Development of a Transfer Function for Interpreting Hybrid-III Lower Leg Data from Axial Loading

Ann M. Bailey, Matthew B. Panzer, Robert S. Salzar

Abstract The Hybrid-III anthropomorphic test device is commonly used for evaluation of human lower extremity injury risk in environments such as automotive intrusion and underbody blast, despite previous literature showing differences between the Hybrid-III and human lower leg mechanical response at faster loading rates. To interpret the Hybrid-III leg data for use in determining human injury risk, transfer functions were developed relating tibia forces from both the Hybrid-III and post mortem human surrogates (PMHS) to axial impact data with loading rates ranging from 100 to 600 g's of peak acceleration with durations of 6.8 to 10 ms and impact velocities between 1.5 to 10 m/s. A parametric study using two finite element models simulated with the same impact conditions was performed, and upper tibia force response was compared between the models. Transfer functions were developed to relate the proximal tibia force responses from each model. Additionally, existing injury criteria were used to determine injury risk for the human based on the Hybrid-III force output. When using existing injury criteria, the transfer function predicted a 50 percent human injury risk when the Hybrid-III axial upper tibia force was 13 kN.

Keywords Axial loading, finite element, Hybrid-III, transfer function

I. INTRODUCTION

The Hybrid-III anthropomorphic test device (ATD) is commonly used for evaluation of human lower extremity injury risk in environments such as automotive intrusion and underbody blast (UBB), despite previous literature showing differences between the Hybrid-III and human lower leg mechanical response at faster loading rates [1-2]. The axial stiffness of the Hybrid-III lower extremities is substantially higher than a human leg, which poses a challenge for interpreting the large database of existing Hybrid-III test data as it relates to human response and injury risk.

Recent research efforts have focused on higher rate axial loading of the lower leg in attempt to characterize the leg's response and injury characteristics [1-4]. While these efforts were performed in the laboratory setting with PMHS, many of the live-fire experiments performed with a goal of determining injury probability for different severity levels of UBB and landmine blasts use ATDs. Despite the existence of more advanced ATD legs like the Mil-Lx and Thor, the Hybrid-III leg is the primary ATD used to gather data from live-fire tests performed by the Live Fire Test and Evaluation Program (LFT&E) [5]. As more information becomes available from these live-fire tests, it is imperative that a standardized procedure exists for determining the response of the human leg from the ATD response. This response can then be mapped to a status of injury or no injury using new or pre-existing lower leg injury criteria [3-4][6-7] in order to be useful for determining survivability of different severities of blast in different vehicle structures.

Previous efforts to correlate human and Hybrid-III lower leg forces have shown that for short duration loading events (<55 ms rise time), the Hybrid-III produces higher forces than the human leg for the same load profile [8]. Prior attempts have been made to develop a relationship between the Hybrid-III and human legs [8-12], however little effort was focused on validating the transfer functions for the higher rates of loading associated with UBB. Kuppa et al. determined stiffness and damping coefficients for one degree of freedom analytical models of the human and original Hybrid-III legs, and used them to create a transfer function for the force output [8]. However, the stiffness coefficient was significantly different from the previous stiffness developed for the human leg because of the loading rates used in the experiment [13]; therefore a transfer function

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created for use in UBB scenarios must either be validated for a larger range of loading rates or specific to UBB loading conditions. Another attempt at developing a transfer function was made by McMaster et al., by determining a relationship between the input and output forces, and the time to peak force in the Hybrid-III and human leg. This relationship was then used to produce the first injury risk curve based on Hybrid-III forces [12]. Despite its existence, the transfer function has not been utilized or validated.

Limitations such as lack of high-rate matched-pair experimental data, and non-rate dependent finite element models have previously prevented the development of a transfer function for determining the relationship between Hybrid-III and human forces under high loading rates. To solve this problem, Hybrid-III and post mortem human surrogate (PMHS) axial impact tests performed using a drop tower with loading rates ranging from 100 to 600 g's of peak acceleration with durations of 6.8 to 10 ms and impact velocities between 1.5 to 10 m/s were used to develop rate-dependent finite element (FE) models. The FE models were then used to develop a relationship between Hybrid-III and human leg forces for use in determining human lower extremity injury risk.

II. METHODS

This research effort combines both finite element modeling and experimental testing to develop a method by which to interpret Hybrid-III lower extremity data. By utilizing PMHS and ATD lower extremity component test data to improve existing human and Hybrid-III lower leg FE models, matched-pair tests could be simulated for specific boundary conditions, which enabled comparisons for rates at which experimental data do not currently exist. These data were then used to form a relationship between the force in the human and the force in the Hybrid-III lower leg.

Experimental Testing

PMHS and Hybrid-III component tests were performed using the drop tower test rig (Fig. 1) described by Henderson et al. [4]. Impact plate, load cell and tibia accelerations were recorded for both PMHS and Hybrid-III tests. Strains along the length of the anterior tibia as well as around the circumference of the distal tibia were recorded for the PMHS tests while upper and lower tibia forces were recorded for the Hybrid-III tests.

In the Henderson et al. study, each leg was sectioned above the proximal epiphysis of the tibia and fibula, and instrumented with accelerometers, strain gauges and load cells. The lower legs were mounted at the bottom of a drop tower equipped with an impactor capable of producing axial loading on the PMHS foot up to 600 g's acceleration over a 1.5 ms duration.

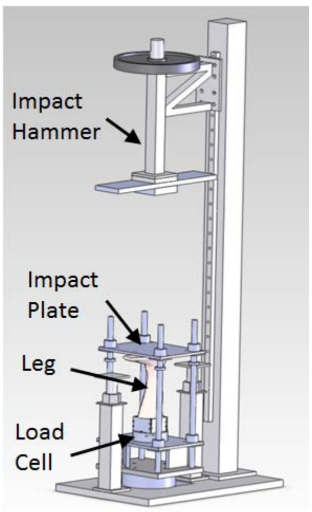

Fig. 1. Drop-tower assembly used to obtain the experimental results [4]

For the Hybrid-III tests, two tests were performed at each of three test conditions, which varied by pulse shaper, with the drop tower hammer dropped from a height of 0.5 meters and the proximal end of the lower leg free to translate in the z-direction away from the applied load. The boundary conditions from the Henderson et al. study were also replicated using the Hybrid-III. Upper and lower tibia axial force and mid-tibia acceleration were collected at 100 kHz using a Slice-Pro data acquisition system (DTS, Inc., Seal Beach, CA), in addition to hammer and footplate acceleration. These data were used for identifying limitations in the current Hybrid-III FE model and helped direct model improvements.

Computational Lower Leg Models

Simulations of the drop tower tests were done using an existing Hybrid-III FE model (Livermore Software Technology Corporation, Livermore, CA) and the Global Human Body Model Consortium (GHBMC) human lower extremity FE model. Both model setups of the University of Virginia drop tower and the leg were the same (Fig. 2): an acceleration boundary condition (taken directly from each experiment) was applied to the rigid aluminum plate that impacted the inferior surface of the foot, while the proximal end boundary condition of the lower leg was modeled using a rigid plate and mounting fixture with masses matching those in the experiment. An automatic surface-to-surface contact was placed between the platen and the foot, and a preload of 100 N was applied at the foot to correspond to the platen resting on the prior to the experimental impact. Comparisons between the FE model and the physical tests were made by focusing on load cell forces and distal tibia strains.

To compare with the PMHS tests performed by Henderson et al. [4], the Phase I lower extremity FE model of the GHBMC 50th percentile male was used. This model was developed by the University of Virginia for automotive loading scenarios, and is validated for a wide range of loading conditions [14-15]. This model was simulated under conditions stemming from 11 of the 18 previously described lower leg impact experiments. An initial series of simulations was run using the original unmodified version of the lower leg model to establish a baseline comparison with the experimental results. Comparison between original model results and experimental data revealed the FE model lacked certain features necessary to accurately capture the biomechanics of the tests. Model improvements were made, including the addition of viscoelasticity to the foot flesh using a visco-hyperelastic constitutive model to capture the dynamic heel-pad mechanics [16], and refining the original model's bone tissue meshes to produce more accurate geometry and better model bony fracture. The latter modification increased the number of through-thickness elements in the fibula and tibia shafts to two, thereby improving the tibia bending stiffness to a more biofidelic response. The simulations were rerun using the modified model with the same input conditions as the original model, and the results were compared (Fig. 3).

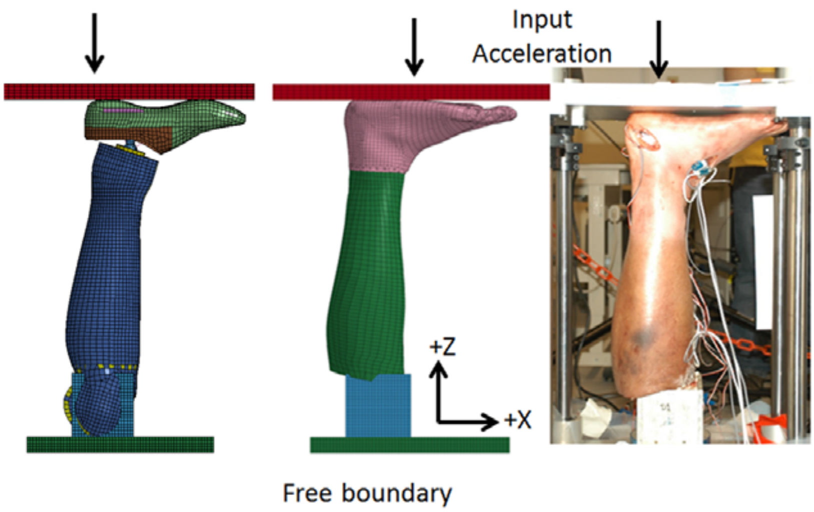

Fig. 2. Hybrid-III and human FE models and the experimental setup

To compare with the Hybrid-III drop tower tests, an initial series of simulations was also run using the original LSTC Hybrid-III leg FE model. The initial comparison of this model to the experimental data also demonstrated the model was inadequate for this test. Improvements were made to the FE model to more accurately represent the geometry of the Hybrid-III foot (including heel thickness) and ankle, and the foot rubber constitutive model was updated to capture the rate-dependent properties of the Hybrid-III skin [17]. These improvements greatly enhanced the fidelity of the ATD FE model in all simulations performed (Fig. 4), and showed that the modified FE model more closely replicates the response of the Hybrid-III tibia in the experimental tests than the unmodified FE model.

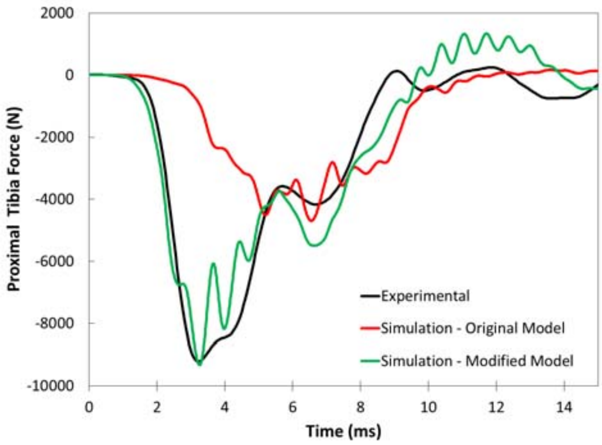

Fig. 3. Modified human leg FE model response

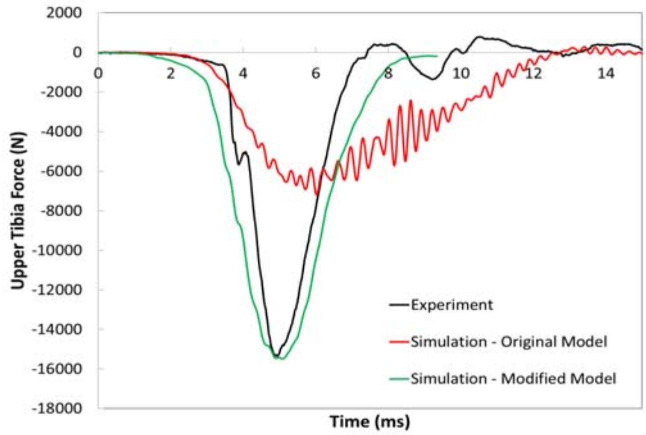

Fig. 4. Modified Hybrid-III leg FE model response

Parametric Analysis

After making modifications to both the human and Hybrid-III leg FE models, a parametric study using the two FE models simulated with the same impact conditions was performed, and upper tibia force response was compared between the models. Five sets of three triangular acceleration pulse impacts (15 total) were used, with each set varying in time to peak, duration and peak acceleration (Table 1). For each set, three different platen acceleration pulse shapes (Fig. 5) were created which had the same peak acceleration, duration and peak velocity, but differed by rate of acceleration. These pulse shapes were designed to represent the wide-range of floor pan accelerations expected in UBB (durations shorter than 10 ms and average acceleration greater than 30 g) [18].

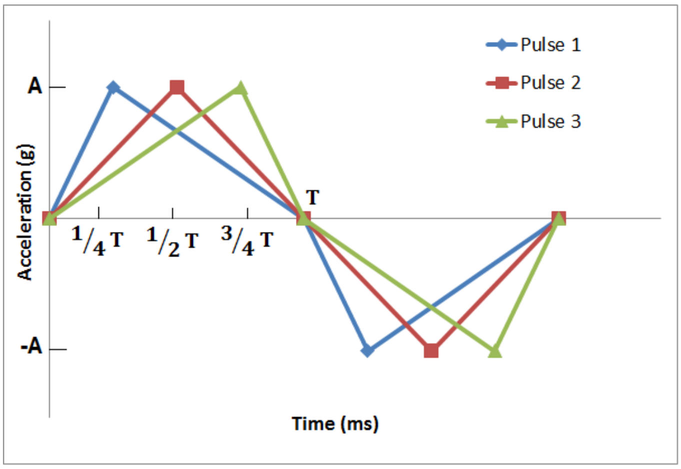

Fig. 5. Characteristic acceleration boundary condition inputs (refer to Table 1 for magnitudes)

TABLE 1
INPUT PULSE PARAMETERS FOR FE MODELS

Pulses	Acceleration [A] (g)	Duration [T] (ms)	Peak Velocity (m/s)
1-3	300	3.4	5
4-6	300	5	7.36
5-9	600	3.4	9.97
10-12	100	5	2.54
13-15	100	3.4	1.67

Transfer Function Development

A transfer function for both the upper and lower Hybrid-III tibia force was developed to require only one load cell to determine the tibia force in the human leg necessary for using existing injury criteria. To develop the transfer function, MATLAB (Mathworks, Inc., Natick, MA) was used to create an isochronic force curve for each of the 15 input conditions using the force-time history from both the FE model output for the Hybrid-III and human legs. A characteristic average curve was then fit to the isochronic force curves (Fig. 6). The average curve was then fit with a second order polynomial equation (1).

$$F_{Hybrid\text{-}III} = A\, F_{human}^{2} + B\, F_{human}, \tag{1}$$

where *A* and *B* are coefficients listed in Table 2 and *F* refers to the axial force in the leg (N).

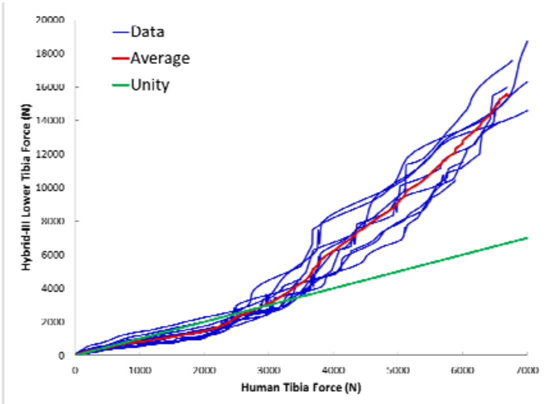

Fig. 6. Characteristic isochronous force curve comparing human and Hybrid-III tibia forces

TABLE 2

FORCE TRANSFER FUNCTION PARAMETERS

$F_{Hybrid\text{-}III}$	*A*	*B*
Lower Tibia Force (N)	3.251_E-4	0.176766
Upper Tibia Force (N)	2.97522_E-4	0.1378235

To compensate for the timing differences between the human and Hybrid-III force curves (Fig. 7), a similar approach was taken, instead creating time curves while holding percentage of peak force constant. An average curve was determined based on the results from the FE models, and a polynomial curve was fitted according to equation (2).

$$t_{human} = \alpha\, t_{Hybrid\text{-}III}^{2} + \beta\, t_{Hybrid\text{-}III}, \tag{2}$$

where *t* refers to time, and the parameters *α* and *β* are presented in Table 3.

TABLE 3

Time Transfer Function Parameters

$t_{Hybrid\text{-}III}$	*α*	*β*	R^2
Lower Tibia	0.0639	0.669	0.9983
Upper Tibia	0.0663	0.6603	0.9982

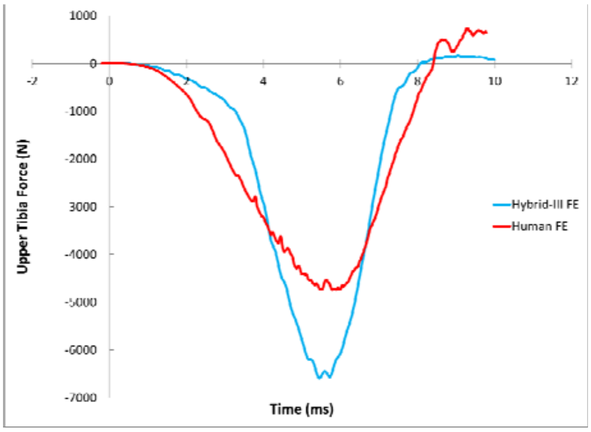

Fig. 7. Timing difference between Hybrid-III and human finite element models for Pulse 10.

III. RESULTS

The transfer functions were used to predict the response of the human FE model by using the output of the Hybrid-III FE model to predict the human FE results, and were found to be in good agreement (Table 4). In general, the human FE force-time history was well captured using the Hybrid-III FE model and the transfer function; the loading rates and maximum force values were consistent, but the unloading rates were slightly higher in the predicted response (Figs. 8-9).

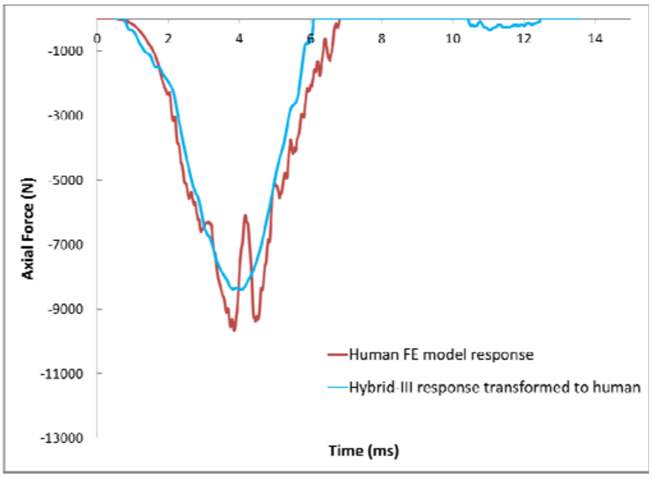

Fig. 8. Human FE model and predicted results based on transfer function applied to Hybrid-III FE response for Pulse 1.

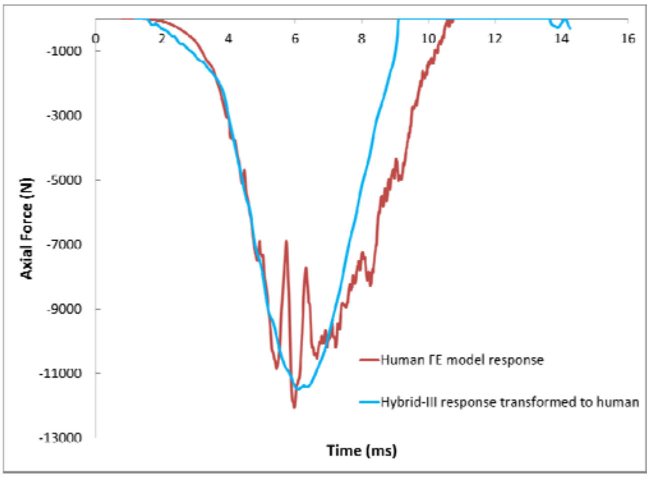

Fig. 9. Human FE model and predicted results based on transfer function applied to Hybrid-III FE response for Pulse 6.

TABLE 4

Coefficient of determination for the comparison of the human FE force response and the force response predicted using the transfer function for each of the 15 tests

Pulse	R^2	Pulse	R^2	Pulse	R^2
1	0.9897	6	0.9959	11	0.9942
2	0.9936	7	0.9920	12	0.9937
3	0.9954	8	0.9959	13	0.9748
4	0.9920	9	0.9969	14	0.9606
5	0.9957	10	0.9938	15	0.9514

The transfer function was then used to interpret experimental Hybrid-III data and predict human leg forces. Results from a series of five matched whole body PMHS and Hybrid-III tests performed by Bailey et al. [19] were used to assess the transfer function as a tool to better predict injury using existing Hybrid-III lower leg injury criteria (Figs. 10-11). The predicted values were also compared to Mil-LX experimental data also from the same test condition, and the peak magnitudes were found to be within 13% of the peak Mil-LX force whereas Hybrid-

III forces over-predicted the Mil-LX force by over 41%. Lines showing probability of injury for a given upper tibia force are shown based on the injury curves developed by Henderson et al [4].

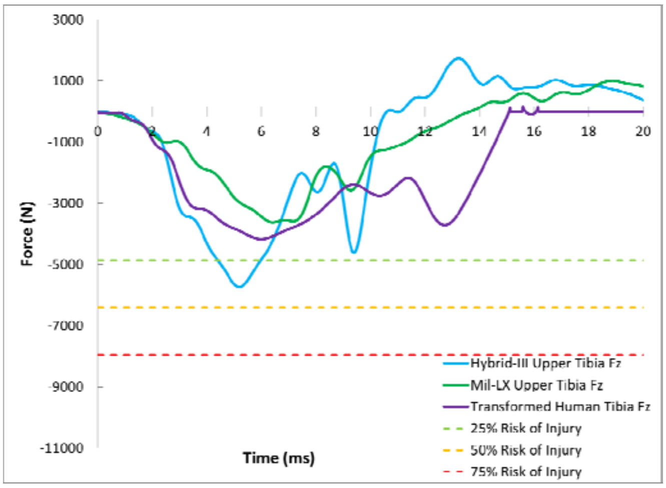

Fig. 10. Transformed human tibia force from whole body test performed by Bailey et al. (Test 1.3) [19]

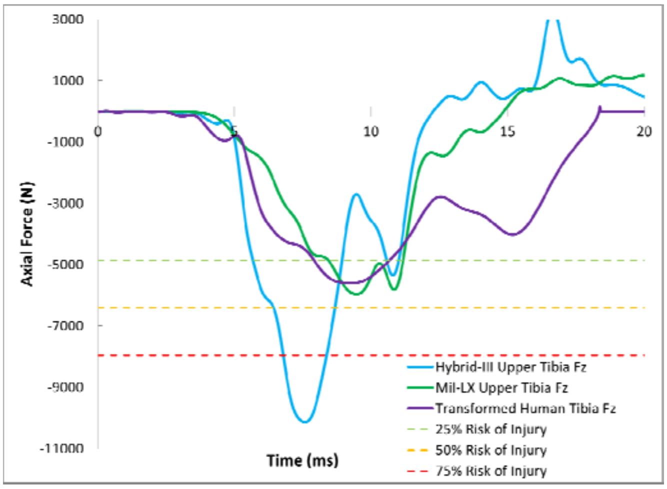

Fig. 11. Transformed human tibia force from whole body test performed by Bailey et al. (Test 1.5) [19]

IV. Discussion

There is a lack of matched PMHS and Hybrid-III lower extremity data for high-rate axial impact, making it difficult to properly assess the risk of injury when Hybrid-III ATDs are used in UBB testing. To better understand the relationship between the human and ATD responses, FE models were used to develop a transfer function between Hybrid-III and human leg forces. The eventual use of the transfer function is to improve the prediction of human lower extremity injury risk in UBB.

While this transfer function allows for determination of force-time histories for the human FE model from the Hybrid-III FE model, it is limited to the fidelity of those models. Differences in timing between the experimental and FE Hybrid-III forces result in phase differences between the predicted human force and the experimental tests. However, despite the slight inaccuracy of the Hybrid-III FE model, the approach presented in this paper lays the foundation for the future when a more accurate FE model is developed. While discrepancies exist in the timing of the predicted human force curves, the predicted magnitudes are reasonable. McKay found Mil-LX peak forces to be comparable to those of the human leg, with slight differences in load duration, which suggests the human leg forces predicted using the transfer function are somewhat accurate (see Fig. 10-11).

Since no matched-pair tests capable of measuring tibia force in the PMHS were performed, experimental work performed by Bir et al. [1] was used to determine the validity of the transfer function. Comparing the human tibia force predicted using the transfer function approximation slightly overestimates the PMHS corridor (Fig. 12). This overestimation may likely be due to the experimental tests by Bir et al. being performed at a different loading rate than for which the FE models were validated. Additionally, variations in the age of the specimens used for the PMHS tests in the Bir et al. study and those used for validating the FE model could contribute to this difference.

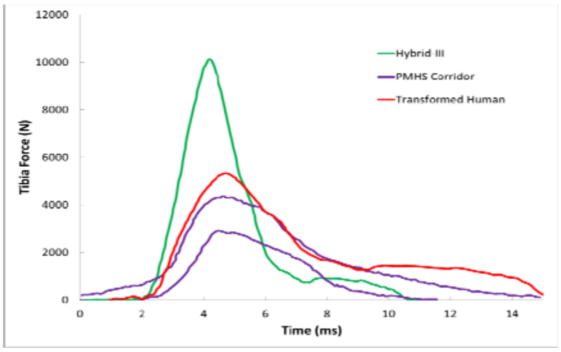

Fig. 12. Comparison of transformed human tibia force to experimental data from Bir et al. [1]

A similar approach was taken to determine the validity of the transfer function for data presented by Manning et al. [20]. A comparison of the human tibia force transformed from the Hybrid-III data from the Manning tests and the average force from the PMHS tests as measured from an implanted tibial load cell is shown in Fig 13. Again the transfer function overestimates the peak force, and the duration of the loading is also overestimated. The timing discrepancy is likely due to the inaccuracy of the timing for the Hybrid-III FE model from which the transfer function was developed, and is even more pronounced due to the much slower impact conditions used for the Manning et al. test series which were around 4 m/s impacts to the heel of the foot with an unknown time to peak. This comparison further emphasizes the necessity of a more robust transfer function with the ability to accurately predict forces over a broad range of loading rates. While the transfer function is satisfactory for use with the Bir et al. data because of the similarity in loading rates, it is unacceptable for use with automotive rate loading conditions similar to those used by Manning et al. [19-20].

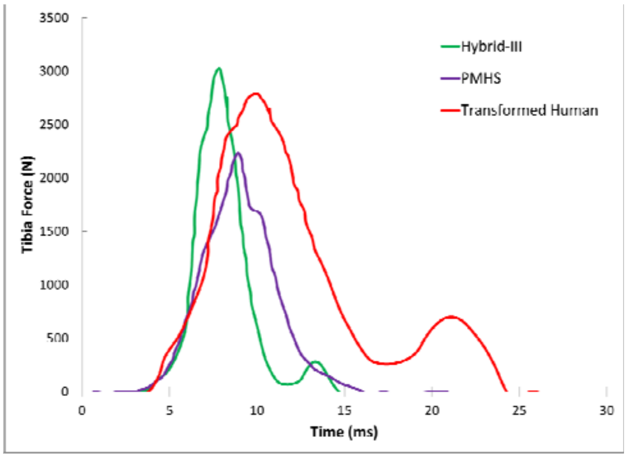

Fig. 13. Comparison of transformed human tibia force to experimental data from Manning et al. [20]

Using the transfer function to predict the forces in the human leg for the five matched-pair tests allows a comparison of the experimental injury outcome and the predicted injury outcome based on the predicted force. An example of how the predicted human leg forces align with the Henderson et al. injury criteria is shown in Fig. 14. The only injurious test (Test 1.1) was the only test that produced a force exceeding 50 percent risk of injury. While the transfer function produced useful results for this particular data set, it should be utilized with more matched-pair tests in order to fully assess its value for determining injury outcome.

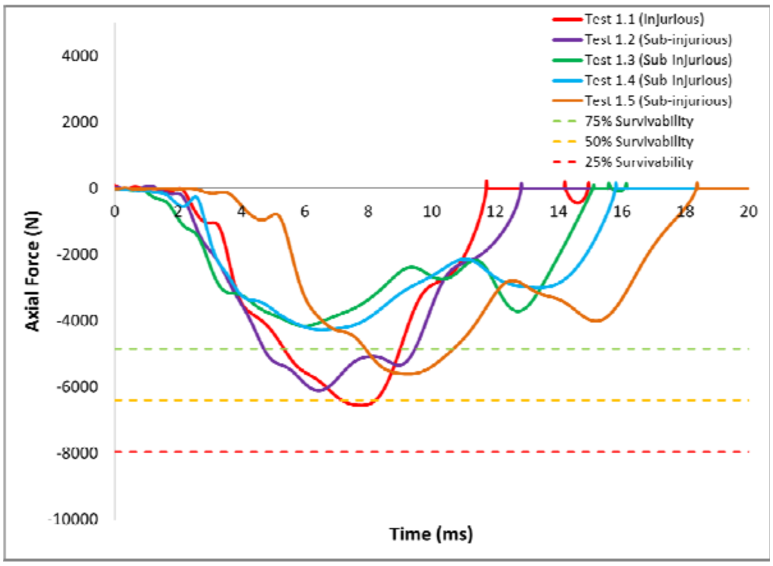

Fig. 14. Human proximal tibia forces transformed from the Hybrid-III lower tibia forces from the Bailey et al. test series [19]

The transfer function could also be used to transform existing injury criteria to equivalent Hybrid-III based risk assessments. For example, the transformation of the human injury criteria developed by Henderson et al. into a

Hybrid-III based assessment results in a 50 percent and 75 percent injury probability correspond to Hybrid-III upper tibia forces of 13 and 17.4 kN, respectively (Fig. 15). These results are significantly higher than the Hybrid-III based risk assessment curve developed by McMaster et al., where 7 kN corresponds to 50 percent risk of injury [12]. This difference may be related to a difference in loading rates or positioning used for each of the test series, or the presence of Achilles tension in the McMaster et al. tests, which has been shown by Funk et al. to increase risk of injury for a given axial load condition [6]. Additional matched-pair testing is required to determine the validity of the current suggested injury curve.

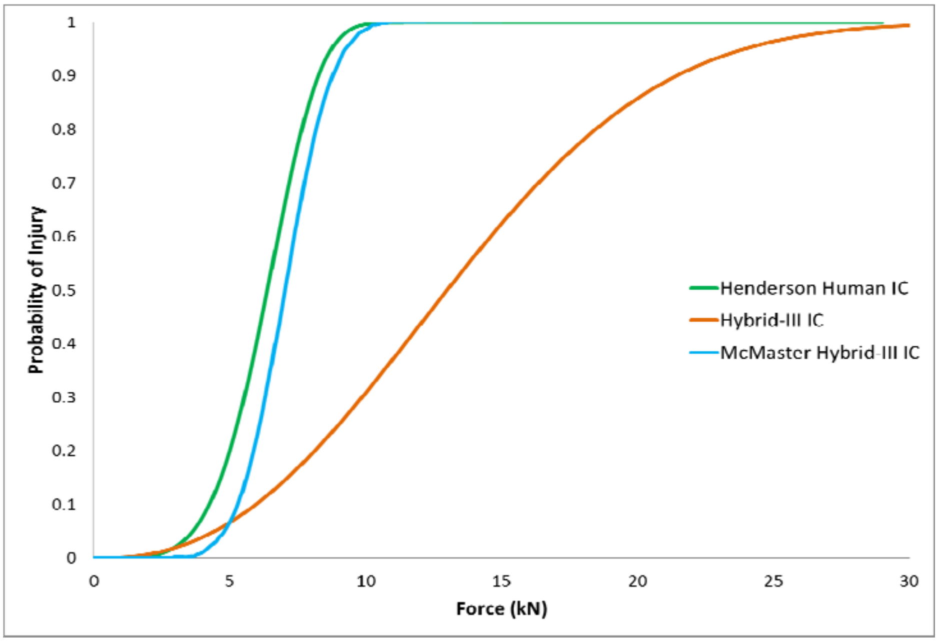

Fig. 15. Hybrid-III injury probability curve based on the human injury probability curve from Henderson et al. [4] compared to the Hybrid-III injury curve developed by McMaster et al. [12]

V. CONCLUSIONS

The transfer function developed in this study is a first step to interpreting the large database of existing Hybrid-III lower leg data from UBB events to better assess the injury potential of the vehicle occupants. Limitations of this work include the use of FE models to provide data for generating a robust transfer function, but this was deemed necessary in the absence of matched PMHS and Hybrid-III experimental data. The transfer function is only applicable for the range of impact conditions investigated, but these conditions are comparable to those measured on the vehicle floor during UBB tests. Future testing efforts will be necessary to fully validate the transfer function for a wider range of input conditions, and to verify that the injury criteria provides accurate injury prediction capabilities for the given loading rates.

While this transfer function has the capability to predict peak human tibia forces which can be used alongside existing injury criteria to determine the probability of injury, it must be used with caution. The FE models used to develop the transfer function are only validated for a certain range of loading rates. Application of the transfer function to a drastically different loading rate could produce erroneous human leg force results. A preliminary comparison to existing literature showed slight overestimation of peak human tibia force [1][19]; it was not clear from those reports if the accelerations for those tests were consistent with the loading rates used to develop the transfer function. Additional matched-pair testing should be performed in order to determine the range over which this transfer function is valid. Future research into a more sophisticated transfer function approach should be used to better capture the phasing and rate-dependent effects. However, the current approach is successful at modeling the general relationship between the Hybrid-III response and the PMHS response. Future efforts into improving the accuracy of FE models will also improve the ability to predict human tibia forces from Hybrid-III tibia forces and decipher the plethora of existing ATD data from UBB testing.

VI. ACKNOWLEDGEMENT

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VII. References

[1] Bir C, Barbir A, et al. Validation of lower limb surrogates as injury assessment tools in floor impacts due to anti-vehicular land mines. *Military Medicine*, 2008, 173(12):1180-1184.

[2] Quenneville C, McLachlin S, Greeley G, Dunning C. Injury tolerance criteria for short-duration axial impulse loading of the isolated tibia. *Journal of Trauma and Acute Care Surgery*, 2011, 70(1): E13-E18.

[3] McKay B. Development of lower extremity injury criteria and biomechanical surrogate to evaluate military vehicle occupant injury during an explosive blast event, Wayne State University Dissertation, 2010.

[4] Henderson K, Bailey A, Christopher J, Brozoski F, Salzar R. Biomechanical response of the lower leg under high rate loading. *Proceedings of IRCOBI Conference*, 2013, Gothenburg, Sweden.

[5] Department of Operational Test and Evaluation. "FY 2012 Annual Report" Internet: [http://www.dote.osd.mil/pub/reports/fy2012], 2012 [March 14, 2014].

[6] Funk J, Khaewpong N, et al. The axial injury tolerance of the human foot/ankle complex and the effect of Achilles tension, *Journal of Biomechanical Engineering*, 2002, 124(6):750-757.

[7] Yoganandan N, Pintar F, et al. Dynamic axial tolerance of the human foot-ankle complex. *Proceedings of Stapp Car Crash Conference*, 1996, 40:207-218.

[8] Kuppa S, Klopp M, et al. Axial impact characteristics of dummy and cadaver lower limbs. *Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles*, 1998.

[9] Manseau J, Keown M. Development of an assessment methodology for lower leg injuries resulting from anti-vehicular blast landmines. *Proceedings of IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications*, 2005, Springer, Netherlands.

[10] Welbourne E, Shewchenko N. Improved measures of foot and ankle injury risk from the Hybrid-III tibia. *Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles*, 1998.

[11] Cheng Z, Pellettiere J. A novel approach for predicting human response from ATD tests. *Stapp Car Crash Journal*, 2004, 48:1-19.

[12] McMaster J, Parry M, et al. Biomechanics of ankle and hindfoot injuries in dynamic axial loading. *Proceedings of 44th Stapp Car Crash Conference*, 2000, 44:357-377.

[13] Mizrahi J, Susak Z. In-vivo elastic and damping response of the human leg to impact forces. *Journal of Biomechanical Engineering*, 1982.

[14] Shin J, Yue N, Untaroiu C. Development and validation of an occupant lower limb finite element model, *Society of Automotive Engineers, SAE World Congress*, 2011.

[15] Shin J, Yue N, Untaroiu C. A finite element model of the foot and ankle for automotive impact applications. *Annals of Biomedical Engineering*, 2012, 40(12):2519-2531.

[16] Natali A, Fontanella C, Carniel E. Constitutive formulation and analysis of heel pad tissues mechanics. *Medical Engineering & Physics*, 2010, 32(5):516-522.

[17]Wood G, Panzer M, Bass C. Viscoelastic properties of Hybrid III skin. *SAE International Journal of Materials and Manufacturing*, 2010.

[18] Wang J, Bird R, Swinton B, Kristic A. Protection of lower limbs against floor impact in army vehicles experiencing landmine explosion. *Journal of Battlefield Technology*, 2001, 4(3):8-12.

[19]Bailey A, Christopher J, Henderson K, Brozoski F, Salzar R. Comparison of Hybrid-III and PMHS response to simulated underbody blast loading conditions. *Proceedings of IRCOBI Conference*, 2013, Gothenburg, Sweden.

[20]Manning P, Wallace A, Owen C, Roberts A, Oakley C, Lowne R. Dynamic response and injury mechanism in the human foot and ankle and an analysis of dummy biofidelity. *Proceedings of the 16th International Technical Conference on the Enhanced Safety of Vehicles*, 1998, Canada.