Investigation of the Biofidelity of the MIL-Lx Foot

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Abstract Foot/ankle injuries from frontal automobile collisions are frequent and debilitating. While post-mortem human subject testing is advantageous as it provides realistic injury responses, it is expensive, highly variable and does not collect internal loading data. Anthropomorphic Test Devices, however, collect load data and are easily used in industry. A biofidelic foot is essential for correctly transmitting load to the tibia where lower leg injury risk is assessed. This study combined Post-Mortem Human Subject feet with an Anthropomorphic Test Device tibia to examine the response of feet while collecting industry-relevant metrics. The Military Lower Extremity, six post-mortem human subject lower legs, and six adapted specimens containing the Military Lower Extremity "tibia" shaft and Post-Mortem Human Subject feet were equipped with an instrumented boot and axially impacted at 5 m/s, representing a frontal automotive collision. No significant differences were found on the plantar surface among all specimens, suggesting this may be a feasible method of evaluating foot response. Significant differences were found in the proximal tibia load cells when comparing the adapted specimens with the Military Lower Extremity, suggesting its foot is overly stiff. These data can be used for redesigning the Anthropomorphic Test Device foot, thus improving lower limb safety assessments.

Keywords Anthropomorphic Test Device (ATD), foot/ankle injury, injury assessment, impact, post-mortem human subject testing

I. INTRODUCTION

In automobile crashes, up to 10% of all non-minor (Abbreviated Injury Scale, AIS, 2+) injuries occur to the foot/ankle complex [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][3]. Although there has been a decrease in the overall frequency of injuries related to car crashes, foot and ankle injuries continue to increase in both severity and incidence [4]. These injuries are impactful and can lead to long-term impairment [1][4]. Typical methods of injury assessment in the automotive industry include the use of Anthropomorphic Test Devices (ATDs), with injury risk to the foot and ankle usually grouped and evaluated using load cells in the tibia. Two primary lower leg ATD models exist, the Hybrid III 50th Male (HIII), and the Military Lower Extremity (MIL-Lx) (Humanetics Innovative Solutions, Plymouth, MI, USA). Another ATD, the Test Device for Human Occupant Restraint Lower Extremity (THOR-Lx) (Humanetics Innovative Solutions, Plymouth, MI, USA), has been designed for increased biofidelity, but has not yet achieved widespread adoption [5]. The response of the MIL-Lx has been investigated both at the lower leg level [6] and as an isolated tibia [7] and is generally accepted as having a more biofidelic response [6-7]. Although this ATD was designed for blast events, it has potential for being a useful tool in the automotive industry for high-force crashes [8]. As such, this ATD leg was the focus herein.

While the biofidelity of the MIL-Lx tibia has been evaluated and compared to other surrogates, few studies have investigated the foot/ankle response specifically [9]. While the isolated MIL-Lx tibia (no foot) has previously been shown to have very good agreement with non-fracture post-mortem human subject (PMHS) tibia data (R^2 = 0.83), an extensive comparative response including the foot/ankle has not been conducted [7-8]. Investigation of the foot/ankle response is important, as load is transmitted through this region to the tibia, where injury risk is assessed. If the foot/ankle does not transmit the load correctly (either over or under attenuating), then tibia safety assessments may be incorrect. As part of ongoing efforts to design biofidelic ATDs, it is important to

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examine the response of the MIL-Lx foot to either validate the current ATD model or provide data for an improved design.

While ATDs are valuable tools with load cells to collect forces and act in a repeatable manner, they do not undergo injuries. Although ATDs are designed in an effort to exhibit a biofidelic response to impacts, they cannot be biofidelic above the fracture level, given that they do not fracture. In contrast, PMHS testing is advantageous as it allows for the identification of fracture limits, locations and mechanisms. However, generally it is also very expensive, requires a lot of preparation, and generates fracture sites that are extremely variable, mostly due to differences among specimen populations. Additionally, testing on PMHS material does not collect internal load data. Researchers have attempted to measure fracture forces in PMHS testing by implanting load cells proximal to the tibia, i.e., [10], or in the tibia itself, i.e., [2]. This makes it extremely challenging to assess fracture risk and compare it to ATD measurements. Furthermore, the addition of load cells can alter stress concentrations, potentially varying the injury mechanism as a result of impact. Differences in stiffness responses among ATDs and PMHSs have been acknowledged, and transfer functions are often used to relate ATD measured values to PMHS injury risk [7][11][12]. However, these functions are dependent on the specific scenario of loading (e.g. automotive collision versus under body blast) and have a great amount of uncertainty (e.g. specific to one given population, as PMHS are highly variable).

Foot/ankle injuries are impactful, as they involve many articular surfaces to disrupt, often leading to post-traumatic osteoarthritis and have poor vascularisation for healing [13]. The foot/ankle region is often neglected when assessing injury risk, by evaluating all injury risk to the lower extremity at ATD load cells in the tibia. Lower leg injury risk curves evaluate risk to the entire lower extremity [14], and although the isolated PMHS tibia has been investigated for its fracture limit [7][15][16], the PMHS foot has not. It is necessary to evaluate foot and ankle fractures separately from tibia injuries, as they are very different anatomically structured regions and have different healing responses [1]. Each component of an ATD should be evaluated independently, as they are acting in series, and the response from the tibia is largely dependent on the foot response. If an ATD foot is overly stiff in comparison to PMHS, forces recorded in the tibia may be much higher than if the stiffness were comparable. As the tibia has already been evaluated as being biofidelic [6-7], the foot must also be examined. The exact force required to fracture a PMHS foot, collected while simultaneously reading forces collected in the tibia shaft, is valuable information that would allow for direct translation into industry how much force is required to generate a foot fracture, and the type of fracture endured. The need for a transfer function (by testing the ATD under parallel conditions to those that pose a specific level of risk to PMHS specimens) between ATDs and corresponding injury risk would thus be eliminated. This is advantageous as data collected from these tests could apply to previous IARVs to update them to account for increased injury risk. To the authors' knowledge, no previous studies exist that examine the axial impact response of the isolated PMHS foot while simultaneously collecting tibia load cell measures.

The purpose of this study was first to compare the axial response of intact PMHS lower legs to the MIL-Lx at impact velocities and durations similar to those experienced by the lower extremity in frontal automotive collisions, and secondly, to develop a method to mount PMHS feet onto the MIL-Lx tibia shaft. This was done to facilitate the investigation of the foot response while collecting the industry-relevant metrics of peak axial force (Fz) and Tibia Index (TI). The objective was to quantify the differences among lower legs to investigate the biofidelity of the MIL-Lx foot to either validate the current MIL-Lx foot or inform future iterations of ATD feet.

II. METHODS

All specimens were fitted with an instrumented boot while tested, which was equipped with eight piezoresistive sensors covering the plantar surface of the foot (Fig. 1). The piezoresistive sensors were sandwiched between plates of 16-gauge cold rolled steel blasted with glass beads to provide a uniform contact surface and were previously calibrated to convert sensor voltage to force [17]. The instrumented boot provides a representation of the regional loading characteristics of the plantar surface of the foot over the entire impact event. This would be valuable information in real-world settings, as it may account for foot entrapment which would not be reflected in standard injury assessment procedures at the tibia. It also provides a consistent footwear model, transferrable between PMHS and ATD impact tests. The boot has previously been used to comparatively assess ATD models and evaluate the effects of ankle posture [18]. In each of the tests, the boot was tightened and laced firmly over each foot, by the same investigator each time. It was tightened in between impact events and all sensors were zeroed after it was fitted to account for any pre-impact loading on the foot.

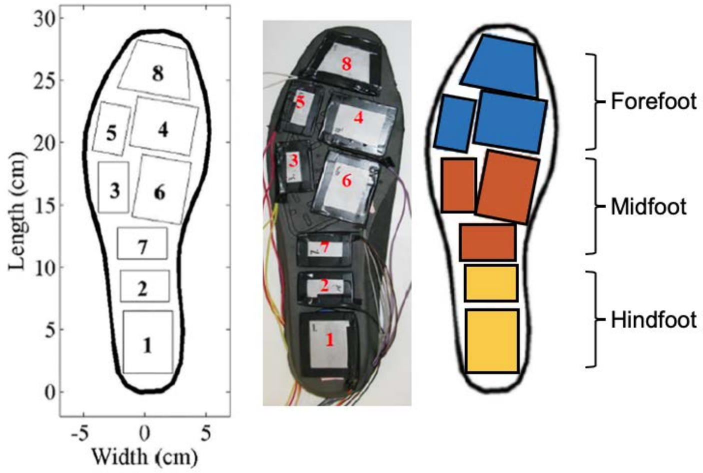

Fig. 1. Piezoresistive sensors on the instrumented boot covered the main loading regions of the insole, based on (a) the schematic, and (b) their corresponding locations on the insole of the instrumented boot, and (c) their regional division. Sensors 4, 5, and 8 were grouped together to form the *forefoot* loading region, highlighted in blue, Sensors 3, 6, and 7 were grouped to form the *midfoot* loading region, in orange, and Sensors 1 and 2 formed the *hindfoot* region, in yellow. Figure adapted from [17].

All impact testing was completed using a pneumatic impacting apparatus, which has previously been used in impact studies [15-18]. Steel impact masses were 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 (Fig. 2). A block of rubber foam (approximately 2" thick) was placed on the surface of the ankle positioning device being struck in order to control pulse duration. This foam was replaced after each specimen (every two impacts) to mitigate any potential accumulated damage. Each ankle was oriented in a neutral position when impacted, defined as when the plantar surface of the instrumented boot was at a 90° angle to the MIL-Lx tibia shaft or PMHS tibial ridge. Ballast weight was secured to the suspension jig to bring the total mass of each specimen to 12.9 kg, the total mass of a 50th percentile male leg [19], in an effort to simulate natural linear inertial properties.

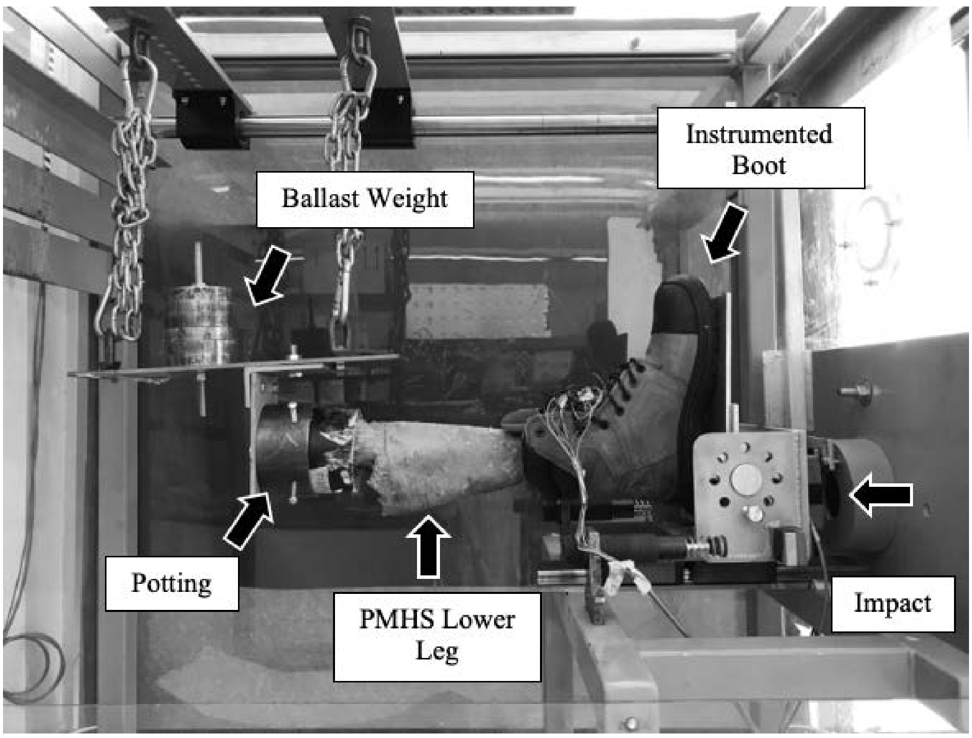

Fig. 2. A PMHS specimen inserted into the instrumented boot and into the impacting apparatus.

MIL-Lx Experimental Testing

The MIL-Lx was fitted with the instrumented boot, and one unrecorded impact at the start of the testing sequence, a *settling* impact to ensure the foot positioned in the boot, was conducted. The MIL-Lx was then axially impacted four times. It was supported within the pneumatic impacting apparatus at the knee clevis.

Intact PMHS Experimental Testing

Six intact lower leg specimens sectioned at the tibial plateau (three male / three female, aged 80 ± 12 years) were tested (Table 1). The specimens were x-rayed in the anterior-posterior and lateral views prior to impacting, and an orthopaedic surgeon declared there were no pre-existing injuries.

Specimens were dissected 2" distal to the tibial plateau and potted at the knee using dental cement in a section of 4"-diameter circular polyvinyl chloride (PVC) piping, to provide a consistent method to support the specimens while testing and ensure proper axial alignment. Consistent alignment was ensured through the use of a laser level projected along the tibial ridge, and the bone was embedded to the full depth of the PVC pipe (2"). All specimens were thawed for a minimum of 12 hours before testing. The leg was inserted into a sealable plastic bag, then into the instrumented boot and mounted in the impacting chamber in a neutral ankle posture. It was secured proximally by fixing the PVC pipe to the ballast plate.

TABLE I
CHARACTERISTICS OF LOWER LEGS TESTED

Test	Age	Sex	Foot Length (cm)
MIL-Lx	N/A	Male	26.1
Specimen 1	69	Female	22.2
Specimen 2	69	Male	27.3
Specimen 3	95	Male	27.3
Specimen 4	95	Male	26.7
Specimen 5	77	Female	22.2
Specimen 6	77	Female	21.6

Adapted Legform Experimental Testing

In order to characterise the response of the natural foot for defining stiffness requirements for the MIL-Lx, isolated PMHS feet were tested. Each intact PMHS was disarticulated at the tibiotalar joint and x-rayed in the anterior-posterior and lateral views after disarticulation. An orthopaedic surgeon again evaluated the x-rays and declared no injuries had occurred during dissection or intact testing.

The MIL-Lx was chosen as the ATD "tibia" shaft due to its superior biofidelic properties in comparison to other ATDs [7][20]. There were several challenges associated with mounting a PMHS foot to an ATD tibia. First, soft tissue support was necessary to facilitate proper alignment of the foot with respect to the artificial tibia shaft, in an effort to replicate natural joint motion during initial positioning. Secondly, natural load transmission between the talus and the ATD shaft was important so that no abnormal stress concentrations on the articular surface could create point loading that would artificially induce a fracture. Finally, the correct alignment and attachment of the MIL-Lx tibia shaft itself to the foot was challenging to replicate a natural line of action of the load path.

In an effort to facilitate proper alignment of the foot with respect to the tibia shaft, and to support the union of the two structures, all soft tissue distal to and surrounding the talus was preserved. The deltoid ligaments and lateral ligaments (posterior tibiofibular, posterior talofibular, and superior fibular) were sutured with a Krakow stitch, using a 2.0 FiberWire suture (Arthrex Inc., Naples, Florida, USA). A Krakow stitch was selected as the suturing mechanism to reduce the risk of shredding or tearing of the tissue. In this, as force is applied, the mechanism tightens around a bundle of fibres and prevents the sutures from pulling through the fibers, providing a secure attachment of the suture to the tendon or ligament [21]. FiberWire sutures were chosen specifically for this application due to their superior strength, allowing for a tight loop to be secured to the ligament group. The flexor digitorum longus medial tendon, extensor halluces longus medial tendon, tibialis posterior medial tendon and tibialis anterior medial tendon were also secured individually with a Krakow stitch, using Coated VICRYL® Plus Antibacterial (polyglactin 910) sutures (Ethicon Inc., Somerville, NJ, USA), which have strong tensile capabilities. These tendons and ligaments have been reported to all play an important role in ankle joint stability [22][23].

A secondary challenge with attaching a PMHS foot to the ATD shaft was attempting to keep load transmission natural. In an effort to do so, the distal tibia and fibula of each specimen were optically scanned and 3D-printed to replicate natural bone geometry. The distal end of each specimen's tibia and fibula were scanned using a handheld optical scanner (Artec Eva, Artec 3D, Hamm, Luxembourg). These scans were processed using Artec Studio 3D and converted to a model as a stereolithography (STL) file using Autodesk Meshmixer (Autodesk, San Rafael, CA, USA). Two channels were added through the medial and lateral malleoli to facilitate attaching medial and lateral ligament suture wires and to control the line of action. The model was designed to replicate the natural articular surface specific to each specimen, and 3D printed in acrylonitrile butadiene styrene (ABS) plastic, to support the union of the MIL-Lx ATD shaft and the PMHS foot.

Finally, to address the challenges associated with aligning the MIL-Lx tibia shaft at a 90° angle to the plantar surface of the foot (neutral posture), a custom steel machined component was designed to secure the 3D printed component to the MIL-Lx ATD shaft. A ball joint was created in the 3D printed tibia to allow the MIL-Lx to rest at a 90° angle to the plantar surface of the foot. This alignment was confirmed by resting the 3D printed component in its natural setting on top of the PMHS talus and placing the steel attachment proximal to this. A bull's eye level rested proximal to the steel attachment, and once precisely levelled, polymethylmethacrylate (PMMA) (Simplex P Bone Cement, Stryker, MI, United States) was used to fill the gap between the 3D printed component and the steel attachment in an effort to prevent the fixture from rotating upon impact (Fig. 4).

The machined component had a series of eight threaded holes around the circumference of the part to allow for tendons to be attached. The entire specimen was then inserted into a sealable plastic bag, to facilitate ease of insertion into the instrumented boot and maintain the cleanliness of the boot. The sutures were secured tightly such that there was visible tension in each of the suture wires, as recommended by an orthopaedic surgeon, and performed by the same researcher each time. The sutured tendons on each specimen were secured to the same bolt in the steel attachment to keep consistency among specimens.

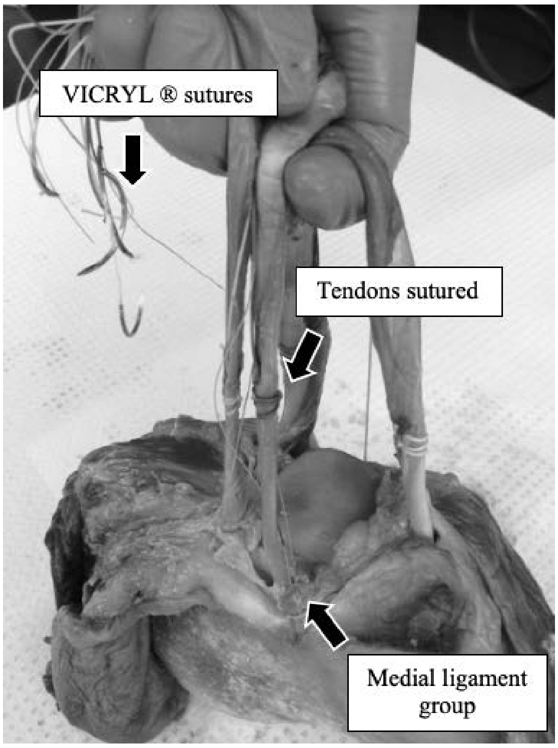

Fig. 3. The four tendons and the medial and lateral ligament groups surrounding the ankle joint were sutured using a Krakow stitch, in order to facilitate securing the PMHS foot to the MIL-Lx tibial shaft.

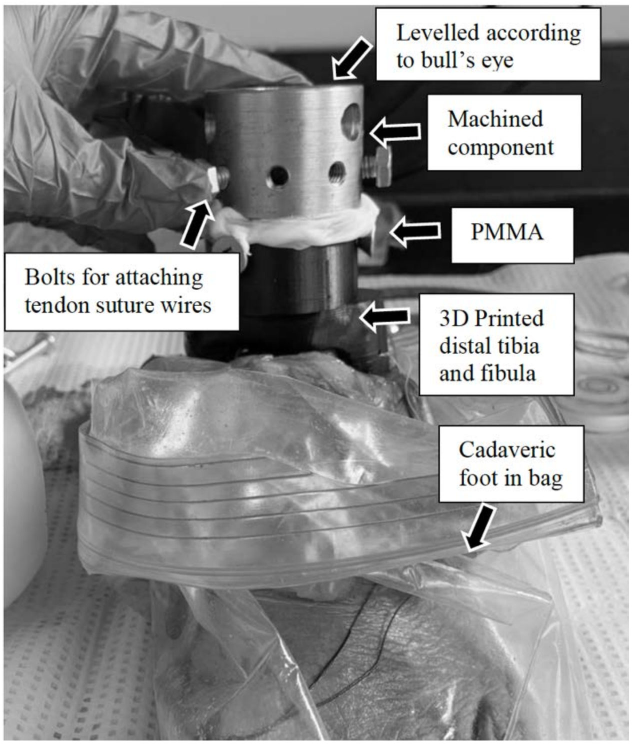

Fig. 4. Components of the newly formed ankle joint, ready for attachment of the MIL-Lx tibia shaft to a PMHS foot.

Impact Conditions

Forces developed during impact are dependent on the stiffness of the object being struck, and as this varies among specimens, is challenging to target. However, the kinetic energy of the impact is controllable, so this was used as the target impact parameter in this study. All impacts were intended to be delivered at a velocity of 5 (± 0.5) m/s and an impact duration of 20 (± 5) ms, intended to be in the range of realistic impact conditions experienced by the lower leg resulting from a frontal collision [8][25]. A settling impact was performed at the start of each specimen to seat the foot within the boot (impact mass 3 kg, kinetic energy of 20-25 J). In order to increase impact energy while duration and velocity remained constant, the projectile mass was increased. An impact mass of 6 kg and impact energy of 80 J was used for all comparative impacts. This was considered a medium-energy impact and was designed to be at a sub-failure level.

The testing procedure was controlled, and data were collected, using a custom-written LabVIEW (National Instruments, Austin, TX, USA) program. All data, including the two 5-axis load cells (Fx, Fy, Fz, Mx, My) in the upper and lower tibia and the eight instrumented boot sensors were recorded at 50 kHz. The data collected from the sensors on the instrumented boot were assessed for peak force and distribution of force along the plantar surface of the foot. Sensors were grouped into three regions for data presentation and analysis: the forefoot, midfoot and hindfoot.

Data Analysis

Data collected from the boot for all three lower leg representations were comparatively assessed. In the adapted leg form and the MIL-Lx, tibia load cell data were dual-pass filtered using a second-order Butterworth low-pass filter with a cutoff frequency of 1,250 Hz, in accordance with industry standards [24]. Impact duration was considered to have begun 1 ms before the boot hindfoot sensor (Sensor 1) decreased to 10% of the peak voltage and concluded 1 ms after the voltage fell below 10% of the peak voltage. Impact duration was calculated in the same manner at the proximal tibia for the MIL-Lx and adapted legform. Changes in sensor voltage on the instrumented boot were converted to force readings according to a previously developed calibration protocol [13]. The mean and standard deviation of each metric were also calculated.

A one-way Analysis of Variance (ANOVA) with post hoc Tukey test was conducted on both the net boot forces and regional forces for all three leg representations. An unpaired *t*-test was conducted to compare the load cell peak axial forces between the adapted leg form and the MIL-Lx, for both the proximal and distal load cells. An unpaired *t*-test was also conducted to compare impact velocities and durations among lower legs. Each of these tests had a significance threshold of $\alpha = 0.05$.

III. RESULTS

Impact velocities ranged from 5.4 to 6.4 m/s, and impact durations were 21 ± 0.4 ms. No significant differences were found among impact velocities ($p = 0.37$); however, impact durations were significantly higher in the adapted specimens as compared to the intact PMHS specimens ($p = 0.02$). The kinetic energies of impacts were significantly higher in the MIL-Lx in comparison to the intact specimens ($p = 0.03$). X-rays pre- and post-impact confirmed there was no damage to the specimens at any stage of the process. The average kinetic energy of all impacts was 78 ± 15 J (Table 2).

TABLE II
LOWER LEG IMPACT INFORMATION

Specimen	Velocity (m/s)	Kinetic Energy (J)	Impact Duration (ms)
MIL-Lx	5.4 ± 0.1	93 ± 2	19.6 ± 3.1
Intact Specimens	4.9 ± 0.7	71 ± 14	17.2 ± 4.2
Adapted Legform	5.1 ± 0.6	74 ± 12	24.7 ± 4.0

For the net boot forces (representing the complete reaction force from the plantar surface of the foot), the MIL-Lx measured 2764 ± 592 N, intact lower limbs measured 3139 ± 871 N, and the adapted legforms measured 2553 ± 899 N (Table 3). Based on the ANOVA, no statistical differences were found among lower leg representations when comparing net boot forces, suggesting that overall reaction force at the plantar surface was comparable among legforms ($p = 0.48$, Fig. 5).

TABLE III
LOWER LEG MEASURED OUTPUTS

Specimen	Total Insole Force (N)	Peak Proximal Force (N)	Peak Distal Force (N)
MIL-Lx	2764 ± 592	2003 ± 60	2005 ± 52
Intact Specimens	3139 ± 871	N/A	N/A
Adapted Legform	2553 ± 889	1538 ± 236	1538 ± 202

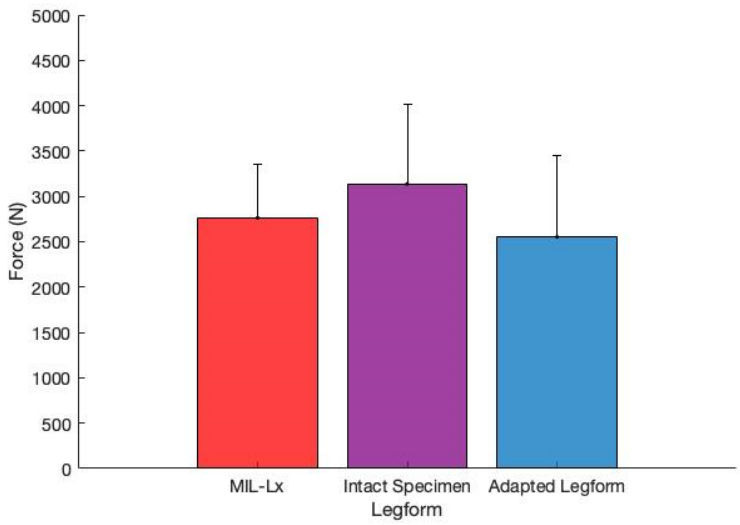

Fig. 5. Comparison of the total insole force collected from the plantar surface of the foot for all legforms.

The proximal tibia forces measured from the MIL-Lx were 2003 ± 60 N, while the adapted specimens measured 1538 ± 236 N. In comparison to the net boot forces, the MIL-Lx proximal tibia measured forces 38% lower than the plantar surface, and similarly the adapted legform read forces 66% lower in the proximal tibia rather than the plantar surface of the foot. The MIL-Lx measured significantly higher forces than the adapted surrogate for both the proximal ($p = 0.005$), and distal ($p = 0.002$) load cells (Fig. 6).

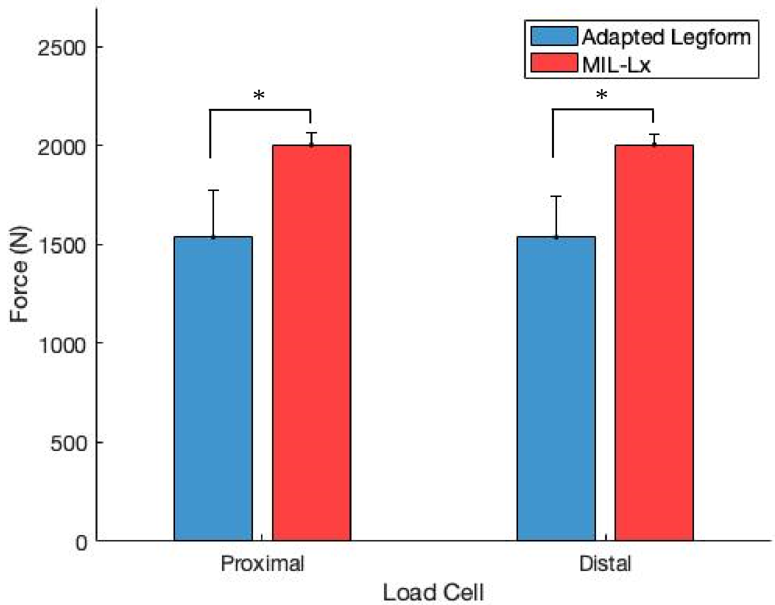

Fig. 6. Proximal and distal tibia load cell forces of adapted specimens compared to the intact MIL-Lx surrogate (* = $p < 0.05$).

The proximal and distal force-time curves for both the MIL-Lx and the adapted legform followed similar trends. Results from the unpaired *t*-test showed durations were significantly higher ($p = 0.008$) in the adapted specimens (24.1 ± 0.3 ms) than the MIL-Lx (18.2 ± 0.01 ms). The MIL-Lx proximal tibia load cells (2003 ± 60 N) were also significantly higher ($p = 0.003$) as compared to the adapted surrogate (1538 ± 236 N).

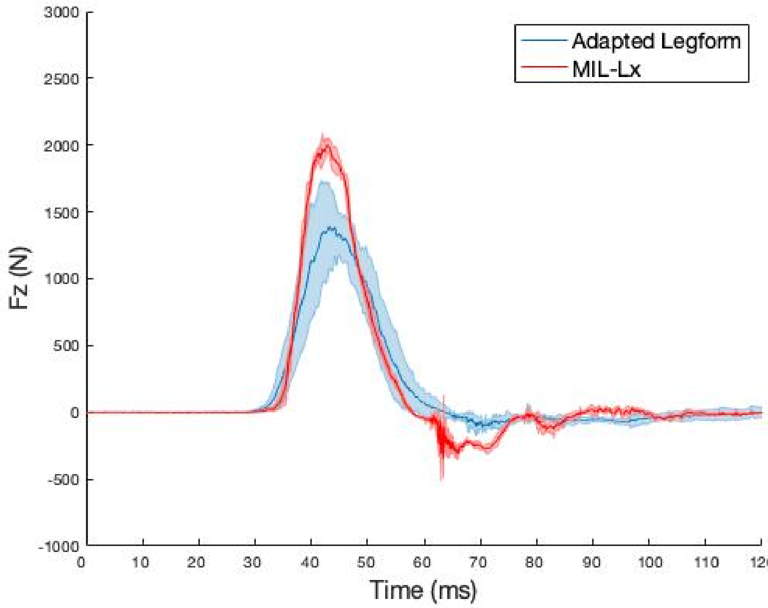

Fig. 7. Force-time curves for the adapted legform and MIL-Lx proximal tibia load cells. The line represents the mean of repeated impacts and the shaded region represents the repeated trials. Impacts were aligned according to the peak proximal force.

The hindfoot region carried the majority of the load for all impacts, ranging from 46-78% depending on the lower leg type, with the MIL-Lx consistently recording the largest hindfoot forces (Fig. 8). Based on the ANOVA, no statistical differences were found among lower leg representations for forefoot readings ($p = 0.7$) or hindfoot readings ($p = 0.2$). However, the intact specimen midfoot readings were significantly different from the adapted legform and MIL-Lx midfoot readings ($p < 0.05$).

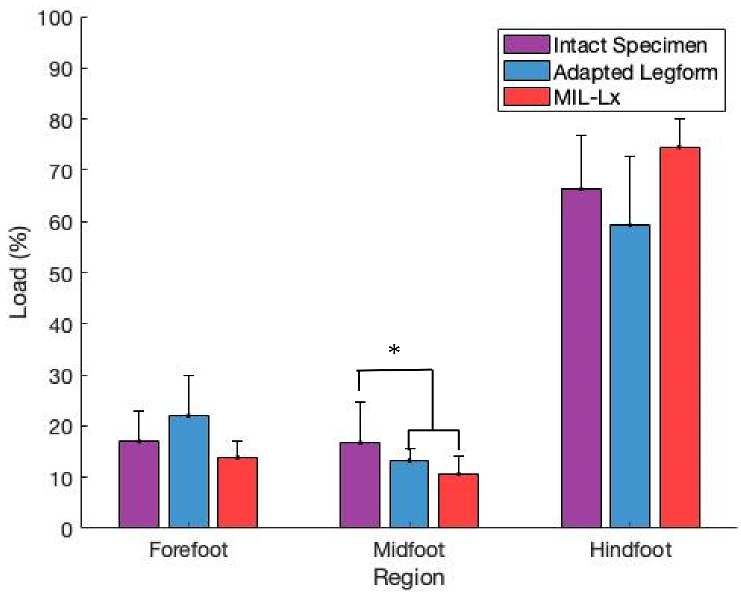

Fig. 8. Regional loading comparisons of all lower leg representations, where the forefoot loads were collected from Sensors 4, 5, and 8, the midfoot loads were collected from Sensors 3, 6, and 7 and the hindfoot loads were collected from Sensors 1 and 2 (* = $p < 0.05$).

IV. DISCUSSION

This study subjected three lower leg representations (the MIL-Lx, intact PMHS lower legs, and adapted lower legs with the MIL-Lx tibia shaft and PMHS feet) to axial impacts for the purpose of evaluating the differences in impact response among surrogates. This study is the first of its kind to investigate the isolated PMHS foot as well as the regional response characterised as a result of axial impacts. Impact time durations and velocities were consistent with those measured during vehicular collision scenarios [8]. The plantar surface force remained generally consistent among all specimens, suggesting this was a feasible method for combining testing subjects. Through the development of a novel technique to evaluate force while collecting ATD metrics, this may be used in the future to refine current injury criteria in an effort to predict foot/ankle injury more accurately. This adapted surrogate technique will be used for evaluating the fracture tolerance of isolated PMHS feet.

Forces measured on the plantar surface of the foot were over 35% higher than forces measured in the tibia, which is unsurprising when considering the location of load cells and direction of loading. No significant differences were found among impact velocities or kinetic energies, so changes in collected forces were likely related to differences in stiffness. Net boot forces among all lower legs were comparable but the adapted legform tibia forces were lower and had longer durations, suggesting the MIL-Lx foot is more rigid. In contrast, perhaps either the MIL-Lx tibia is too compliant or having many compliant elements in series (like the boot) alters the response, as suggested by [26]. This emphasises the importance of understanding the relative stiffness in series and creating transfer functions as needed that account for the various compliant elements.

Forces measured in the proximal tibia load cells were 30% higher with the MIL-Lx foot in comparison to the adapted lower leg representation. The differences in force dissipation that the MIL-Lx exhibited in comparison to the PMHS foot suggest the MIL-Lx foot may be overly stiff. This is likely a result of changes in material composition, and emphasises the importance of developing an ATD foot that has similar characteristics to PMHS feet, as the foot transmits the load to the tibia, where injury risk is assessed. This also has implications that testing completed with the MIL-Lx foot may '*fail*' the 2.6 kN injury threshold for the MIL-Lx, but with a more biofidelic foot may '*pass*' [27].

The significantly higher midfoot forces in the intact legform may have been a result of the stiffer ankle in the MIL-Lx in comparison to the intact and adapted leg forms, which allow for more ankle joint motion. Another explanation may be due to variations in the arch and soft tissue as a result of ankle stiffness. Interestingly, the distribution of forefoot and hindfoot forces were not significantly different among all lower leg representations, despite variations in foot size. The PMHS feet varied in size, which caused the forefoot readings to be slightly reduced in some specimens. Although Sensor 8 (toe sensor) did not record much data for these specimens, the forefoot sensor group was not substantially affected.

The majority of previous studies have developed injury criteria for the entire lower leg. Although some studies have examined foot/ankle injuries, these were completed in combination with injuries to the tibia and fibula [25]. By focusing on isolated PMHS feet, while also collecting tibia load data, this study assessed the impact characteristics of feet specifically. This is important as the foot/ankle complex is a vulnerable and frequently injured area in frontal collisions and the long-term implications can be debilitating. The injury mechanism is dependent on orientation of loading [28], and as such must be appropriately characterised.

There were a few limitations to the current study. Firstly, this study was completed with a small sample size (N = 6). Due to limited donor availability, the specimens were from an older population (average age of 80 years) and with both males and females. However, when analyzing male and female results separately, no significant differences were found among total insole forces, or tibia forces. The specimens also had varying foot lengths, which may have had implications when comparing to the MIL-Lx foot, considering its larger size. The smaller feet had reduced forefoot readings, which likely led to larger standard deviations observed herein. Furthermore, tests were conducted with lower forces than would likely have been conducted in vehicular collision tests. However, evaluation of the adapted legform was compared to the original same specimens, which is advantageous as this allowed comparison of adapted lower legs to its intact case, serving as its own control. Next, the MIL-Lx was optimized with respect for biofidelity for impacts of higher energy and shorter duration that were conducted here. Although this does not represent exact durations that the MIL-Lx was designed for, this study aim was to present a method of attachment, which may be used at injurious levels in the future. Testing a greater number of specimens with the protocol described herein will establish better foot and ankle injury standards when tested at injury-generating levels.

The sensors used for this study have been tested under a range of impact conditions corresponding in speed and impact duration to automotive crashes [17]. Boot forces had higher variations among impacts as compared to the MIL-Lx, as the boot will be a tool for determining regional loading and equivalence between things but would need improvements to be a reliable enough tool for any injury prediction application in the real world. Pilot testing of the sensors indicated that they were relatively insensitive to shear loading, so axial impacts are characterized well [17]. Differences in stiffness between PMHS and ATD foot may have affected bending of the sensors differently. However, the sensors were situated just proximal to the outsole of the shoe, a rigid structure, limiting the amount of bending of the sensors during loading. The sensors were calibrated using a very similar impact protocol, so the effects of bending likely did not affect these results.

Next, although the optical scanning procedure was intended to replicate natural human anatomy as closely as possible, by printing a unique mating component for each PMHS foot. However, the lack of cartilage and the artificial fixation of tendons and ligaments could have altered the responses of the feet. Every effort was made to replicate the natural ankle; however, this model could not be verified. It is possible that the load transfer between the PMHS ankle and MIL-Lx is not a biofidelic method of force transmission. As impacts were delivered in a neutral ankle posture, and in compression, this likely was not a substantial issue and although it may have affected post-impact joint motion, that was not the focus of this work.

The technique developed herein does not include the PMHS tibia or fibula. If this approach is used in future injury tests, this technique would not evaluate injury risk to the tibia and fibula, which can be injured in these types of events. However, the tibia has been well characterised in other studies [7][15][16]. Findings from [2] indicated that the primary location of fracture in axial impact tests is the calcaneus. Additionally, in tests that involved both calcaneal fractures and pilon fractures, acoustic emission results indicated the calcaneus fractured before the tibia. It is likely that fractures of the foot would occur prior to fractures of the tibia and fibula, so testing isolated PMHS feet is acceptable to determine the exact force at which fracture may occur but should be used with the understanding that tibia and fibula risk is not captured. Furthermore, it has been shown that posture affects the load pathway in the foot, resulting in different injury outcomes [11][28], however ATDs are relatively insensitive to postural changes [11][18].

The proximal and distal tibia load cells recorded forces that were very similar, indicating that the compliant element located proximal to the distal load cell was not engaging during impacts. This likens the ATD to behave more like a rigid tibia, similar to the Hybrid III, but with straight geometry. The proximal and distal load cell forces have been shown to start to diverge around 3 kN in the MIL-Lx, which corresponds to forces around 6 kN in the Hybrid III [7]. The MIL-Lx injury threshold is 2.6 kN, and the authors wished to conduct non-injurious tests in this study. Net boot forces were around 3 kN and work boots have been shown to absorb up to 65% of tibia forces [7], which was similar to findings from the present study. As there were many components acting in series during these tests, the authors wanted to ensure injurious levels of PMHS feet were not reached. In the future, fracture testing will be completed using this method, and careful attention will be paid to assess how absorption of the compliant element affects forces measured. Next, variations in tibia load cell data were likely a function of a greater number of compliant elements and interfaces in series. It is likely the system variation played some part in the differences in tibia measured forces and is a limitation of this study.

Often times when vehicular occupants see an impending collision, they will start to panic brake, activating muscles in the calf through tensioning the Achilles tendon. No Achilles tensioning was applied for testing in this study. The exact amount of tension actually activated through the Achilles is relatively unclear. Reference [2] based their 1.5-2 kN of Achilles tension on pedal forces measured during braking from volunteer driving simulations. The effects of muscle tension may be better investigated using numerical models [29]. Furthermore, the ligaments and tendons surrounding the ankle joint were sutured and secured to the machined and 3D printed components. The level of tensioning was not recorded for each specimen; however, this was not meant to pretension the tendons and ligament groups, but rather was conducted in an effort to mimic natural load transmission pathways. The purpose of suturing the tendons and ligaments surrounding the ankle was primarily to hold the PMHS talus to the artificial tibia and fibula. This role is highly focused on the post-impact kinematics, which are outside of the scope of the present study, and less important in compressive tests such as the present study. Current ATD models used in automotive collision testing do not incorporate Achilles tension, although it has been proposed for the THOR-Lx. In order to provide the best comparison with the MIL-Lx, which does not take this into consideration, muscle activation was neglected herein. Although Achilles tension may affect fracture mechanism, for the purposes of evaluating the adapted legform, it was considered not necessary.

In the future, it would be interesting to investigate the MIL-Lx response at higher energy levels, as well as with the foot situated in different ankle postures. Repeating this investigation with both the Hybrid III and the THOR-Lx would also compare the biofidelity of different surrogates. Furthermore, sub-injurious impact energies were used in this study to ensure repetition of impacts without damage to the PMHS. Future studies using this technique at injurious levels would allow for determination of exact tibia cutoff values that may result in a fracture from axial loading.

The data collected in this study could be used for future generations of ATD feet. This study found that ATD feet are stiffer in comparison to PMHS feet. Improvements in ATD properties to provide a more biofidelic ankle joint, or segmented foot to provide enhanced representation of the types of loading the foot/ankle complex will experience in these scenarios would be advantageous to reduce the incidence and severity of foot and ankle injuries.

V. CONCLUSIONS

This study presented a method to assess injury risk to the isolated foot while collecting data that are immediately relevant to the automotive industry. This is important as these findings may be retroactively applied to previous studies, with an adjustment factor considered due to differences in compliance of the ATD foot. The similar force readings collected at the plantar surface of the foot showed that doing this did not affect the load response to the foot. Results suggest that the MIL-Lx foot could be improved in stiffness characteristics, and this study has provided data that can be used for this design. It is the first study of its kind to propose an adapted lower leg in order to assess isolated foot injuries while gathering axial force data.

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