The Effect of Occupant Posture on the Risk of Fracture in the Human Tibia Under Dynamic Impact Loading

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ABSTRACT

Vehicle floor intrusion is a common cause of injury to the lower leg in automotive collisions. For the most part, experimental investigations into the fracture tolerance of the leg in these collisions have assumed that a vehicle occupant is seated in an idealized neutral posture, with force directed along the long axis of the leg (producing pure compression). This neglects the non-standard postures an occupant may actually assume during a crash, which result in bending in addition to compression. A new injury criterion that accounts for these non-standard postures and the associated combined loading is needed in order to better assess risk and evaluate protective measures in collision scenarios.

A custom-built apparatus was used to deliver impulses representative of automotive collisions to eight isolated human tibiae (four pairs, female, aged 48-73). Impacts were delivered at an average velocity of 6 m/s and over an average duration of 23 ms, in order to represent realistic conditions of a frontal collision. The mass of the impacting projectile (and correspondingly impact energy) was increased until fracture was achieved. One specimen from each pair was held at a posture of 15° from pure axial loading and the contralateral at 30°. Forces, moments, and impulse were collected from the tests and analyzed after fracture to determine the effect of posture on injury risk. It was found that specimens held at the smaller angle tended to withstand higher resultant forces applied to the bone before fracture, and that the Revised Tibia Index (RTI) in its current formulation was not reliably able to predict fracture.

This work to quantify the effects of leg posture on injury risk will lead to the development of more comprehensive design criteria for vehicle occupant protection measures.

INTRODUCTION

Frontal automotive collisions frequently result in fractures of the lower leg and foot for vehicle occupants. While not often life-threatening, these injuries may be debilitating, and can result in long-term pain and impairment for the person (Read, 2004). The mechanisms of injury in this scenario have been widely studied through experimental testing, whereby impacts were delivered to the lower limb via the plantar surface of the foot (Crandall., 1998; Funk, 2002; Gallenberger, 2013; McKay, 2009; Yoganandan, 1996). In these studies, it was generally assumed that force during these injurious events would be directed along the long axis of the leg. However, an occupant may take on a range of postures due to anthropometrics and the design of a vehicle interior, and this may result in the load being applied off-axis (Figure 1). In a review of tibial shaft injuries sustained in automotive collisions, bending was found to be the most common failure mechanism regardless of fracture site (Ivarsson, 2008). While some of this bending may be due to tibia curvature, computational models of floorpan intrusion have shown that lower limb posture has an effect on injury risk, with off-axis loading lowering the threshold for fracture (Dong, 2013; Hardin, 2004).

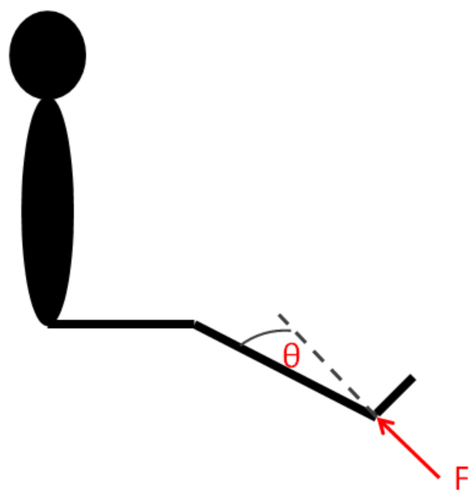

Figure 1: Illustration of off-axis loading resulting from a non-standard posture. Force, F, is applied along the impact path. The angle, θ, that the lower leg makes with the impact path is defined as the leg angle.

One of the most notable injury criteria for the lower leg, the Tibia Index (TI) (Mertz, 1993), expresses injury risk as a linear combination of compressive and bending loads measured by an Anthropomorphic Test Device (ATD):

$$TI = \frac{F}{F_c} + \frac{M}{M_c} \qquad \text{(Eq. 1.)}$$

In this equation, F and M are the applied force and moment as measured by an ATD load cell, and F_c and M_c are critical force and moment values, initially suggested by Mertz to be 35.9 kN and 225 Nm for a 50th percentile adult male and 22.9 kN and 115 Nm for a small female. In the initial formulation, a TI value of 1 or greater corresponded to fracture risk. Due to concerns about the derivation of these critical loads, a Revised Tibia Index (RTI) was suggested with critical values of 12 kN and 240 Nm for a 50th percentile male (Kuppa, 2001). Both of these formulations were based on results from mechanical tests of the tibia's strength at midshaft, with compression and bending being tested independently of one other. This is despite the fact that most tibial shaft fractures occur in the distal third region (Ivarsson, 2008). Additionally, bending strength was assessed in a 3-point test, in which load is applied perpendicular to the bone's axis at a point along its diaphysis (Kuppa, 2001). However, in a frontal collision load is delivered through the foot, generating different bending. The goal of this work was therefore to experimentally subject tibias to off-axis loading to study the effects of simultaneous bending and compressive loads on its injury limit.

METHODS

Specimen Preparation

Eight fresh-frozen cadaveric isolated tibiae (four pairs) were used for this study. The specimens were from female donors, aged 48-73. Specimens were fixed in four inch PVC pipe sections using dental cement at their proximal ends to a depth of approximately three inches. The pipe sections provided a method to securely affix them in the test apparatus. Specimens were aligned using laser levels in the pots based on anatomical axes. Each specimen was centered within the pipe section, and the anterior crest at midshaft was used to define vertical in the frontal plane, and the midpoint of the malleolus used for the sagittal plane (Figure 2).

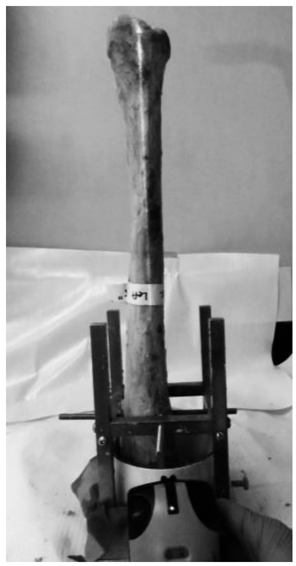

Figure 2: Specimen alignment based on anatomical axes during potting process. The bright vertical line on the specimen is from the laser level used for alignment.

After potting, the bones were returned to the freezer. Prior to testing, specimens were removed from the freezer again and thawed for a minimum of four hours.

Test Apparatus

Experiments were conducted using a custom-built apparatus (Figure 3) that allows the user to deliver impulses to a specimen via a pneumatically-propelled projectile. The specimen is suspended within the test chamber by a set of chains connected to an overhead rail and bearing system. Ballast masses are fixed to the top of the specimen support subsystem in order to simulate the inertial effects of the rest of the leg. The total mass of the specimen, the support system, and the ballast mass for this study was 12.9 kg, corresponding to the mass of a 50^{th} percentile adult leg and foot (Huston, 2013). The projectile does not strike the specimen directly, instead delivering impulse through an intermediate footplate. A uniaxial accelerometer (MMA1200, Freescale Semiconductor, Austin, TX, USA) is fixed to the footplate.

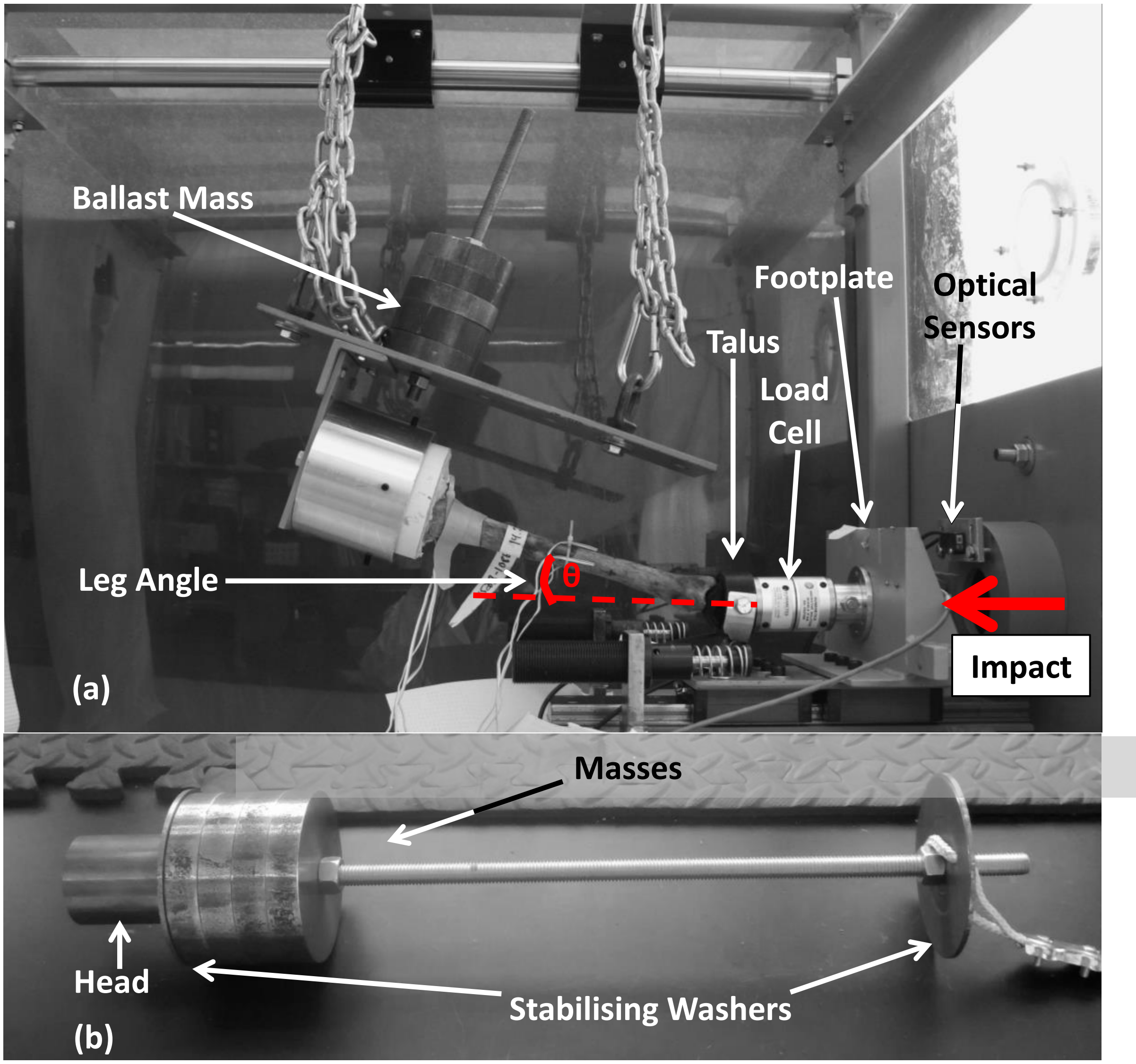

Figure 3: (a) Test specimen suspended in impacting apparatus, with ballast mass, artificial talus, load cell, optical sensors, and impact path shown. The leg angle of the specimen is also identified, and was varied between contralateral specimens in this study. (b) Impacting projectile with head, stabilizing washers, and removable masses indicated.

Load is delivered to the specimen's distal articular surface via a rapid prototyped ABS artificial talus based on geometry extracted from a Computed Tomography scan of an average male adult ankle. This talus is mounted to the end of a six-axis load cell (IF-625, Humanetics Innovative Solutions, Plymouth, MI, USA) attached to the intermediate footplate, and the specimen is securely pressed up to the talus and aligned in all directions prior to each test. The load cell allows for measurement of the reaction loads at the tibial plafond during each impact.

The projectile has a solid steel head that contacts the intermediate footplate and is supported by two stabilizing washers that keep it supported in its barrel. The user may vary the projectile's mass by adding and removing weights between these washers. The pressure of the compressed air supplied to the apparatus (controlled using a regulator) is used to vary the projectile's impact velocity, and impact duration is modulated via a layer of foam covering the intermediate footplate. The apparatus is controlled using a custom-written LabVIEW program. The program collects signals from the load cell at 50 kHz and calculates the velocity of the projectile using two optical sensors (PZ-V31P, Keyence Corporation, Osaka, Japan). All tests are captured on high speed video at 1250 frames per second.

Testing

One specimen from each pair was held with a leg angle of 15° and the contralateral was held at a larger angle of 30°. In order to represent the conditions of a frontal collision, impacts were targeted to be at a velocity of 6 m/s (Crandall et al., 1998) and duration of 25 ms (McKay & Bir, 2009). In order to keep these properties consistent among trials, the projectile mass was increased between tests on each specimen. This resulted in increased energy and force delivered to the intermediate footplate (and correspondingly, the specimen). Impacts were increased until a distal tibia fracture was produced, defined as the splitting of the bone into two distinct regions. In order to minimize the effects of fatigue loading, attempts were made to limit the number of impacts to each specimen. The goal was to strike each specimen at sub-failure loading once, before producing a fracture on the second strike.

A best subsets regression was performed on data collected from impact tests (with the response being 1 for tests resulting in fracture and 0 for no fracture) to determine the best predictors of injury. The factors considered were leg angle, projectile mass, projectile impact velocity, peak footplate acceleration, peak load cell vertical force (F_y), horizontal force (F_z), resultant force (F_{res}), bending moment (M_x), projectile impact energy, and total applied impulse (calculated as the resultant of the integral of the y and z force-time curves). Paired t-tests were used to determine whether or not there was a significant difference in the means of various parameters between the two postures tested, with $\alpha = 0.1$.

RESULTS

Intra-articular fractures were produced in all specimens (Figure 4). An average of 2.6 (1.1) strikes were required to achieve failure in all specimens, and the average projectile velocity across all trials was 6.3 (0.9) m/s with an average impact duration of 22.8 (9.5) ms. The loads measured in trials leading to impact are summarized in Tables 1 and 2.

Figure 4: Examples of fractures produced in the study. Various fracture patterns were noted, and with different levels of comminution.

Table 1: Highest achieved and fracture impact test conditions for specimens tested at 15 degrees

Specimen #	Fracture Vertical Force, F_y (N)	Fracture Horizontal Force, F_z (N)	Fracture Resultant Force, $F_{res,f}$ (N)	Highest Achieved Resultant Force, $F_{res,h}$ (N)	Fracture Bending Moment, M_x (Nm)	Fracture Impulse, I_F (Ns)	Highest Achieved Impulse, I_h (Ns)
1547L	4422	5549	7073	7073	213	21.9	41.3
1582L	2393	6829	7319	7776	123	21.3	51.2
1640L	3287	6050	6885	6885	131	16.9	16.9
1653R	3048	8667	8853	8853	133	21.9	43.4
Average (±S.D.)	3287 (846)	6774 (1368)	7532 (898)	7647 (891)	150 (42)	20.5 (2.4)	38.2 (14.8)

Table 2: Peak and fracture impact test conditions for specimens tested at 30 degrees

Specimen #	Fracture Vertical Force, F_y (N)	Fracture Horizontal Force, F_z (N)	Fracture Resultant Force, $F_{res,f}$ (N)	Peak Resultant Force, $F_{res,h}$ (N)	Fracture Bending Moment, M_x (Nm)	Fracture Impulse, I_F (Ns)	Peak Impulse, I_P (Ns)
1547R	4146	4606	5647	5647	194	20.7	37.9
1582R	4576	6113	7555	8683	208	27.1	40.8
1640R	2634	2711	3780	3780	134	21.8	33.6
1653L	3766	4404	5484	5484	86	26.8	41.8
Average (±S.D.)	3781 (833)	4459 (1392)	5617 (1544)	5899 (2039)	156 (57)	24.1 (3.3)	38.5 (3.7)

In all cases, impact duration and impulse were greater in the trial preceding fracture than in the fracture case itself (Figure 5).

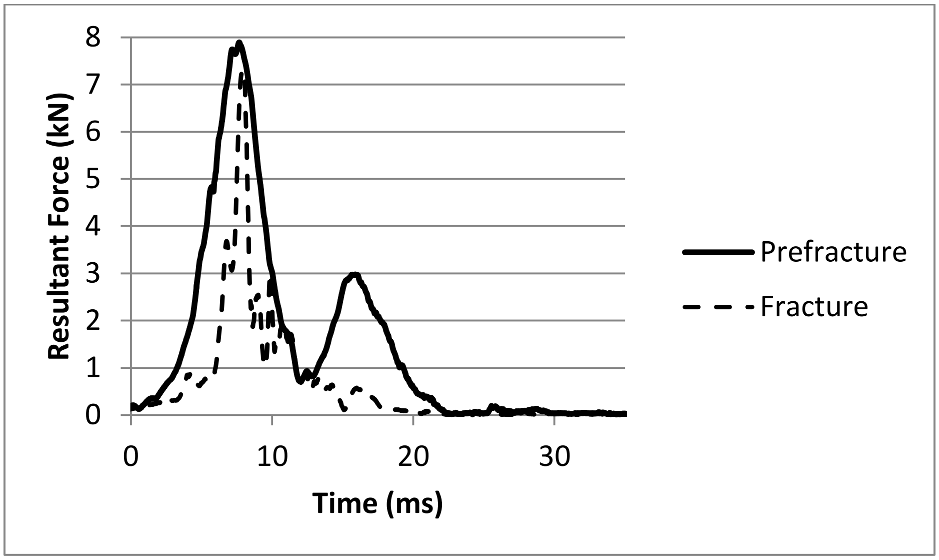

Figure 5: Force-time curve for pre-fracture test of specimen 1582L. Note the shorter impact duration and reduced impulse in the fracture case.

Based on the value of R^2-adjusted from the best subsets regression analysis, the best model for predicting injury includes leg angle, projectile mass, peak horizontal force, peak resultant force, peak bending moment, and applied impulse. For this model, R^2 was 90.4 and R^2-adjusted was 85.2.

In six out of the eight specimens, the highest peak resultant force corresponded to the fracture case. However, for specimens 1582R and 1582L, peak force was achieved in the trial preceding the fracture impact. In order to rule out the effect of potential accumulated damage, the highest achieved resultant force was used for the paired t-tests in addition to the peak force at fracture.

These t-tests indicated that the specimen's angle had a significant effect on fracture horizontal force (p=0.087) (Figure 6).

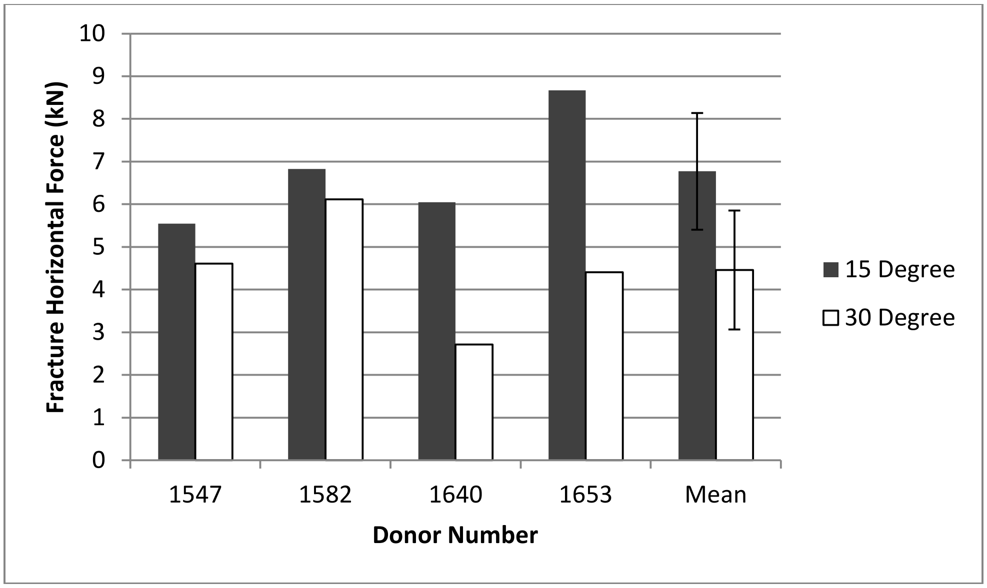

Figure 6: Comparison of horizontal force (F_z) measured during fracture for each specimen, separated by leg angle.

The difference between the two postures tested was not significant for impulse at fracture (p=0.108), impulse at peak resultant force (p=0.959), fracture vertical force (p=0.493), horizontal force at peak resultant force (p=0.135), resultant force at fracture (p=0.106), peak resultant force (p=0.174), bending moment at fracture (p=0.851), and bending moment at peak resultant force (p=0.873). However, impulse at fracture and resultant force at fracture were very near the defined limit for significance, and will be examined further to determine if a relationship can be derived between these values and posture.

The loading applied to each specimen was converted to RTI values for both the fracture tests and the non-fracture test immediately preceding it. RTI was calculated using resultant force (from F_x, F_y, and F_z) and resultant bending moment (M_x and M_y) measured by the load cell. The difference between the means of RTI values at fracture were not significant between the 15° and 30° cases (p=0.595).

Based on the guidelines for RTI set out by Kuppa et al (2001), an RTI value of 1.16 corresponds to a 50% chance of serious injury, and the calculated RTI for each specimen was compared to this value. RTI values for the pre-fracture case ranged from 0.75 to 1.48 (mean = 1.11), and for the fracture case ranged from 0.90 to 1.50 (mean = 1.20, Table 3).

Table 3: Revised Tibia Index values at peak resultant force case for each specimen

Specimen #	Angle (degrees)	Revised Tibia Index Value – Pre-fracture	Revised Tibia Index Value - Fracture
1547L	15	0.75	1.48
1547R	30	0.89	1.29
1582L	15	1.45	1.13
1582R	30	1.48	1.50
1640L	15	-	1.13
1640R	30	0.87	0.90
1653L	30	0.90	0.90
1653R	15	1.41	1.32
Average (±S.D.)		**1.11 (0.32)**	**1.20 (0.23)**

DISCUSSION

This study examined the tibia's response to complex but realistic loading in an experimental set-up that has not been seen previously in the literature. Impacts that were intentionally representative of a frontal collision were applied to isolated tibias, allowing for easy identification of the mechanism of failure in the bone. The use of paired specimens also allowed for close examination of the effects of posture on injury tolerance while controlling for variation among individuals. Older female specimens were also utilized in order to provide a conservative estimate of injury tolerance for the general population. While the effects of lower leg posture on injury tolerance have been assessed numerically (Dong, 2013; Hardin, 2004), the isolated tibia has not been experimentally tested under off-axis loading.

It was noted that during the repeated testing, impulse and impact duration decreased between the pre-fracture and fracture cases. This is likely due to energy absorption during the fracture event. It is also possible that during the repeated testing, a crack initiated that was not visible, thus weakening the specimen. However, efforts were made to minimize the number of impacts, and the specimen that fractured on the first impact did so at a load that fell within the range of the other specimens.

Based on the four pairs tested, it was found that specimens with an initial leg angle of 15° have a higher tolerance for force in the z direction than those held at 30°, likely due to the dominance of compression in the overall loading. However, the bending moment measured by the external load cell was very similarly distributed between the two postures, and therefore the measured forces from the load cell should be examined in further detail to determine their contribution to bending.

Calculating Tibia Index values for these tests was challenging due to the off-axis loading as well as the fact that the standard critical values for RTI are based on tests conducted on male specimens, but it appears that the model cannot be applied to this case in its current formulation. Further investigation of critical loads for off-axis loading to refine the Tibia Index is required.

While no other work has ever been done in this exact configuration, the peak resultant force values of 5.6-7.5 kN fall within the range of expected values based on the results of similar previous studies. The compressive tests that provide the critical force value for the RTI give an average compressive failure load of 12 kN for the tibia. As well, one study that involved impacts to isolated tibias achieved an average peak force of 12.6 kN before failure (Quenneville, 2011). This is higher than the force values measured in the current study, but may be attributed to the fact that the previous study made use of male donors, and investigated shorter duration and axial impacts.

The small sample size should be kept in mind when considering the significance of these results. Ongoing testing will expand the sample size, and hopefully lead to further clarity on the optimal injury risk function for combined loading.

Because of the non-standard postures used in these tests as well as the irregular individual geometry of each articular surface, it was sometimes challenging to achieve full contact between the specimen and the artificial talus, which likely produced stress contours that would not match those of the native joint. Therefore, it is anticipated that peak forces and moments will provide more reliable injury thresholds than impulse, despite its identification as a predictor of fracture. In addition, peak forces and moments are easy to extract from experimental tests and can generally be compared simply between cadaveric specimens and surrogates such as synthetic bones and ATDs.

When developing any posture-specific load limits for ATDs, it will be important to recreate the experimental fracture conditions in order to scale the resulting loads. Fracture loads from cadaveric bone should not be used directly because they will not account for the relatively higher stiffness of ATD legforms.

CONCLUSIONS

Specimens with a smaller leg angle had a higher tolerance for force than those held at a larger angle, likely due to the increased contribution of bending in the overall loading. Force in other directions and impulse absorbed by the bone before fracture did not differ significantly between the two postures tested, although they were approaching the level of significance and may be of interest as the work is expanded with greater statistical power. The authors intend to test more specimens according to this protocol, and these relationships will be reevaluated with the larger sample size.

Ultimately these data will be useful in the development of a novel injury risk criterion that accounts for combined loading due to posture. This criterion may be used in the development of design metrics for safety features in automobiles.

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