 1.8 kN of Achilles tension was almost identical to the injury distribution reported by Kitagawa et al. (1998), who tested with 2 kN of Achilles tension.

Achilles tension was investigated as a predictor variable in the survivor function (eq. 1) for two reasons. First, Achilles tension was associated with the experimental production of tibia pilon fractures in axial loading. Therefore, an axial loading injury criterion designed to protect against tibia pilon fracture should incorporate Achilles tension in the formulation. Second, Achilles tension was expected to affect the peak axial tibia force associated with fracture. Because of the geometry of the test setup, the axial force measured by the tibia load cell included the contribution of the Achilles tension. As a predictor variable, Achilles tension was not quite statistically significant at the $\alpha = 0.05$ level ($p = 0.08$). However, the sample size for tests with nonzero Achilles tension was only half of the total, so an increased sample size might have lowered its p-value. In the end, peak Achilles tension was included in the model because it had an obvious influence on the peak axial tibia force associated with fracture (Figure 11), and excluding the term would have biased the statistical model.

The finding that Achilles tension increases the peak axial tibia force associated with fracture does not necessarily prove that active muscle tension will protect the foot/ankle complex in a dynamic axial impact characteristic of a car crash. First, the amount of force experienced by the foot/ankle complex in a crash is not prescribed, but rather is a complicated function of parameters such as impact velocity, specimen mass and stiffness, and reciprocal loading due to knee entrapment or muscle force. In sled tests, simulated muscle tension has been shown to increase the stiffness and effective mass of the lower extremity, which tends to increase the efficiency of load transmission through the specimen (Klopp et al., 1995). Therefore, the same crash pulse may result in higher forces in a tensed limb compared to an untensed limb. Biomechanical component tests are only the first step in answering the question of whether active muscle tension due to pre-impact bracing will protect or endanger the foot/ankle complex in a real world crash. Crash tests and finite element modeling of the lower extremity may be useful in analyzing the load transmission through the foot/ankle complex as a function of Achilles tension and impact velocity.

In addition to altering the load transmission properties of the leg, Achilles tension affects the peak axial force measured in the tibia for the purely geometric reason that tibia force includes the contribution of Achilles tension. If the axial failure strength of the foot/ankle complex were to be characterized based on the force applied to the foot, the effect of Achilles tension might be entirely different. For example, Kitagawa et al. (1998) reported that peak axial footplate loads were approximately 2 kN lower than peak axial tibia loads in tests with Achilles tension. In this study, the peak axial load was almost always higher at the footplate compared to the tibia for a given test. This discrepancy between studies is due to differences in specimen fixation. Because the leg specimens were rigidly fixed in the tests by Kitagawa et al. (1998), they could not accelerate, so there was relatively little inertial difference in the axial loads measured at the footplate and the tibia. In this study, the tibia and fibula were not rigidly fixed. The knee was constrained in a block, but some translation was possible due to compression of the padding and soft tissue surrounding the knee. This more realistic type of fixation allowed the leg specimen to accelerate early in the impact event, which created an inertial

difference in the axial loads measured at the footplate and the tibia.

The observation that load transmission through the lower limb can change as a function of specimen fixation, loading rate, and muscle tension points to a limitation of this study. The survivor function developed in this study estimates the risk of fracture based on the axial force measured in the tibia. However, acoustic emission demonstrated that fracture initiation was correlated with peak local axial force. Tibia pilon fractures occurred at the time of peak axial tibia force, but calcaneal fractures occurred at the time of peak axial footplate force. This finding suggests that axial footplate force would be a better predictor of calcaneal fracture than tibia axial force.

The fact that the survivor function does not use axial footplate force to predict injury is a limitation of this study. However, this limitation was deemed necessary, because the purpose of this research was to obtain biomechanical injury tolerance data that could be applied to tests with anthropomorphic test dummies. Unfortunately, current crash test dummies cannot measure the axial force applied to the bottom of the foot or heel. The Hybrid III lower extremity and the Thor-LX are both equipped with load cells situated in the upper and lower tibia. The location of these load cells is similar to the location of the tibia load cell in this study, which was implanted mid-shaft. For that reason, the survivor function derived in this study calculates the risk of foot/ankle fracture using axial tibia force.

Nonetheless, peak axial tibia force was felt to be an effective predictor of foot/ankle fracture in this study. Many previous studies have used peak axial tibia force to describe the fracture tolerance of the foot/ankle complex (Yoganandan et al., 1996; Begeman and Aekbote, 1996; Kitagawa et al., 1998). In this study, AE showed that the amount of load that could be transferred to the mid-shaft of the tibia was limited by and directly related to the fracture tolerance of the calcaneus. Therefore, it is reasonable to assume that a consistent relationship existed between the fracture tolerance of the calcaneus and the peak axial force developed in the tibia under the loading conditions studied. It should be noted that the relationship between peak axial tibia force and peak axial footplate force might change under different loading conditions or ankle orientations.

The intent to apply biomechanical data from this study to anthropomorphic tests dummies raises additional issues. An anatomical difference between the dummy and cadaver lower limb is the lack of a fibula in the dummy. For that reason, it is important to specify whether a dummy biofidelically replicates the axial loading response experienced by the tibia and fibula together, or just the tibia. Preliminary testing indicates that the 50th percentile male Thor-LX accurately reproduces the dynamic axial load response of the tibia in a cadaver (Rudd et al., 1999). However, more testing is required to evaluate the biofidelity of the Thor-LX over a range of loading conditions and test populations.

The biomechanical data in this study indicate that age, gender, body mass, and the level of active Achilles tension all significantly affect the peak axial tibia force associated with fracture of the foot/ankle complex in a cadaver. Anthropomorphic test dummies do not include all of these factors in their design. Test dummies are used as surrogates for people of a specific body mass and gender (i.e. 5th percentile female or 50th percentile male). Although dummies do not necessarily need to be age-specific, the data from this study suggest that injury assessment reference values should be age-specific. The survivor function (eq. 1) was used to calculate potential injury assessment reference values for different dummy types at two different ages and levels of injury risk (Table 2). These injury assessment reference values were estimated assuming no Achilles tension because neither the Hybrid III lower limb nor the Thor-LX is equipped with active musculature at this time. Although the Thor-LX does simulate passive muscle tension acting through the Achilles tendon, the amount of passive muscle tension is negligible when the ankle is in the neutral position, which was the ankle orientation examined in this study. It is anticipated that the survivor function reported here may allow the calculation of injury risk to be modulated according to the measured level of Achilles tension in a future dummy design that does include active muscle tension.

Table 2. Matrix of injury assessment reference values for blunt axial impact loading of the foot/ankle complex assuming no Achilles tension

Dummy type	Age (yrs)	50% risk	30% risk
50th percentile male	45	8.3 kN	7.3 kN
	65	6.1 kN	5.3 kN
5th percentile female	45	5.0 kN	4.4 kN
	65	3.7 kN	3.2 kN

CONCLUSIONS

This study investigated the injury tolerance of the human foot/ankle complex to dynamic axial impact loading. Clinically realistic axial loading injuries were produced, including calcaneal, talar, and tibia pilon fractures. Acoustic emission (AE) demonstrated that fracture occurred precisely at the

time of peak local axial force. Because the biomechanical data was intended for application in crash test dummies, peak axial tibia force was used as a predictor of foot/ankle fracture. Achilles tension was shown to modulate not only the fracture mode, but also the peak axial tibia force associated with fracture of the foot/ankle complex. A survivor analysis of the peak axial tibia force data using a Weibull regression model was developed using donor age, gender, body mass, and level of Achilles tension as predictor variables. The resulting survivor function predicts the risk of fracture to the foot/ankle complex under axial impact loading as a function of axial force measured in the tibia. Several potential injury assessment reference values were presented to address below-knee fractures caused by blunt axial impact loading.

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Appendix 1. Summary of tests in which the specimen did not sustain a fracture in the foot/ankle complex. Artifactual fractures occurred at the interface of the bone and potting material in the mid-shaft of the tibia

Test #	ID #	Age (yrs)	Sex	Height (cm)	Mass (kg)	BMD (g/cm^2)	Footplate Fz (N)	Tibia Fz (N)	Achilles Fz (N)	Comments
3A	104-L	67	F	163	63.6	1.239	3027	2205	0	No injury
3C	95-L	57	F	173	72.7	1.010	7782	3243	0	Artifactual fx only
4A	76-R	47	M	178	52.3	1.295	N/A	N/A	0	Data collection error
4D	98-R	44	F	163	60.0	1.301	6838	5336	0	Artifactual fx only
4E	98-L	44	F	163	60.0	1.301	7353	5682	0	Artifactual fx only
5B	77-R	63	F	160	55.9	1.086	6876	5887	0	No injury
5E	77-L	63	F	160	55.9	1.086	7593	3996	0	Artifactual fx only
6A	91-L	65	F	165	64.5	0.861	5113	3591	1646	Artifactual fx only
6B	105-R	57	M	173	74.5	1.383	10316	6898	1715	Tibia Plateau fx only
6G	106-L	72	F	178	66.8	1.137	3707	2726	1456	Artifactual fx only
6H	111-L	67	M	175	70.5	1.340	5744	5263	936	No injury
6I	103-R	67	M	175	73.6	1.306	3909	4159	2642	No injury
7C	4-R	43	M	175	78.6	1.462	9588	9602	2558	Artifactual fx only
8C	4-L	43	M	175	78.6	1.462	9131	9573	2644	Artifactual fx only