Evaluation of the Distribution of Forces on the Foot Under Axial Impact Loading to Assess Variations Among Altered Ankle Postures Between Two ATD Models

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Abstract Anthropomorphic Test Devices (ATDs) are used to assess injury risk during impact scenarios, such as motor vehicle collisions. Risk of foot/ankle injuries in these scenarios is traditionally evaluated by analyzing the peak axial force and the Tibia Index, collected from tibia load cells. This neglects assessment of foot injury risk and does not account for potential changes in injury assessment at altered postures. Two commonly used ATDs, the Hybrid III and Military Lower Extremity, were exposed to impacts at five different ankle postures, at varying degrees of flexion. An array of piezoresistive sensors located on the insole of a boot was employed to assess the load distribution variations between postures and ATD models. Both posture and ATD model affected the load distribution on the foot, highlighting the need to establish injury limits for non-neutral postures. The increase in forefoot loading during plantarflexion was not reflected in the standard industry metrics, suggesting that increased fracture risk to the forefoot would not be detected. The differences in load distribution between the models also emphasizes the importance of selecting the correct surrogate. These data will be useful for establishing ATD measures that correspond to future PMHS injury tests at altered postures.

Keywords Anthropomorphic Test Device (ATD), Foot/ankle injury, Impact, Injury assessment, Ankle posture

I. INTRODUCTION

With advancements in seatbelt and airbag technology providing better protection for the head, neck and torso, lower extremities are now the most frequent site for “non-minor” (AIS 2+) injuries to occur in a frontal collision, of which trauma to the foot and ankle make up 30% [1]. Axial loading of the lower extremity is responsible for injuries with the most significant long-term impairment and causes an estimated 40% of all injuries to this region in frontal collisions [2]. Anthropomorphic Test Devices (ATDs) are commonly used to assess injury risk to the foot-ankle complex, where the lower leg is typically instrumented solely with an upper and lower tibia load cell. Although ankle and toe load cells exist for the Hybrid III foot, they are not often employed in industry testing. Injury risk to this region is typically grouped with the rest of the lower leg and evaluated based on tibia peak axial force (Fz) or the Tibia Index (TI) measured by the tibia load cells, and sometimes foot acceleration [3]. The calculated TI attempts to combine the axial load and bending moment in the lower leg into a single value related to injury potential. A TI score less than one indicates a passing test, whereas a TI score greater than one indicates a failing test, and therefore a poor safety rating.

There are many limitations to the current methods of developing and applying injury risk curves to lower extremities. Injury risk curves, which relate the probability of injury to metrics obtained from the aforementioned instrumentation, are developed using post mortem human subjects (PMHS). Research efforts have historically focused on the ankle positioned in a 90° tibia-to-foot (neutral) posture [4,5], though vehicular occupants may assume a range of ankle postures while driving [6]. A limited number of PMHS studies have investigated the effect of ankle posture on injury mechanism, severity and type, and results have suggested that dorsiflexed ankle postures are more resistant to injury [7] and that plantarflexed ankle postures are more likely to result in distal tibial fractures [8,9]. Ankle posture and its effect on injury risk have not been reflected in current injury risk curves. Furthermore, PMHS testing is generally conducted without the use of footwear, while industry testing using ATDs usually incorporates clothing and footwear. The employment of footwear in impact scenarios has been shown to reduce tibia axial forces by 50-65%, depending on the test conditions [10,11]. Footwear offers a unique

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opportunity to measure impact loads to the foot/ankle complex and load distribution along the plantar surface of the foot, metrics which are typically neglected, using an identical measurement tool for both PMHS and ATDs.

The Hybrid III 50th Male (Humanetics Innovative Solutions, Plymouth, MI, USA) is the most widely-used ATD in automotive impact testing, approximating the height and weight of the 50th percentile adult male. However, this model has been shown to be relatively insensitive to ankle posture changes [8][12]. The shaft of the Hybrid III leg is angled, positioned slightly posteriorly above the ankle. This causes the Hybrid III legshaft to be angled when positioned in a neutral position. Concerns over the Hybrid III's response to under body blasts, particularly it's overly stiff structure, led to the development of the MIL-Lx (Military Lower Extremity, Humanetics Innovative Solutions, Plymouth, MI, USA), which is commonly used for analyzing anti-vehicular land mine protective systems. The MIL-Lx leg incorporates a straight knee clevis, tibia shaft, and ankle. Studies have demonstrated the effect of ATD selection on injury risk assessment, indicating that the Hybrid III and MIL-Lx legs generate considerably different force-time and peak force measurements [8][10][13,14]. This is partly due to the Hybrid III foot lacking cushioning elements, whereas the MIL-Lx has compression-absorbing elements in the tibial shaft and heel [14]. Differences between the surrogate ankle design elements may alter the post-impact kinematics (i.e. by rotating the foot about the tibia differently) which may alter the load transmission through the lower leg. While the MIL-Lx is accepted as having a more biofidelic tibia, neither ATD model currently provides any indication of regional loading the foot may encounter during impacts. The differences in load distribution between the two ATD models could alter injury risk assessment in frontal collisions.

To address some of these limitations, our lab designed a custom boot instrumented at the insole with eight piezoresistive sensors (Fig.1) and calibrated it to convert resistance to force over a range of impact energies typical of frontal automotive collisions [15]. The instrumented boot also provides a representation of the regional loading characteristics of the plantar surface of the foot over the entire impact event. It also provides a consistent footwear model, transferrable between PMHS and ATD impact studies. However, this boot has not yet been used to comparatively assess the two ATD models, or to assess the effects of ankle posture.

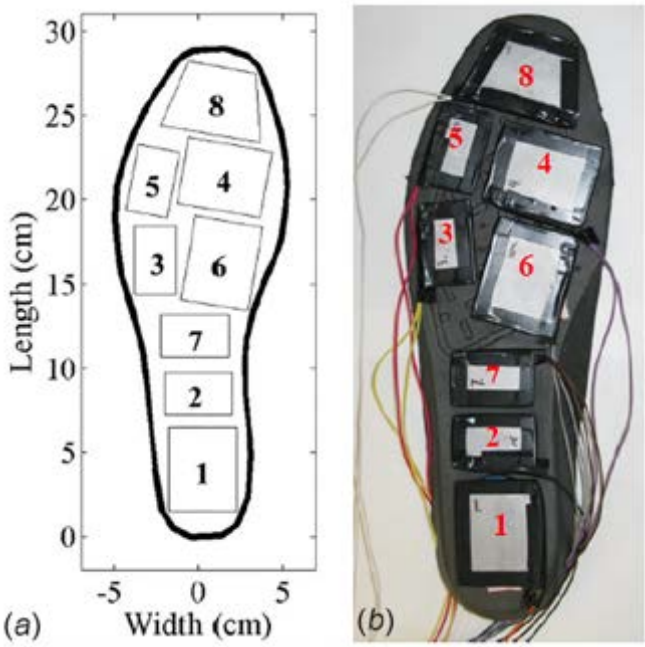

Fig. 1. Piezoresistive sensors covered the main loading regions of the insole, based on (a) the schematic, and (b) their corresponding locations on the insole of the instrumented boot. Figure adapted from [15]. Sensors 4, 5, and 8 were grouped together to from the "forefoot" loading region, Sensors 3, 6, and 7 were grouped to form the "midfoot" loading region, and Sensors 1 and 2 formed the "hindfoot" region.

The purpose of this study was to assess the force distribution across the plantar surface of the foot using two common ATD surrogates: the Hybrid III and the MIL-Lx 50th percentile male legforms. Varying degrees of ankle flexion were investigated to evaluate how ankle posture affects the axial load and corresponding injury risk, both assessed using the boot and standard tibia metrics.

II. METHODS

Experimental Testing

All testing was completed using a pneumatic impacting apparatus (Fig.2), which has previously been used in many impact studies [15-17]. Impacts were conducted at a kinetic energy of approximately 280 J, in which a steel projectile was propelled down an acceleration tube by compressed air towards the testing chamber. Impulse was transmitted to the plantar surface of the foot *via* an ankle positioner, which was mounted on low-friction linear bearings. A block of rubber foam was placed on the surface of the ankle positioning device being struck in order to control pulse duration. This foam was replaced every four impacts to mitigate any potential accumulated damage.

The Hybrid III and MIL-Lx lower leg models were each in turn fitted with the instrumented boot and fixed to a proximal bracket at the knee which was suspended on a linear rail and bearing system (Figure 2). This allowed free translation and rotation of the surrogate after impact. Ankle posture was controlled by the ankle positioner, and care was taken to ensure each ATD was axially aligned in the direction of impact, and only the foot was rotated about the ankle joint. The flesh analog of both ATD models were removed to facilitate proper alignment of the ATDs, and neutral posture of the leg was defined as the long axis (defined as line connecting knee clevis to center of rotation of the ankle joint) at a 90° angle to the plantar surface of the boot. Ballast weight was secured to the suspension jig to bring the total mass of each legform to 12.9 kg, to compensate for the total mass of a 50th percentile male leg, simulating linear inertial properties in of the remainder of the leg [18].

All impacts were delivered at a velocity of approximately 5.8 m/s and a duration of approximately 20 ms, intended to be in the range of realistic impact conditions resulting from a frontal collision [7][19]. Each ATD was axially impacted in five ankle postures: 20°-dorsiflexion, 10°-dorsiflexion, neutral, 10°-plantarflexion, and 20°-plantarflexion, with five repeated impacts conducted at each posture. Two unrecorded impacts were performed at the start of each ATD testing sequence to ensure the boot was sufficiently worn in.

The testing procedure was controlled, and data were collected, using a custom-written LabVIEW (National Instruments, Austin, TX, USA) program. A uniaxial accelerometer (MMA1200, Freescale Semiconductor, Austin, TX, USA) was mounted on the ankle positioner to collect footplate acceleration. Two optical sensors (PZ-V31P, Keyence Corporation, Osaka, Japan) were mounted over the projectile exit from the acceleration tube to collect impact velocity. All data, including the two 5-axis load cells (Fx, Fy, Fz, Mx and My) in the upper and lower tibia and the eight boot insole sensors, were recorded at 50 kHz.

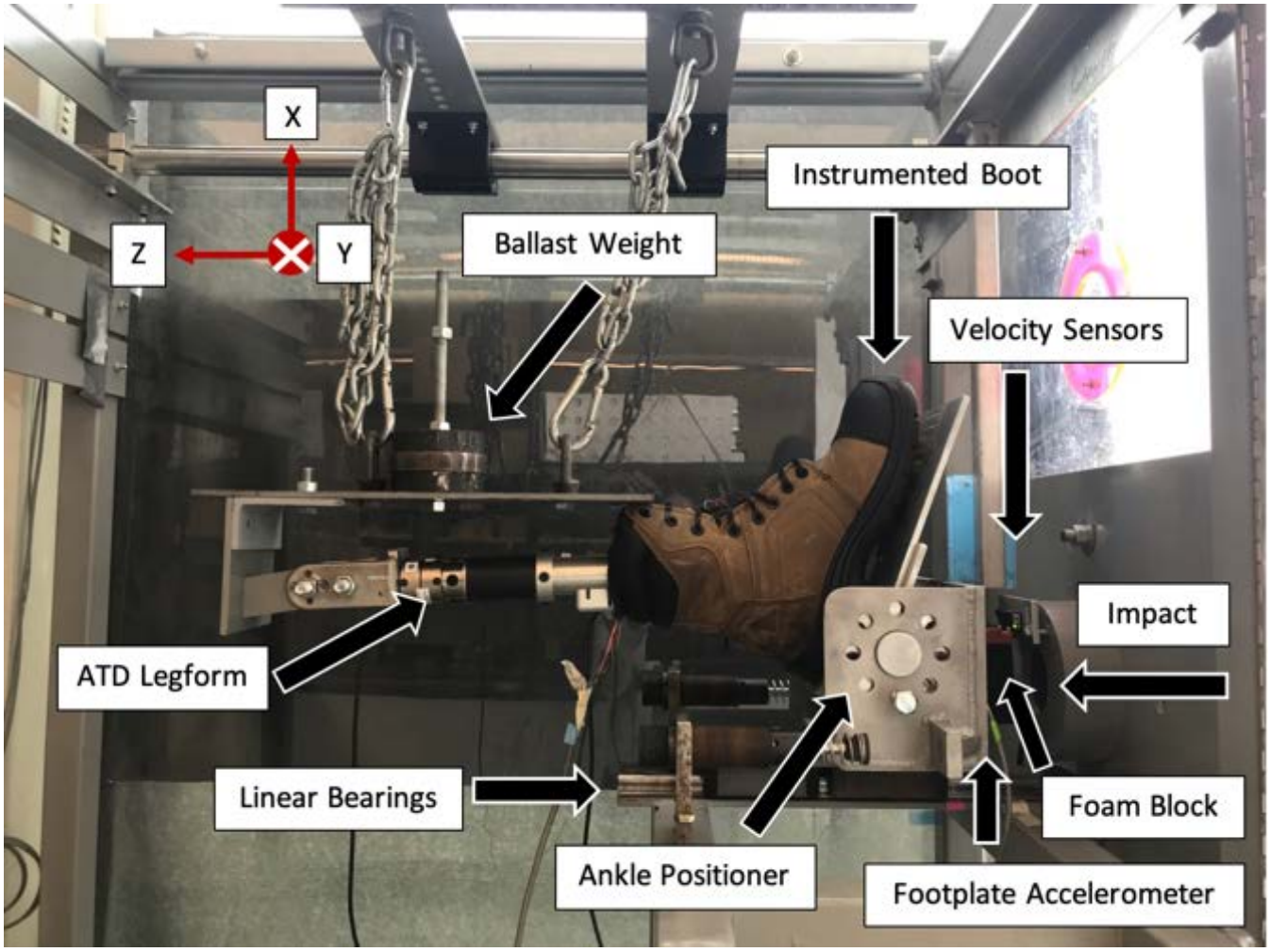

Fig. 2. The MIL-Lx suspended in the pneumatic impacting apparatus fitted with the instrumented boot and positioned in 20°-plantarflexion. A coordinate system is included to indicate axis used in tibia load cell data collection.

Analysis

For data presentation and interpretation, Sensors 4, 5, and 8 were grouped together to from the "forefoot" loading region, Sensors 3, 6, and 7 were grouped to form the "midfoot" loading region, and Sensors 1 and 2 formed the "hindfoot" region. Tibia load cell data were dual pass filtered using a second-order Butterworth low-pass filter with a cutoff frequency of 1,250 Hz [20]. Metrics were assessed at the distal tibia load cell for the Hybrid III leg and the proximal tibia load cell for the MIL-Lx, in accordance with industry standards [18]. Impact duration was considered to have begun 1 ms before the distal Fz data increased to 10% of the peak Fz and concluded 1 ms after the distal Fz fell below 10% of the peak Fz. The five repeated trials were averaged, and the standard deviation was determined. The primary outcomes from the ATD were the peak axial forces in the upper and lower load cells, as well as the Tibia Indices (TI).

The Tibia Index is a widely accepted injury criterion for the lower leg [21]. The Adjusted TI (TI_{Adj}) was developed to account for the geometry of the Hybrid III leg [22, 23]. Tibia Indices were calculated according to the following equations:

$$TI = \frac{F}{F_c} + \frac{\sqrt{{M_x}^2 + {M_y}^2}}{M_c}, \tag{1}$$

$$TI_{Adj} = \frac{F}{F_c} + \frac{\sqrt{{M_x}^2 + (M_{y,\,distal} - 0.006398\,F_z)^2}}{M_c}, \tag{2}$$

where *TI* corresponds to injury risk (where values greater than 1 indicate failure), *F* is the applied axial force and

M_x and M_y are measured tibia moments. F_c and M_c are the critical values previously determined using cadaveric testing, 35.9 kN and 225 Nm, respectively [21]. Equation (1) was used to calculate the proximal TI for MIL-Lx impacts, while Equation (2) was used for Hybrid III distal tibia load cell analysis [18].

An unpaired *t*-test was conducted to compare net boot forces between ATDs at each posture. A one-way ANOVA with post-hoc Tukey test was also conducted to compare the net boot forces between postures, with a significance threshold of α=0.05.

III. RESULTS

A total of 50 axial impacts were conducted for the purposes of this study, with an average impact velocity of 5.86 ± 0.11 m/s. The footplate acceleration was 125 ± 10 g for the Hybrid III and 127 ± 8 g for the MIL-Lx, showing no significant difference between ATD models (p=0.43). The impact duration differed for the two ATD models (p=0.03) and was an average of 16.4 ± 1.2 ms for the Hybrid III and 22.1 ± 2.5 ms for the MIL-Lx. Five force-time curves were produced for each ATD, one for each posture, with the mean trace graphed and shaded regions bounding it representing the standard deviation range (Fig 3). In each posture, impacts were aligned by their peak force. All five repeated tests had very similar time durations, indicating the methodology of producing these curves was repeatable. The Hybrid III force traces were generally noisier and had a larger standard deviation range than the MIL-Lx model. The MIL-Lx also reached a lower peak force over a generally longer impact duration. The axial force measurements (Fz) and Tibia Indices were plotted for each ATD in each posture (Figs. 4 and 5).

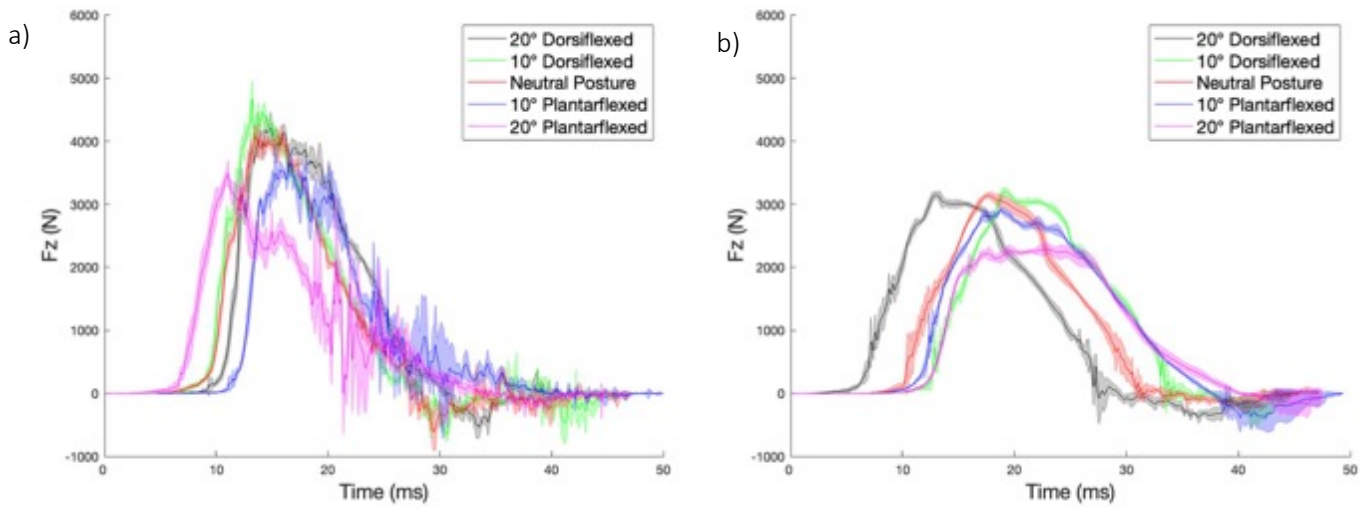

Fig. 3. Average axial force-time traces for (a) Hybrid III distal load cell, and (b) MIL-Lx proximal load cell, presented for the five postures and with standard deviation regions indicated with shading.

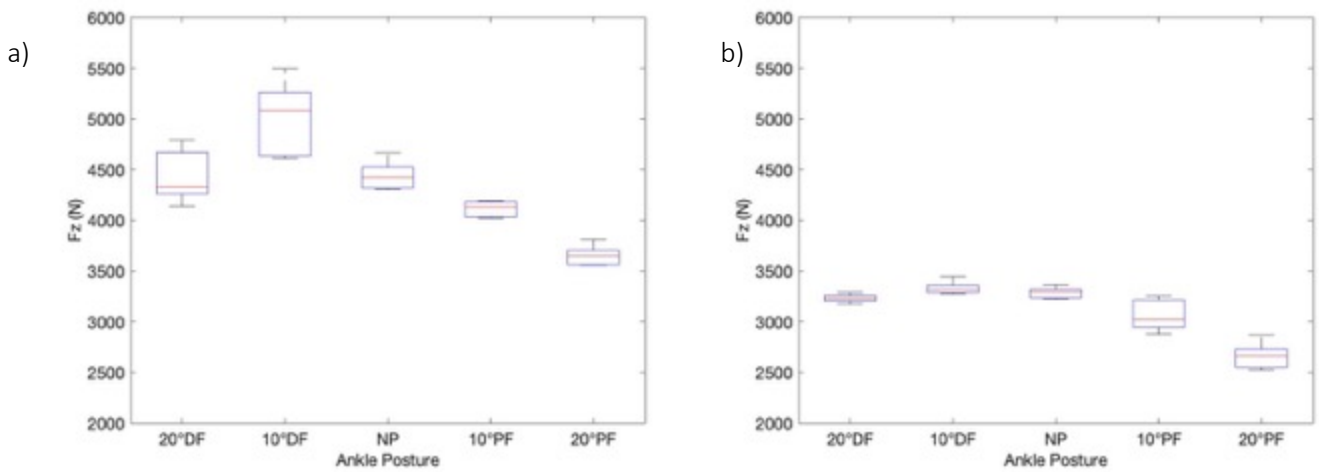

Fig. 4. Peak Axial Forces of the (a) Hybrid III model and the (b) MIL-Lx model, measured at varying degrees of flexion. Data from the distal tibia load cell were analyzed for the Hybrid III trials, and the proximal tibia load cell was used in the MIL-Lx trials.

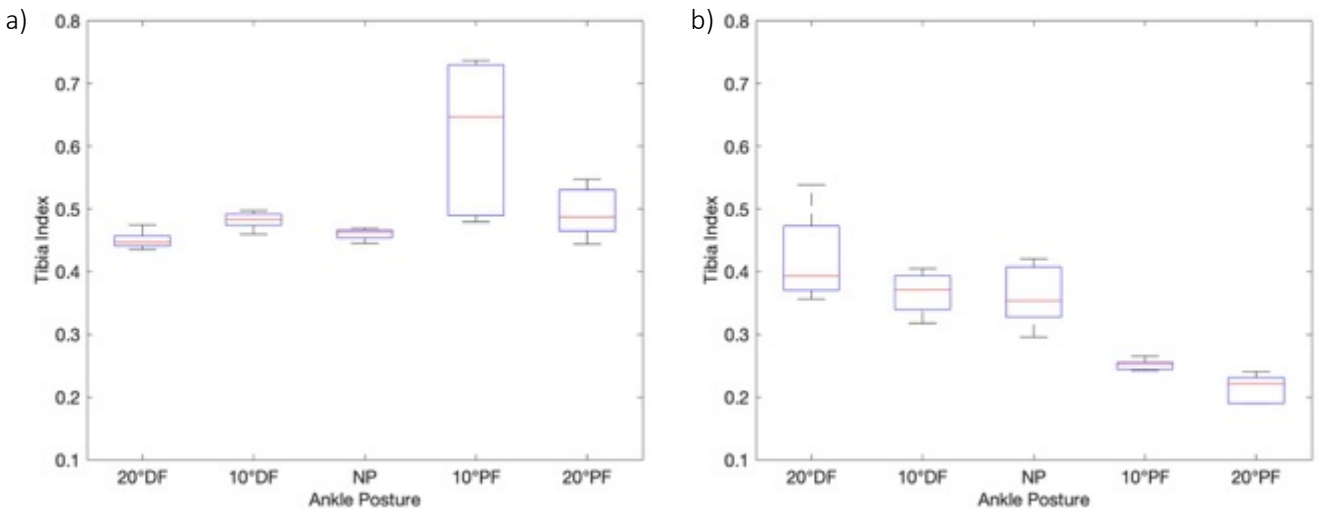

Fig. 5. Tibia Indices of the (a) Hybrid III legform and (b) MIL-Lx legform, measured at varying degrees of flexion. Data from the distal tibia load cell were analyzed for the Hybrid III TI_{Adj}, and the proximal tibia load cell for the MIL-Lx TI.

A sum of each sensor's peak axial force collected from the instrumented boot were compared to the tibial load data collected in the ATDs across all postures (Fig. 6). The boot read forces greater than the tibia load cell across all postures, for both ATDs. In the Hybrid III model, the boot collected forces 24-26% higher than forces collected at the distal tibia load cell, and in the MIL-Lx, the boot collected forces 46-58% higher than forces collected at the proximal tibia load cell (depending on posture). The tibia load cells collected more of the boot forces when the ankle was in the plantarflexed posture. The boot collected the highest loads in the 10°-dorsiflexed posture for the Hybrid III. The dorsiflexed and neutral postures were significantly different from the 20°-plantarflexed postures in the Hybrid III, and the plantarflexed postures were significantly different from the 20°-dorsiflexed posture. In the MIL-Lx, the boot collected highest loads when the ankle was positioned in neutral. The dorsiflexed postures were significantly different from the neutral posture, and the 20°-plantarflexed postures, and the plantarflexed postures were significantly different from the neutral posture and the 20°-dorsiflexed posture. Results from the *t*-test indicated that when ATDs were in the same posture, all net boot forces were statistically different from one another, with the exception of the 20°-dorsiflexed posture.

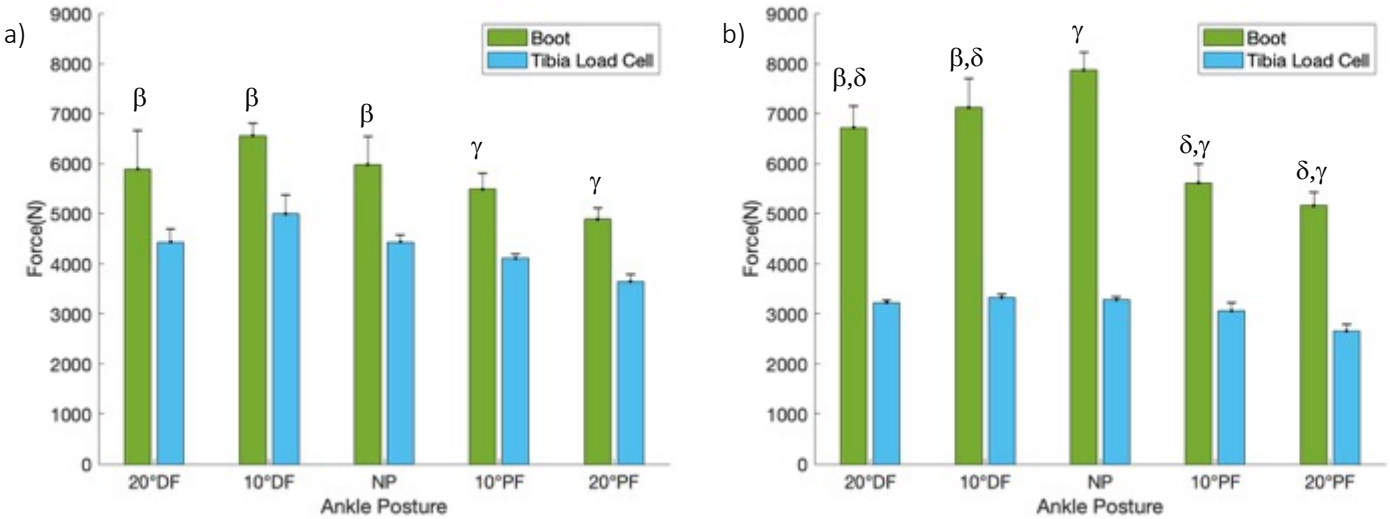

Fig. 6. Peak ATD forces compared to net peak boot sensor forces from (a) Hybrid III legform and (b) MIL-Lx legform, measured at varying degrees of flexion. Data from the distal tibia load cell were analyzed for the Hybrid III TI_{Adj}, and the proximal tibia load cell for the MIL-Lx TI. β indicates a significant difference from the 20°-plantarflexed posture, γ indicates a difference from the 20°-dorsiflexed posture, and δ indicates a significant difference from the neutral posture.

When examining the distribution of load across the plantar surface of the foot in varied postures, two primary trends were observed in both ATD legforms (Fig. 7). Firstly, in all impacts the hindfoot carried the majority of the load, which was followed by midfoot and then forefoot (except for the MIL-Lx at 20°-plantarflexion, where this was reversed). Secondly, as the ankle moved from dorsiflexion through neutral and into plantarflexion, a portion of the hindfoot load was transferred to the other regions, primarily the forefoot. This effect was most pronounced in the MIL-Lx.

At neutral (NP), the hindfoot recorded 71% of the load in the Hybrid III model and 80% of the total load in the MIL-Lx model. The forefoot recorded 14% of the loads in the Hybrid III and 5% of the loads in the MIL-Lx. When the ankle was positioned in the plantarflexed (PF) postures, the sensors in the hindfoot region decreased substantially in both ATD models. The Hybrid III model recorded hindfoot forces of 49% and 51% in 10°- and 20°-plantarflexion, respectively, while the MIL-Lx hindfoot loading decreased to 67% in the 10°-plantarflexion and 52% in the 20°-plantarflexed posture. This load was mostly transferred to the forefoot, which increased from 18% to 25% in the Hybrid III model (but demonstrated no trend with increasing angle) and for the MIL-Lx increased from 12% to 32% in the 10°-plantarflexed postures in the 20°-plantarflexed postures, respectively.

In dorsiflexion (DF), the hindfoot sensor loads also decreased from 65% at 10°-dorsiflexion to 64% at 20°-dorsiflexion in the Hybrid III and reduced from 75% in the 10°-plantarflexed posture and to 70% in the 20°-plantarflexed postures in the MIL-Lx. Forefoot loading collected 9% at the 10°-dorsiflexed posture and 12% at the 20°-dorsiflexed posture in the Hybrid III model. In contrast, in the MIL-Lx the forefoot loading in this posture increased from neutral to 6% in the 10°-dorsiflexed posture, and 11% in the 20°-dorsiflexed posture. In general, for both surrogates, the dorsiflexed postures exhibited lower loads in the forefoot as compared to plantarflexed postures.

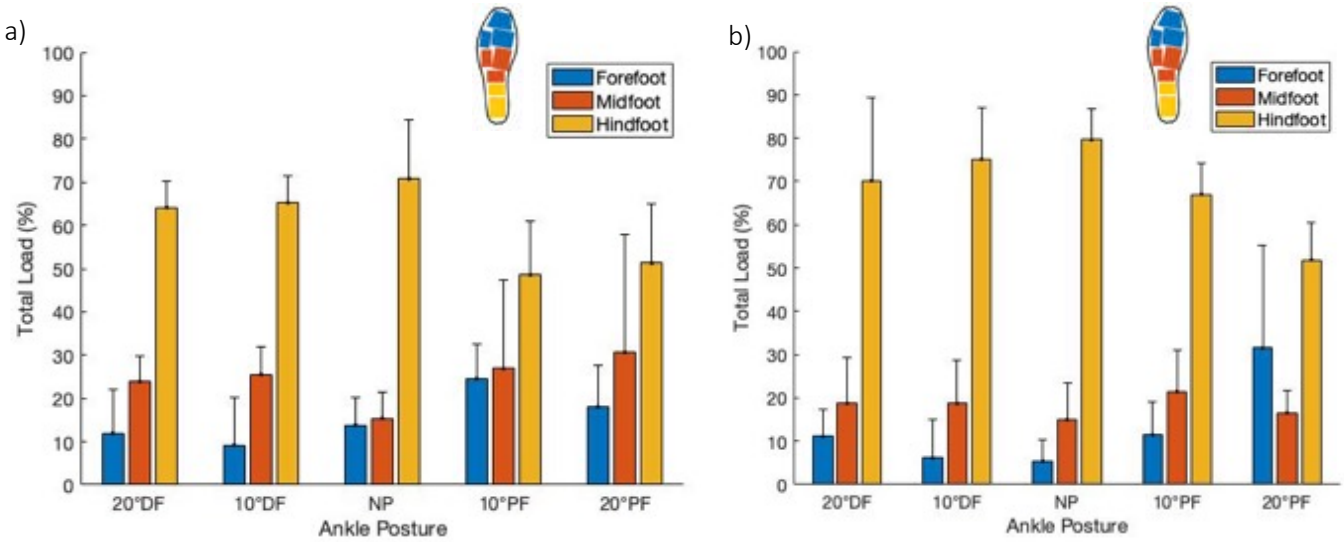

Fig. 7. The average distribution of forces that each sensor carried during (a) the Hybrid III impacts and (b) the MIL-Lx impacts, measured at five postures at varying degrees of flexion: 20° dorsiflexion (20°DF), 10° plantarflexion (10° PF), neutral (NP), 10° plantarflexion (10° PF), and 20° plantarflexion (20° PF). Error bars represent the standard deviation across all trials.

IV. Discussion

It has been demonstrated by multiple researchers that injury risk to the lower leg may be altered when the foot is positioned in a non-neutral posture [7-9][24,25]. However limited studies have examined the effect of initial posture on the response of ATD model lower legs [8][12], which must have a biofidelic response in order to be considered an appropriate model for predicting injury to this region. Of these previous studies, none has conducted as extensive an evaluation of ankle posture and foot/ankle injury as the present study. The effect of ankle posture on both the traditional injury metrics of axial load and tibia index, as well as the new location of measurement of the plantar surface of the foot, were collected under conditions representative of frontal motor vehicle collisions. Impacts were delivered in the range of the results of a frontal crash simulation by Crandall *et al.*, that reported a floor velocity of 5 m/s [26]. Similarly, impact duration was in the range of 15 to 45 ms,

representative of a frontal automotive impact [19]. Impacts were conducted using both the Hybrid III and the MIL-Lx ATD models, and both surrogates proved to be consistent models with the MIL-Lx showing slightly better repeatability results in comparison with the Hybrid III, which is consistent with a previous study [14].

The peak axial force (as measured by the industry standard load cell in each surrogate) showed a decreasing trend as the ankle was moved from dorsiflexion through neutral and into plantarflexion. The Hybrid III force values were consistently higher than the MIL-Lx, which is unsurprising given the acknowledged stiffer nature of this surrogate. As the MIL-Lx is more compliant, it has a correspondingly lower injury criterion, at 2.6 kN, versus the 5.4 kN threshold typically used for the Hybrid III [20]. Interestingly, the Hybrid III would have 'passed' all of the tests conducted herein, as all peak axial forces were below the 5.4 kN threshold. This is in contrast to the MIL-Lx which would have 'failed' 4 of 5 tests, in which the peak axial force was greater than the 2.6 kN injury criterion, which suggests that the two surrogates and their respective injury criteria are not entirely equivalent. This is in contrast with a previous study [13] that found that the Hybrid III lower leg exceeded its criterion at lower impact conditions than the MIL-Lx. However, the impact durations in the present study were greater than that previous one (which was simulating anti-vehicular mine blasts, with impact durations in the 10-15 ms range). The same previous study noted that the relative performance shifted when an energy attenuating mat was included, which in effect extended the impact durations, and may be more in line with the impact conditions applied in the present study.

When comparing the summation of peak sensor forces to tibia load cells, the sensors consistently collected forces that were higher than the tibial load cells. This is unsurprising due to the force dissipation that occurs between the plantar surface of the foot and the tibia, and these results align well with a previous study conducted by Acharya, who found ankle and toe load cell forces that were consistently 120-130% higher than tibia forces [15]. The statistical analyses that were conducted on the data demonstrated that postural affects between postures are significant.

The increased forefoot loading in plantarflexed postures that were observed in this study are similar to those observed in a previous study conducted by Grigoriadis *et al.* [8], with a notable difference of an increase in hindfoot loading in the plantarflexed posture as compared to the neutral posture in the Hybrid III ATD. This study displayed similar force-time traces to the current study, where the MIL-Lx measured force values consistently lower and with a longer impact duration than that of the Hybrid III. In neutral postures, the surrogate tibia shaft was most aligned with the direction of impact, and therefore most stiff. This resulted in higher peak forces and shorter impact durations, a trend also observed in a previous study [12]. The TI_{Adj} and peak Fz values reported in [12] also showed very similar trends in varying degrees of flexion in comparison to the current study.

In all impact scenarios, for each posture and for each model, the hindfoot sensors carried most of the load. This is unsurprising as the hindfoot lies directly under the ankle joint. Furthermore, ATDs do not mimic muscle loading in the ankle, so forces acting on the forefoot cause the ankle to rotate with minimal resistance. Measuring the force distribution in PMHS under similar impact conditions and recording injuries in specimens with the distribution of forces in the ATDs would be a useful step in developing regional injury criteria for the foot.

While the axial force measures did not show much variation in either model with altered postures, the same cannot be said about the Tibia Index measures. In the Hybrid III, moving from dorsiflexion to plantarflexion increased Tibia Index. As this move decreased axial force, this means that the bending moment was substantially increased with altered posture, which may be a function of the irregular geometry of the Hybrid III lower leg. Because the 'neutral' lower leg posture for a Hybrid III has the tibia already at an angle, this likely led to the development of artificial bending moments under axial loading, which has been previously observed [10]. Conversely, the MIL-Lx decreased in Tibia Index when moving from dorsiflexion to plantarflexion. This trend is in contrast to PMHS studies, which suggest the dorsiflexed posture is more resistant to injury in comparison to other postures [7], indicating that TI may not be a reliable assessment of injury risk as it relates to posture. These data also suggest that the TI and TI_{adj} are sensitive to different factors, making them difficult to comparatively assess, showing only similar values in the 20° dorsiflexion posture. This also highlights the interdependence of axial force and Tibia Index as assessment metrics.

This study found that both posture and ATD model affected the load distribution across the insole of the boot. As the ankle was moved from dorsiflexion to plantarflexion, loads in the hindfoot tended to decrease and get redistributed to the mid- and forefoot regions. The hindfoot loading trend paralleled the tibia axial force trend, suggesting these two are highly correlated, but while the Hybrid III had higher axial forces, the MIL-Lx had higher

hindfoot percentages. This may have been the result of ankle rotations during impact, or load being converted into bending, and highlights the complex kinematics associated with these impacts. In plantarflexion, the forefoot proved to be particularly vulnerable, carrying a large percentage of the total load, and thus injuries to this region may require separate injury criteria and evaluation of risk for more complete assessments of safety. Interestingly, none of the metrics evaluated herein showed the dorsiflexion posture as being more resistant to injury, as was noted in a previous cadaveric study [7], which may be the function of the simpler joint representation in an ATD when compared to PMHS (with ligamentous structurers and numerous irregular bones that may alter the load path).

Both the Hybrid III and the MIL-Lx showed greatest loading in the hindfoot when positioned in a neutral posture, a trend that was not observed in the Fz and TI data. This suggests that there may be important information to be gained by examining plantar surface force distribution. This lack of correlation with existing metrics may be a reason why the mid- and forefoot regions have not been previously studied in impact testing, although they would be particularly vulnerable in scenarios of impingement. However, the boot sensors had very large coefficients of variation (CoV). This was particularly problematic for the forefoot and midfoot regions, due to the smaller magnitudes of values at these locations, however for the hindfoot region, CoV values ranged between 0.07 and 0.23. These values exceed the acceptable level of variability in a test surrogate for injury risk assessment [27] and highlight that more work needs to be done to improve the repeatability of the insole as a tool. The CoV values for axial force in the ATDs were all acceptable (<8%), as were most of the Tibia Index CoV (but not all, with values exceeding 10% for one Hybrid III posture, and three MIL-Lx postures). The large standard deviations of the regional boot sensor load (Fig. 7) are a compound effect of test variability and ATD variability (similar to the standard deviation bars in Figs. 2 and 3), as well as boot sensor and ankle kinematic variabilities. Due to these factors, the boot forces have larger error associated the presented results.

Analysis of the data suggests that ankle posture is a good indicator for peak Fz, hindfoot loading, and forefoot loading, as these data showed correlating results. Interestingly, there was not a correlation of TI as it relates to ankle posture, the basis upon which many current injury limits are set. This further emphasizes the need for greater load characterization to this region of the body. The significant differences between impact duration in the models, 16.3 ms and 22.1 ms for the Hybrid III and MIL-Lx models, respectively, are not surprising due to the increased compliance of MIL-Lx model. The footplate accelerations were not significantly different between models (125 ± 10 g for the Hybrid III and 127 ± 8 g for the MIL-Lx), which is also unsurprising as this metric is a direct result of the applied kinetic energy.

There were a few limitations to the current study. The five trials conducted in each posture were done subsequently and were not randomized, due to time constraints of altering the ankle positioner between tests. It is possible that there was some accumulated damage or relaxation to the surrogate (or instrumented boot) with the repeated impacts. However, the number of impacts were similar to the number of impacts applied during the calibration process, and therefore any insole compression and relaxation should have been conducted prior to data collection. There may have also been small differences in the lacing of the boot on the two ATDs, but this was minimized as best as possible by having the same researcher tighten the boot on each surrogate. There were some inconsistencies in trends with the altered postures (e.g. the Hybrid III forefoot load percent increasing for 10° of plantarflexion, then decreasing at 20°. Testing a greater number of ankle postures may make trends in the data more evident. Ballast weight was affixed at a single location on the supporting bracket within the impacting apparatus, and as such did not realistically represent the distribution of mass that would occur in the natural lower limb. This may not have provided natural rotational inertia and could have altered post-impact kinematics. However, given the axial impact scenario, and the relatively modest percent of ballast weight as compared to total weight of the lower leg, overall kinematics were likely similar, and minimal rotation was observed upon examination of high-speed video. Furthermore, this was consistent among all tests. Also, while the data collected at the mid- and forefoot regions are of interest herein, there are no established injury criteria for these regions, making it difficult to evaluate the potential importance of this loading. Finally, tests were conducted only at a single impact speed, and always with the impact aligned with the lower leg axis. Further investigation on the implications of impact velocity, duration and acceleration may affect the force distribution on the foot. Any out-of-posture of the lower limb (not just ankle) would require separate and further investigation.

V. CONCLUSIONS

This study emphasized the importance of selecting the correct surrogate as well as considering initial ankle

posture when developing injury criteria for the lower extremity. This study also outlined the importance of developing regional injury criteria for the foot and moving beyond the gross measures indicating global mechanics. Increased loading to regions of the foot in different postures would go undetected without the use of instrumentation on the plantar surface of the foot. Finally, it is evident from these findings that ankle posture plays a large role in the force distribution on the plantar surface of the foot (as well as in tibia load cell measures, as previously noted). Since injury criteria are typically developed when the ankle is in a neutral posture, care must be taken when aligning the ATD foot according to the tibia shaft during collision testing to provide an accurate representation of injury risk. Finally, posture-specific injury risk curves should be developed to account for the different varying load transmission pathways, and these data may guide future developments of more biofidelic foot/ankle models. To the authors knowledge, this is the first study to assess the load distribution at the plantar surface of the foot under impact loading. It is also the first application of the novel instrumented insole and will form the baseline for comparison with future post-mortem human subjects to develop regional injury metrics.

VI. References

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