

Biomechanical Response of the Lower Leg under High Rate Loading

Kyvory A. Henderson, Ann M. Bailey, John J. Christopher, Fred Brozoski, Robert S. Salzar

Abstract Severe lower extremity injuries have been observed in military vehicle occupants in an underbody blast (UBB) event. Characterization of how these loading conditions, different from those seen in automobile collisions, affect the response of the lower limb provides valuable understanding of the relevant injury mechanisms from a UBB. This study investigates the effect of a high-rate axial load to post-mortem human surrogates' (PMHS) lower extremities. Eighteen PMHS were tested on a drop-tower test fixture designed to simulate the occupant loading environment and an unconstrained knee boundary condition. Strain-time histories found on the lateral and medial sides of the tibia showed inhomogeneous deformation patterns along the length of bone. The response of the lower leg to high rate loading showed significant local stiffness at the distal tibia-fibula region $k_1 = 2028.38 \pm 525.02 \text{ kN/m}$ and $k_2 = 9058.35 \pm 2043 \text{ kN/m}$. The 50% survivability for lower extremities subject to these loads and load rates was calculated at 7.7kN for the distal tibia and 6.8kN for the proximal tibia. Additionally, an analytical three-mass model of the lower leg was developed for the tested condition and the predicted model fit compared to the measured response.

Keywords High rate loading, lower extremity injuries, underbody blast

I. INTRODUCTION

High-rate axial loading from an underbody vehicular explosion has been identified as one of the most common war time injury mechanisms in the lower limb [1]. The rates at which the lower limbs are loaded in these events are significantly higher than those previously characterized for automotive intrusion. Rate-dependent characterizations of how these loading conditions affect lower limb components will provide information towards understanding injury mechanisms from these elevated loading rates, in addition to providing vehicle designers with injury data necessary to reduce or even prevent these injuries in future vehicle designs. This information will also serve as essential in the development of future biofidelic anthropomorphic test devices.

Medical injury data from theater reveal that 32% of soldiers wounded in underbody blast events sustained foot/ankle fractures, while 16% sustained both foot/ankle and tibia/fibula fractures [2]. For soldiers killed during an underbody blast (UBB) event, 28% sustained foot/ankle fractures, while 27% sustained both foot/ankle and tibia/fibula fractures. This evidence details the vulnerability of lower extremities in UBB events and suggests that the environment is different from what is traditionally seen in automotive intrusion events.

For comparison, lower limb injuries account for 8%-12% of all injuries and foot-well intrusion is often labeled as the primary mechanism of lower limb injuries [3-5]. Injury mechanisms associated with foot-well intrusion include inertial loading, entrapment, excessive motion of the joints and contact with other compartment surfaces. Of these mechanisms, the most severe injuries are sustained from axial loading of the lower limbs. Axial impact tests to the foot/ankle complex found the mean dynamic force of fracture to be 6.8kN [6]. Studies such as these have provided valid injury criteria for the automotive crash environment. Though the average automotive intrusion produces typical toe pan velocities of 5 m/s with a load duration of 10ms at peak accelerations of 50g [6], UBB events have been shown to have accelerations of 100g or more [7].

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Currently there is no validated test methodology for determining the transmission of forces, nor the risk of injuries to the lower limbs due to intrusion of a vehicle toe pan, from a UBB. Since theater data suggest these injuries are a result of axial loading at very high rates and low levels of stroke, one goal of this study is to recreate the environment of an UBB event using PMHS components and to develop an injury criterion for axial loads at these rates. In this study, 18 PMHS lower limbs are subjected to a high rate impulse applied to a foot-pan with acceleration levels of 200 and 600g and durations of 3ms, and the resulting injuries are recorded.

II. METHODS

Eighteen fresh frozen PMHS lower limbs, disarticulated below the patella, were procured for this study and thawed for use (TABLE I). Pre-test DEXA (bone density) and CT images were taken to verify that specimens did not have pre-existing or recently healed fractures, and that the bone mineral density was greater than the threshold between osteopenia and osteoporosis (T-score > -2.5) [8]. All test procedures were approved by the University of Virginia Cadaver Use Committee.

TABLE I
SPECIMEN DATA

Specimen #	Test #	Age	T-Score	Tibia Length (mm)	Body Mass (kg)	Height (cm)
549L	1.1,1.2	39	0.6	360	95	183
549R	1.3,1.4	39	1.7	360	95	183
537L	1.5	54	-1.3	389	76	178
537R	1.6	54	-1.3	389	76	178
544L	1.7	59	-1	404	118	185
547R	1.8	59	-	400	118	185
548L	1.9	53	5.3	432	127	185
543L	1.10	64	-0.1	405	104	183
548R	1.11	53	3.9	428	127	185
540L	1.12	47	-2	383	73	175
543R	1.13	60	0.3	405	104	183
530L	1.14	54	0.1	364.4	117	185
540R	1.15	47	-2.2	377.92	73	175
556R	1.16	60	-	380.39	82	183
556L	1.17	60	-	376.4	82	183
530R	1.18	54	-0.1	363.06	117	185
564L	1.19	53	-1.1	385.21	73	178
564R	1.20	53	-1	369.02	73	178

First removing the soft tissue, the proximal tibia was placed in a steel mating cup and secured with Urethane Pour Foam (US Composites, Inc., West Palm Beach, FL). The specimens were then placed in the UVA drop tower fixture with the proximal mating cup attached to a 6-degree-of-freedom load cell and a load cell plate. The specimens were mounted in the drop tower fixture with the plantar surface of the foot facing vertically up and in contact with a second plate (impact plate) of mass 10.2 kg that was constrained in the X and Y directions, but free to translate in the Z direction (SAE coordinate system). The tibia boundary conditions consisted of a reaction mass comprised of the load cell, load cell plate and the potting cup. This mass totaled 7.2kg, which is equivalent to the effective mass acting at the proximal tibia in a 50th percentile male. This mass was determined through a finite element (FE) simulation of a representative UBB event at 200gs and 600gs, which calculated the forces and accelerations at the proximal tibia. The second boundary condition represented a fully constrained proximal tibia. The stroke of the impact plate was limited by the use of shaft collars (Fig. 1).

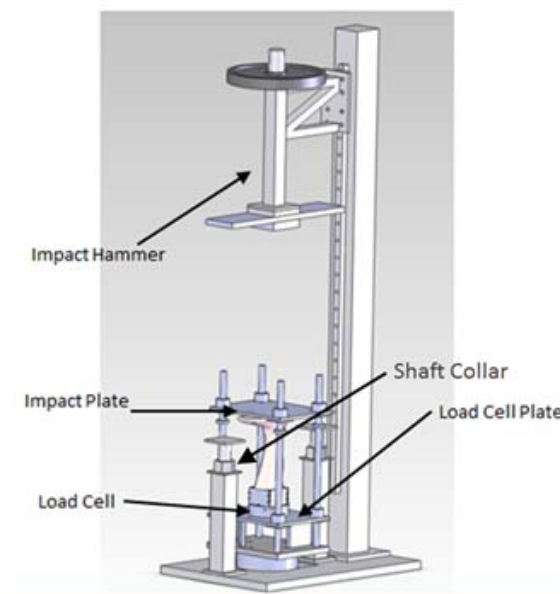


Fig. 1. Drop Tower Schematic

For this series, multiple input conditions were investigated. These conditions were obtained by combining different drop heights, hammer masses and pulse shapers. Drop distances ranged from 1 to 2.3 m. Hammer masses ranged from 38.5 to 61.2kg. Two different pulse shapers, as listed in TABLE III, were used between the drop hammer and the impact plate in these tests to extend and limit the region of interest: (1) a 38 mm thick sheet (155 x 155 mm) of expanded polystyrene ($E=413$ kPa) and (2) a 155 x 155 mm block of crushable aluminum honeycomb ($E=10.3$ MPa) with a 6.4 mm layer of latex.

A combination of load cells, accelerometers, angular rate sensors as well as a laser displacement device for capturing impact velocity and displacement, were used in this study. Load cell, strain and accelerometer data were acquired through a TDAS (Diversified Technical Systems, Inc., Seal Beach, CA) data acquisition system at sampling rates of 20kHz with an anti-aliasing filter of 4kHz. The laser displacement (Keyence America, Elmwood Park, NJ) data were collected by a Synergy CS (Hi-Techniques, Inc., Madison, WI) high speed data acquisition system at a sampling rate of 1 MHz with an anti-aliasing filter of 200 kHz.

Each specimen was instrumented with 12 strain gages. Four were placed in pairs at anterior and medial points of the distal and proximal ends of the tibia. Six strain gages were positioned in a circumferential bone cell configuration at the distal tibia, and a pair of strain gages were placed along the tibia diaphysis and the calcaneus. These gages were applied to the bone using cyanoacrylate glue. Data were collected from four accelerometers. One was placed at the impact plate (x-direction), the load cell plate (x-direction), the proximal tibia (z-direction) and on top of the navicular bone (z-direction). One angular rate sensor was placed at the distal tibia (y-axis) using a circular metal clamp and at the dorsal surface of the navicular bone (y-axis) using wood screws. Accelerometers were also placed on the impact and ballast plates (z-axis). The load cell was located between the potting cup and the reaction mass. All instrumentation is listed in TABLE II. After instrumentation, an additional pre-test CT scan was performed to determine precise sensor location.

TABLE II
INSTRUMENTATION LIST

Channel	Sensor	Location
Acceleration (SAE-Z)	Endevco 7264B-2000	Distal Tibia
Angular Rate (SAE-Y)	DTS-ARS-8k	Distal Tibia
Acceleration (SAE-Z)	Endevco 7264B-2000	Navicular
Angular Rate (SAE-Y)	DTS-ARS-8k	Navicular
Force and Moment	Denton 3868	Proximal Tibia
Acceleration (SAE-Z)	Endevco 7264B-2000	Distal Impact Plate
Acceleration (SAE-Z)	Endevco 7264B-2000	Proximal Plate
Strain	Micro-Measurements C2A-06-062LW-350	Tibia Diaphysis and Calcaneus
Displacement	Keyence LK-G502	Drop Hammer

The bone cell consisted of six strain gages oriented longitudinally along the axis of the tibia, and placed along the circumference of the distal tibia as pictured in Fig. 2. This configuration allowed for determining both the level of compression and the extent of bending experienced at the distal tibia. The bone cell was calibrated using the static weight of the impact plate and the hammer to estimate the modulus of the bone. All strain gages were then zeroed prior to testing. Following the tests, a custom Matlab (The MathWorks, Inc) program was used to determine an effective modulus for the bone. The program used CT images of the cross-section of the instrumented tibia to determine the area centroid of the bone.

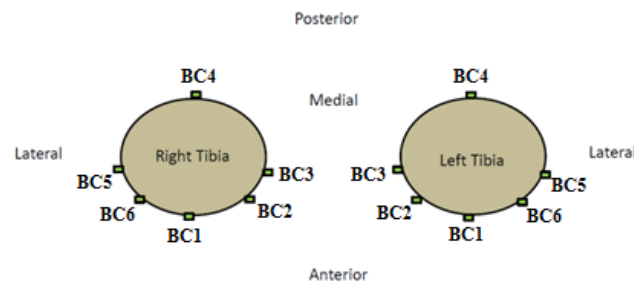


Fig. 2. Bone Cell Configuration

Once a specimen was fully instrumented, it was pre-test CT scanned to determine the exact locations of the instrumentation. As instrumentation location was confirmed, specimens were placed in the drop tower with the impact plate resting statically on the plantar surface of the foot. The hammer was slowly lowered until it was bearing its weight on the impact plate. Triggers were then armed and the hammer was dropped from the pre-determined height.

All data were analyzed in Excel 2010 (Microsoft). Load cells were inertially compensated. Data were recorded at 20KHz and filtered at cfc1000 (SAE J-211). Strain rates, load rate and rise times were determined by locating the time of arrival, beginning at the first sign of transducer output, and the time of the first peak in the data.

A survival analysis was performed on the data from this study using Minitab® (Minitab 16, Minitab Inc.), and compared to previous values found in the literature on the lower extremities. The forces at the proximal end were corrected for inertia effects produced by the movement of the load cell using the recorded acceleration at

the load cell plate. Forces at the distal tibia were found using strain measurements recorded from the bone cell which were assumed to be due to pure axial loading which was also used accurately in reference [9]. The results of this analysis are shown later, where the probability of fracture was calculated based on the proximal tibia force and the distal tibia force. These probabilities are calculated from tests where specimens were not re-tested, and only include tests where the proximal tibia was in a free, unrestrained condition. Values for forces at the distal tibia were calculated from strain values given by the distal tibia bone cell. The forces recorded for the proximal tibia were collected from the load cell against the load cell plate.

Bone cell forces were derived using a methodology similar to that used in [9]. However, the application of additional strain gages as well as the positioning of the strain gages allowed for a more accurate calculation of the force. Instead of being placed at the tibia mid-shaft, the bone cell was placed at the distal tibia in order to decrease the bending contribution due to the curvature of the tibia. The additional number of strain gages in the bone cell enabled a more accurate determination of the neutral axis.

Pre-test CT images of the tibia were used to determine the moments of inertia of the bone cross section at the gage and the positions of the bone cell strain gages. These positions, in addition to the strain-time histories for each gage, were used to determine the position and orientation of the neutral axis for each time step. The main assumption used in this bone cell calculation was that the bone cell, placed distally on the tibia, would primarily record axial load and that minimal bending would be present at that level. The average orientation angle of the neutral axis was calculated for the time prior to the peak strain of the first strain gage and then applied to the remaining time steps. This average orientation was used and the neutral axis was translated in order to make it pass through the geometric centroid of the tibia cross-section. The two strain gages furthest from the centroid on either side of the neutral axis were chosen in order to calculate the axial strain time history Equation 1. Using this axial strain, Young's modulus was calculated using Equation 2 which relates Young's modulus to strain rate, and the cross-sectional area of the tibia at the bone cell location was calculated. Using Matlab and the CT images, the force-time history was calculated.

$$\varepsilon_{axial} = \varepsilon_i + (\varepsilon_j - \varepsilon_i) \frac{m_{N/A} x_i^g - y_i^g}{m_{N/A} x_i^g - y_i^g - m_{N/A} x_j^g + y_j^g} \quad (1)$$

where $m_{N/A}$ is the slope of neutral axis, x and y are the positions of the i^{th} and j^{th} strain gage, and ε are the strains associated with the i^{th} and j^{th} strain gage.

$$E = 29.258(SR)^{0.0441} \quad (2)$$

where E is Young's modulus and SR is strain rate [10]

A three-degree of freedom model with spring and dashpot (Fig.3) was utilized to model the elastic compliance and the dissipative properties of the leg. As it is shown, m_1 is the mass of the foot and ankle to the distal tibia (at accelerometer location), m_2 is half the mass of the remaining lower extremity, and m_3 is the other half of the lower extremity plus the 7.2 kg effective mass of the leg [11]. The stiffness and compliance of the ankle region are represented by k_1 and c_1 and the stiffness and compliance of the tibia/fibula are represented by k_2 and c_2 . The parameters k_1 , k_2 , c_1 , and c_2 were estimated using a least squared optimization method modeled in Matlab ©(Mathworks, Natick, MA). The equation of motion for this system is written in Equation 3.

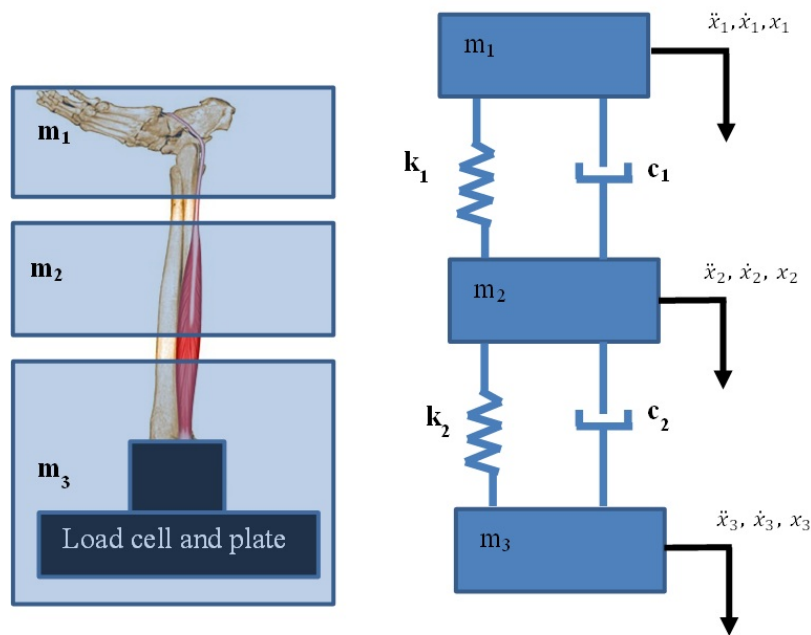


Fig. 3. Schematic of the three mass model.

$$\begin{aligned}
 & \begin{bmatrix} -m_1 & 0 & 0 \\ 0 & -m_2 & 0 \\ 0 & 0 & -m_3 \end{bmatrix} \begin{bmatrix} \ddot{x}_1 \\ \ddot{x}_2 \\ \ddot{x}_3 \end{bmatrix} + \begin{bmatrix} c_1 & +c_1 & 0 \\ c_1 & -c_1 - c_2 & c_2 \\ 0 & c_2 & -c_2 \end{bmatrix} \begin{bmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{x}_3 \end{bmatrix} \\
 & + \begin{bmatrix} -k_1 & k_1 & 0 \\ k_1 & -k_1 - k_2 & k_2 \\ 0 & k_2 & -k_2 \end{bmatrix} \begin{bmatrix} x_1 \\ x_2 \\ x_3 \end{bmatrix} = \begin{bmatrix} 0 \\ c_2 \dot{x}_2 + k_2 x_2 \\ 0 \end{bmatrix}
 \end{aligned}
 \tag{3}$$

where $\ddot{x}'s$ are the acceleration and ($\dot{x}'s$ and $x's$) are the integrated velocity and displacement.

Of the 20 total tests performed, 6 specimens incurred no injury, while 14 cases produced injury. Three of the injurious tests showed fracture of the distal tibia while the other 11 tests produced injuries to the calcaneus, the talus, or both. TABLE III summarizes the loading conditions, the resulting peak forces and acceleration for each test. Preliminary test refer to tests performed with additional boundary conditions and included specimens that may have been repeatedly tested. Primary tests were all performed under similar boundary conditions. Stroke implication tests allowed for greater stroke length in hammer impact.

TABLE III
TEST Conditions

TEST #	HAMMER MASS (kg)	Hammer Drop Height (m)	STROKE	PULSE SHAPER	REACTION MASS (kg)	NOTES
1.1	38.5	2	1"	1	7.2	Preliminary Test
1.2	38.5	2	1"	1	7.2	Preliminary Test
1.3	38.5	2	1"	1	7.2	Preliminary Test
1.4	38.5	2	1"	2	7.2	Preliminary Test
+1.5	38.5	2	1"	1	∞	Primary Test
1.6	38.5	2	1"	2	∞	Primary Test
+1.7	61.2	2	1"	2	∞	Primary Test
*1.8	61.2	2	1"	2	7.2	Primary Test
+*1.9	61.2	2	3"	2	7.2	Stroke Implications Test
+*1.10	61.2	2	1"	2	7.2	Primary Test
+*1.11	61.2	2	1"	2	7.2	Primary Test
+1.12	61.2	2	1"	2	7.2	Primary Test
+*1.13	61.2	2	1"	2	7.2	Primary Test
+*1.14	34.2	2.3	1"	2	7.2	Primary Test
+1.15	34.2	2.3	1"	2	7.2	Primary Test
+1.16	34.2	1.5	1"	2	7.2	Primary Test
+1.17	34.2	1	1"	2	7.2	Primary Test
+*1.18	34.2	1.4	1"	2	7.2	Primary Test
+*1.19	34.2	1.25	1"	2	7.2	Primary Test
+*1.20	34.2	1.25	1"	2	7.2	Primary Test
+ (Test was used in survivability analysis) three mass model development)					* (Test was used in	

III. RESULTS

In the 20 tests that were performed, various load rates, rise times and accelerations were reached, all of which could be quantified as blast rates. The results are summarized in Table IV. For the tests used in the survivability analysis, the mean impact plate acceleration was $422.05 \pm 130.622g$, the mean impact plate rise time was $3.3175 \pm 2.56ms$, the mean proximal tibia load rate was $5.15 \pm 4.17kN/ms$, and the mean strain rate located at the mid-tibia was $2272.289 \pm 1171.87u/ms$.

Fig. 4 shows the results of the survival analysis performed for the distal tibia force. The 50% survivable probability for the distal tibia under these loading rates was found to be 8.91kN. The 75% survivable probability was found to be 6.41kN (Table V).

Fig. 5 shows the results of the survival analysis performed for the proximal tibia force. The 50% survivable probability for the distal tibia under these loading rates was found to be 6.41kN. The 75% survivable probability was found to be 4.86kN (Table VI).

Table IV
Test Results

TEST #	IMPACT PLATE ACCELERATION (g)	IMPACT PLATE RISE TIME (ms)	HAMMER VELOCITY (m/s)	Prox Tibia Load Rate (kN/ms)	Strain Rate (u/ms)
1.1	181	9.8	5.73	1.3	872.3
1.2	178	7.5	5.88	0.3	417.14
1.3	200	8.5	5.63	8	5762.6
1.4	450	4	4.18	4	2635.1
1.5	200	7.2	5.93	3.5	1279.96
1.6	400	2.1	5.14	19	2432.82
1.7	490	2	4.66	11	2218.97
1.8	530	2.4	5.73	6	1597.71
1.9	530	1.6	5.16	-	3587.77
1.10	510	1.75	4.09	7	3072.69
1.11	430	2.2	5.57	8.4	-
1.12	460	1.5	5.60	3.3	3219.58
1.13	540	2.5	5.35	3.4	3070
1.14	570	1.7	5.46	3.3	1586.83
1.15	600	1.8	-	3.5	2020.66
1.16	520	1.9	4.98	2.7	1542.8
1.17	340	2.3	4.44	1.3	-
1.18	480	1.8	4.93	3.3	1723
1.19	430	2	4.66	4.5	2052.37
1.20	420	2.1	4.70	4	1808.9

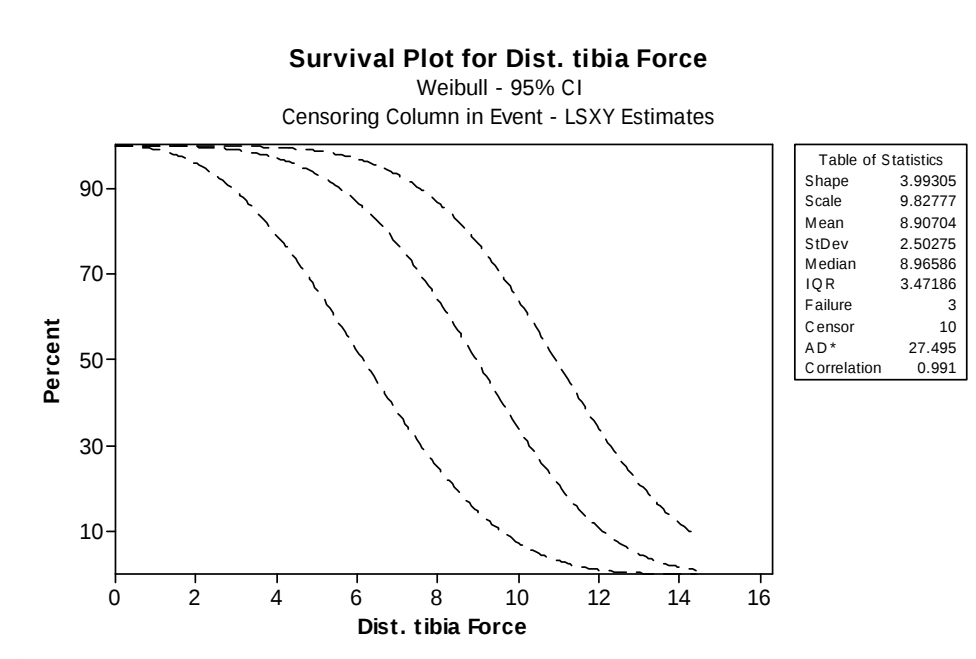


Fig. 4. Survivability Plot for distal tibia forces.

TABLE V
Survivability Analysis for Distal Tibia Forces

25.0%		50.0%		75.0%	
Force (kN)	Std. Error	Force (kN)	Std. Error	Force (kN)	Std. Error
11.41	1.38	8.91	0.99	6.41	0.83

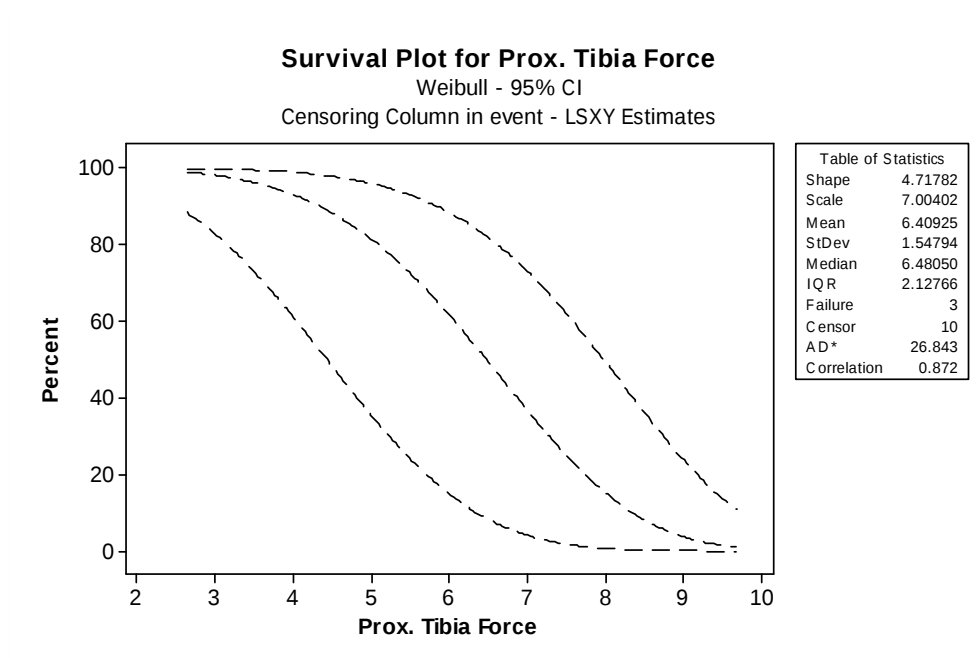


Fig. 5. Survivability Plot for proximal tibia forces.

TABLE VI
Survivability Analysis for Proximal Tibia Forces

25.0%		50.0%		75.0%	
Force (kN)	Std. Error	Force (kN)	Std. Error	Force (kN)	Std. Error
7.96	4.165	6.41	0.59	4.86	0.31

Table VII summarizes the input energies and the resulting injuries for each test performed. The mean energy of the impact was calculated to be $831.6 \pm 301.63\text{J}$, while the mean time for this energy to be transmitted from the bottom of the foot to the proximal tibia was $4.5 \pm 1.6\text{ms}$. The energy values were obtained by calculating the potential energy of the hammer for each test with consideration to test-specific weight and drop height. Tests performed with the fixed boundary condition using the 38.5 kg hammer produced more severe injuries including comminuted calcaneus fractures. Injuries ranged from calcaneus fractures to talus, cuboid and distal tibia fractures. The injuries were coded with AIS; all were of severity level 2.

TABLE VII
Injury Matrix

Specimen #	Test #	Input Energy (J)	Force Distal Tibia (kN)	Force Proximal Tibia (kN)	AIS	Injury Description
549L	1.1	755.37	2.70	4.6	-	None
	1.2	755.37	1.92	2.83	857361.2	Calcaneus: fx line into one joint surface
549R	1.3	755.37	3.11	5.35	-	None
	1.4	755.37	2.92	6.75	857371.2	Calcaneus: fx line into greater than/equal to 2 joint surfaces
537L	1.5	755.37	15.5	10.42	857300.2 857200.2	Calcaneus: fx NFS Talus: fx NFS
537R	1.6	755.37	6.20	18.31	857361.2	Calcaneus: fx line into one joint surface
544L	1.7	1200.75	9.01	14.51	857600.2	Cuboid: fx NFS
					857361.2	Calcaneus: fx line into one joint surface
					854361.2	Distal Tibia: fx, partial articular
547R	1.8	1200.75	4.11	5.82	857361.2	Calcaneus: fx line into one joint surface
548L	1.9	1200.75	8.50	6.83	-	None
543L	1.10	1200.75	7.34	6.06	854361.2	Distal Tibia: fx, partial articular
548R	1.11	1200.75	7.30	6.72	857200.2 857300.2	Talus: fx NFS Calcaneus: fx NFS
540L	1.12	1200.75	2.03	4.02	857371.2	Calcaneus: fx line into greater than/equal to 2 joint surfaces
543R	1.13	1200.75	7.20	6.16	854361.2	Distal Tibia: fx, partial articular
					857351.2	Calcaneus: fx, extra articular
530L	1.14	773.27	4.53	3.04	857361.2	Calcaneus: fx line into one joint surface
					857200.2	Talus: fx NFS
540R	1.15	773.27	5.80	3.92	857300.2	Calcaneus: fx NFS
556R	1.16	503.25	2.37	1.92	857351.2	Calcaneus: fx, extra articular
556L	1.17	335.50	4.79	3.7	857351.2	Calcaneus: fx, extra articular
530R	1.18	469.70	4.85	7.55	-	None
564L	1.19	419.78	5.40	4.53	-	None
564R	1.20	419.78	6.52	4.59	-	None

Fig. 6 presents the results from a representative group of tests (Tests 1.15, 1.16, 1.17) that consist of similar boundary conditions and hammer mass, but different drop heights. For the distal tibia force (Fig. 6a), a clear pattern of force increase is seen as the drop hammer height (therefore, energy) is increased. The force-time histories for the proximal tibia (Fig. 6b) show a similar pattern of increased force with increased drop height, but with a significant phase shift becoming evident. Fig. 6c shows the strain-time histories for the distal tibia and a slight dependence on energy. Fig. 6d shows that the proximal tibia strain has less dependence on this increase in energy.

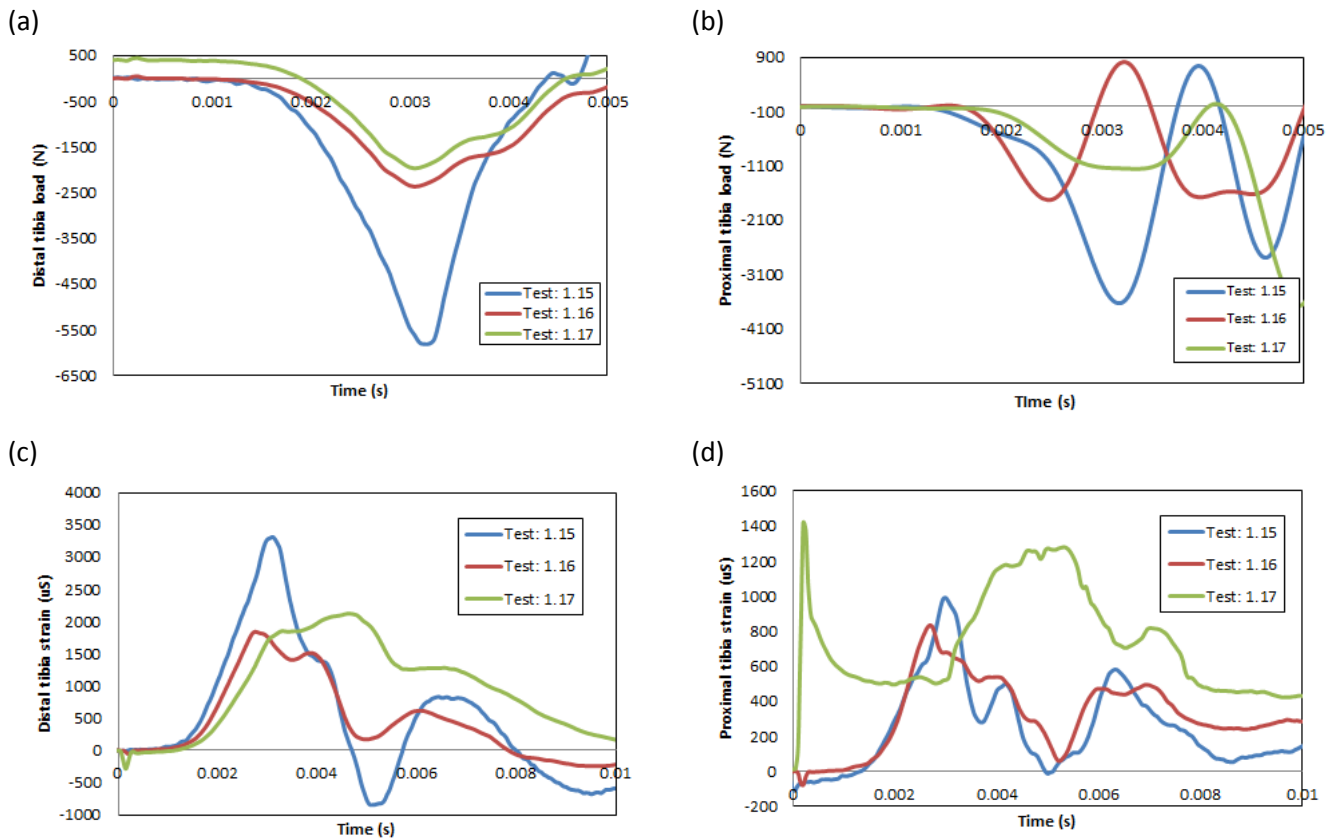


Fig. 6. (a) The distal tibia force-time histories for three different drop heights, same hammer mass. Note the increase in the distal response stiffness with the increase in energy; (b) Proximal tibia force-time history. Note that the response has less dependence on input energy; (c) Distal tibia strain-time histories. Note strain rate dependence on input energy; (d) Proximal tibia strain-time histories. Note insensitivity of strain rate to energy input (proximal strain gage on test 1.17 failed).

IV. DISCUSSION

The purpose of this test series was to investigate the effect of higher accelerations, stroke limitation, and a “free-to-translate” knee on the response and resulting injuries to the lower extremities. Previous tests involving automotive intrusion injuries have detailed the response and resulting injuries from similar boundary conditions, typically involving high accelerations, longer intrusion strokes [7] and an entrapped knee [11]. For UBB events, the boundary conditions are typically higher accelerations than those seen in automotive collisions, limited floor-pan intrusion (< 3 cm) and an unconstrained knee [6]. The resulting differences begins to resemble an event typical of energy deposition rather than an inertial event, focusing injuries at the point of application rather than the weakest part of the component. In this case, the resulting contact forces typically damage the calcaneus and distal tibia, where the automotive rate impacts focus more on the talus and distal-to-mid tibia [4]. Through the non-intrusive use of the bone-cell technique rather than the implantable tibia load cell historically used, this test series is able to detail the distribution of both forces and strains along the length of the lower extremity, without the risk of artifactual fractures commonly encountered previously.

The survivability forces produced are comparable to those from other studies. In one automotive-rate study, 26 PMHS lower extremities were tested in axial loading and a calculated 50% risk of injury was determined to be 6.8kN as recorded by an implanted tibia load cell [12]. Another study calculated a 50% risk of injury at 5.9kN recorded from an implanted tibia load cell [13]. These forces are comparable to the 50% survivable probability of 6.41 kN for the proximal tibia found in this study, showing some load rate insensitivity at this location. Additionally, the 50% survival probability for the distal tibia found in this study (8.91 kN) is reasonable considering the higher load rates and the “energy deposition” rather than inertial event.

The coefficients for the three-mass model of the lower extremity are shown in Table VIII. A two-mass model of the lower extremity developed for automotive intrusion found the spring coefficient to be $k=141.77$ kN/m and the dampening coefficient to be $c=932.4$ N s/m [11]. The increase in stiffness found in the current study demonstrates the rate dependence seen at these higher rates of loading. Interestingly, dampening at the distal tibia/ankle could not be identified at these higher loading rates. The utility of such lumped-mass models is clear, as they can aid in vehicle design, the evaluation of injury mitigation concepts for UBB protection and the development of suitable lower extremities for more biofidelic ATDs.

TABLE VIII
Spring and Compliance Constants

Model Parameters	Magnitude	SE
k_1 (kN/m)	2028.38	± 525.02
k_2 (kN/m)	9058.35	± 2043.382
c_1 (N.s/m)	0	± 0
c_2 (N.s/m)	800	± 169

While both the results of the survivability analysis and the lumped-mass model compare well to automotive studies, there are other factors that need to be considered for underbody blast events. An improved way of measuring the energy delivered to the lower extremity needs to be developed. The force measured in the tibia represents only a fraction of the force seen in the entire leg, as the fibula also carries some of the axial load [14]. Additionally, personal protective equipment, specifically military issued boots, needs to be accounted for in testing, as they will displace significant energy and lower the loading rates seen in the lower extremities.

V. CONCLUSIONS

Lower extremity injuries from toe-pan intrusion have been extensively researched for the automotive safety industry. This research must be considered as the foundation for higher levels of acceleration such as those seen in UBB events. Along with the higher acceleration/shorter duration intrusions seen in underbody blasts are smaller intrusion displacements. At very high acceleration/short durations, intrusion into the lower extremities can be much smaller than those seen in automotive impacts, yet the injuries can be as, or more, severe. While the loadings to the lower extremities in automotive impacts can be described as inertially driven, the injuries from the higher accelerations propagate distally and are beginning to resemble an energy deposition, with injuries resulting from focused contact. This study attempts to characterize the changes in lower extremity response to these higher acceleration events, and suggests that the mechanism of injury is transitioning from the inertially driven event typically seen in automotive collisions to something more focal, at least partially caused by the structural stiffening seen at these elevated load rates. Acknowledging the high-rate response of the lower extremities is critical when designing new military transport vehicles and protective footwear.

VI. ACKNOWLEDGEMENT

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