Modular Incorporation of a Detailed Lower Extremity Model into a Simplified Human Body FE Model

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ABSTRACT

Finite element human body models (HBMs) can be used to investigate injury mechanisms and tolerance of the human body during various loading conditions. The modeling complexity required to achieve good biofidelity in HBMs can lead to long compute times. This has prompted the development of faster running simplified models (GHBMC M50-OS) that can be used primarily for kinematic/kinetic comparisons. These simplified models share the same body habitus as the detailed counterparts, but with rigid bony structures and simplified modeling approaches for viscera, joints etc. Previous studies have shown the ability to modularly incorporate organs with high biofidelity models, such as the detailed (GHBMC M50-O) brain, into the simplified model. This technique allows for localized analysis of a region of interest in a fraction of the computational time required for the detailed model. The purpose of this study is to expand on this previous work by incorporating the previously-validated lower extremity of the M50-O into the computationally efficient M50-OS (M50-OS+LEx) and compare physical loading response to the M50-O in a localized knee bolster impact and a frontal sled simulation. The modularly-incorporated components include all deformable bony structures from the sacrum to the foot, detailed soft tissue structures from the femur flesh to the foot flesh, and all explicitly meshed tendons and ligaments. Total force from the femur, upper tibia, lower fibula, and ilium were obtained and compared between the two models during the knee bolster and frontal sled simulation. The implementation of the detailed lower extremity provided force time histories that tend to agree with the overall shape and magnitude of the data from the M50-O. The method introduced here, allows researchers to obtain similar force data at a much reduced computational cost (~71% time savings). Further investigation into the validation of this technique may include full vehicle buck simulations and optimization of included detailed components to ensure the best balance between time savings and performance.

INTRODUCTION

Computational human body models (HBMs) are a form of finite element models that can be used as tools to investigate injury mechanisms and tolerance of the human body during various loading conditions (Arun, et al., 2016; Chang, et al., 2008). The Global Human Body Model Consortium (GHBMC) is a family of models of various body sizes and complexities that are currently utilized for vehicle crash response. The GHBMC is an industry-sponsored and

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government supported consortium with the goal of developing computational human models for blunt impact injury evaluation.

The GHBMC average male occupant (M50-O v4.4) finite element model was used in this study. This model was developed using a multi-modality medical image and external anthropometry dataset of a volunteer representing a 50th percentile male in terms of height (174.9 cm) and weight (78.6 ± 0.77 kg). More details on the development of the M50-O can be found in Gayzik et al. (Gayzik, et al., 2011; Gayzik, et al., 2012; GHBMC, 2015). The M50-O has been extensively validated at both the regional and full body levels (Hayes, et al., 2013; Li, et al., 2010; Shin, et al., 2012; Soni, et al., 2013; Yang, et al., 2006).

The modeling complexity required to achieve good biofidelity in HBMs can lead to long compute times. This has prompted the development of a faster running simplified average male occupant model (GHBMC M50-OS) that can be used primarily for kinematic/kinetic comparisons. The simplified model retains the same body habitus and rigid bone structures as the detailed model but has simplified soft tissue structures and reduced mesh density. For example, biofidelic complexities in the joints such as ligaments were removed and the properties of bone in most of the body, including the leg, were modeled as rigid parts to improve computational performance. These factors help the M50-OS run in a fraction of the computational time of the M50-O, while retaining similar response in blunt impact scenarios. The M50-OS has demonstrated the ability to run approximately 32 times faster than the M50-O (Decker, et al., 2016; Schwartz, et al., 2015).

An increase in airbag availability and stricter seatbelt laws has reduced the amount of deaths from serious motor vehicle collisions (Kuppa, et al., 2003). However, injuries to the lower limb during motor vehicle crashes have increased and become the most common form of AIS 2+ injuries for occupants. These injuries are not usually life threatening but can lead to complications such as functional impairment and permanent disability (Dischinger, et al., 2004; Kuppa, et al., 2003). Injuries to knee-thigh-hip (KTH) complex, in particular, has been seen to make up more than half of these lower limb injuries (Kuppa, et al., 2003; Laituri, et al., 2006). Traditional research techniques with Post Mortem Human Subjects (PMHS) and crash test dummies aim to develop more complex injury criterion to determine injury risk. Conversely, computational HBMs allow researchers to calculate stress/strain distributions within the areas of injury risk. Extensive validation of the GHBMC M50-O lower limb was performed by Untaroiu et al. which included various component level testing such as tensile ligament testing (Untaroiu, et al., 2005), multi-directional loading of the long bones (Untaroiu, et al., 2013), full leg impactor validation (Untaroiu, et al., 2005), and ankle motion analysis (Shin, et al., 2012). The current state of the lower limb within the M50-O has not been altered since these validation studies.

Previous studies have demonstrated a technique to incorporate various detailed components of the M50-O into the M50-OS, in order to obtain localized analysis of a region of interest in a fraction of the computational time required for the detailed model. This technique has been used in the regions of the brain and the thoracoabdominal cavity (Decker, et al., 2017). In this study, we expand on this concept by incorporating the previously-validated lower extremity of the M50-O into the computationally efficient M50-OS (M50-OS+Lex) and compare

physical loading response to the M50-O in a localized knee bolster impact and a full vehicle buck simulation.

METHODS

Detailed Lower Extremity Modular Incorporation

The lower extremity of the M50-OS v1.8.4 was modeled with rigid bones surrounded by a coarse 3D mesh of flesh, coated with a layer of 2D skin. The knee flesh is composed of two solid mesh hemispheres that are not in contact with each other. This allows for the use of a pre-simulation model mover to position the legs without requiring a simulation. Both the femur and the tibia have cross-sectional load cells modeled near the knee, and the lower leg has passive 1D musculature. The joints of the lower extremity have varying implementations: the hips are defined by a spherical joint, the knees are defined by a revolute joint, and the ankles do not have a defined joint and are restricted to move by the skin connection to the lower leg flesh. More details on the development and validation of the simplified model can be found in Schwartz et al and Decker et al.

The modularly-incorporated components of the detailed model included all deformable bony structures form the sacrum to the foot, detailed soft tissue structures from the femur flesh to the foot flesh, and all explicitly meshed tendons and ligaments. The psoas and iliac pelvic muscles were incorporate as well but were not defined in contact with the simplified pelvic flesh due to intersections. The discrepancy in mesh size between the simplified pelvis flesh and the detailed femur flesh led to the incorporation of a 3D mesh transition that uses that uses the same material properties as the femur flesh. This M50-OS with the modular lower extremity will be referred to as the M50-OS+LEx.
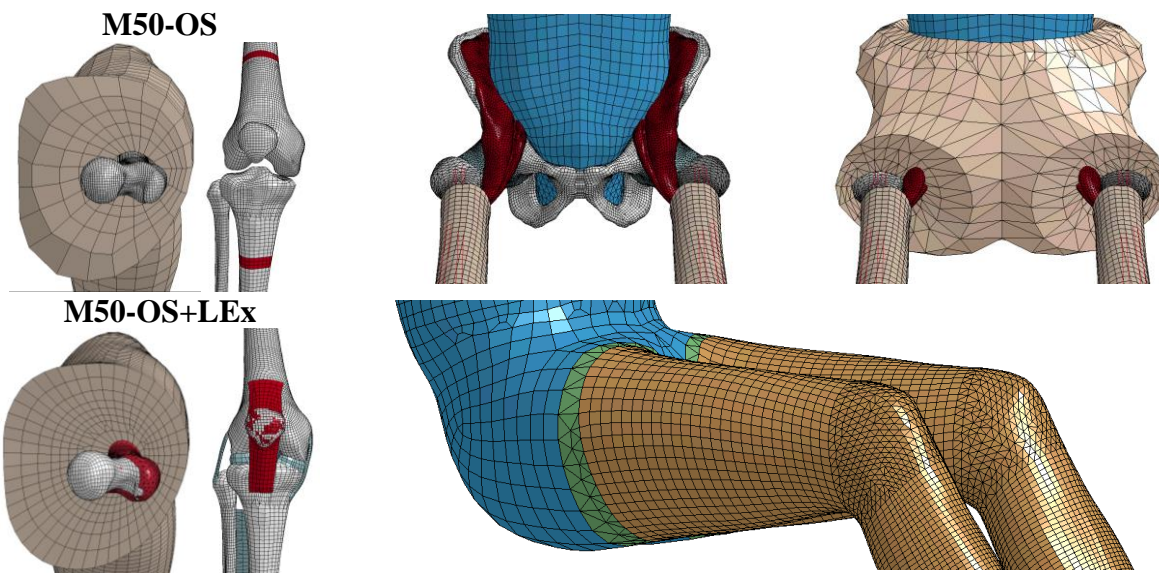

Figure 1: Comparison between the M50-OS and M50-OS+LEx (Left) and two views of the pelvic muscle intersections, as well as a view of the flesh transition shown in green (Right).

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Simulation Test Setups

The functionality of this modular approach in the lower extremity was evaluated through a localized knee bolster impact to the knees and a frontal sled simulation to replicate that of a common vehicular impact (Shaw, et al., 2009). Settling of both models was performed prior to the frontal sled, as well as a 100 ms settling period at the beginning of the pulse simulation. The knee bolster impact consisted of the model hitting a knee bolster at 4.9 m/s and the frontal sled pulse with an 11.8 m/s rearward acceleration to represent a frontal impact. A time history comparison for both simulations can be seen in Figure 3. The force response was compared between the M50-O and M50-OS+LEx for the two simulations. A total of 4 simulations were performed, with all simulations run on the Distributed Environment for Academic Computing (DEAC) high performance computational cluster at Wake Forest University using LS-DYNA R.7.1.2 (LSTC, Livermore, CA).
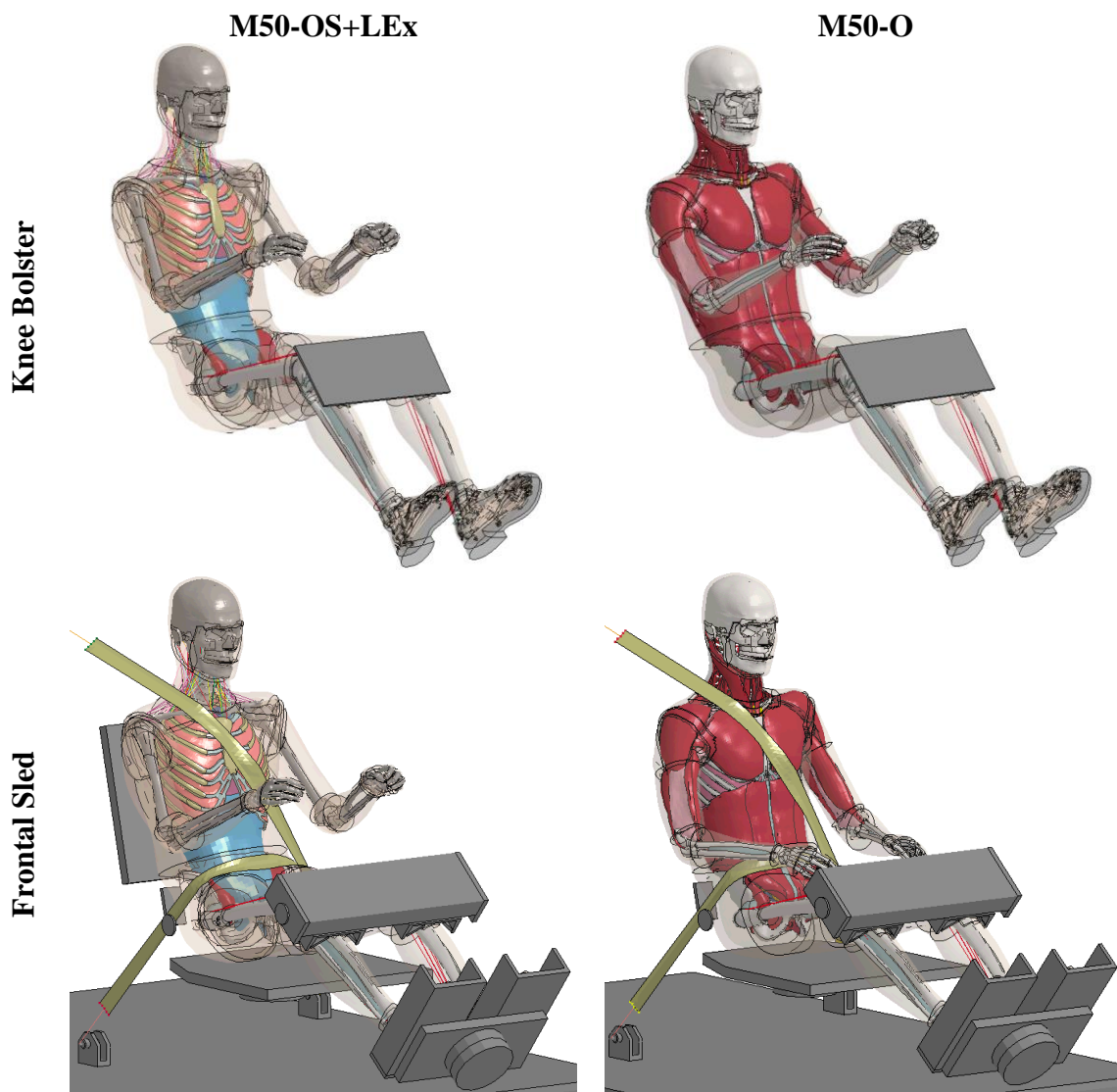

Figure 2: Visual of the M50-O and M50-OS+LEx with the knee bolster and frontal sled setups.

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RESULTS

Total force from the femur, upper tibia, lower fibula, and ilium were obtained and compared between the two models during both the knee bolster impact and the frontal sled simulation. These points of analysis were chosen to gather data from a variety of locations where lower extremity injuries are prevalent. The force vs time results for both simulations are shown in Figure 5.
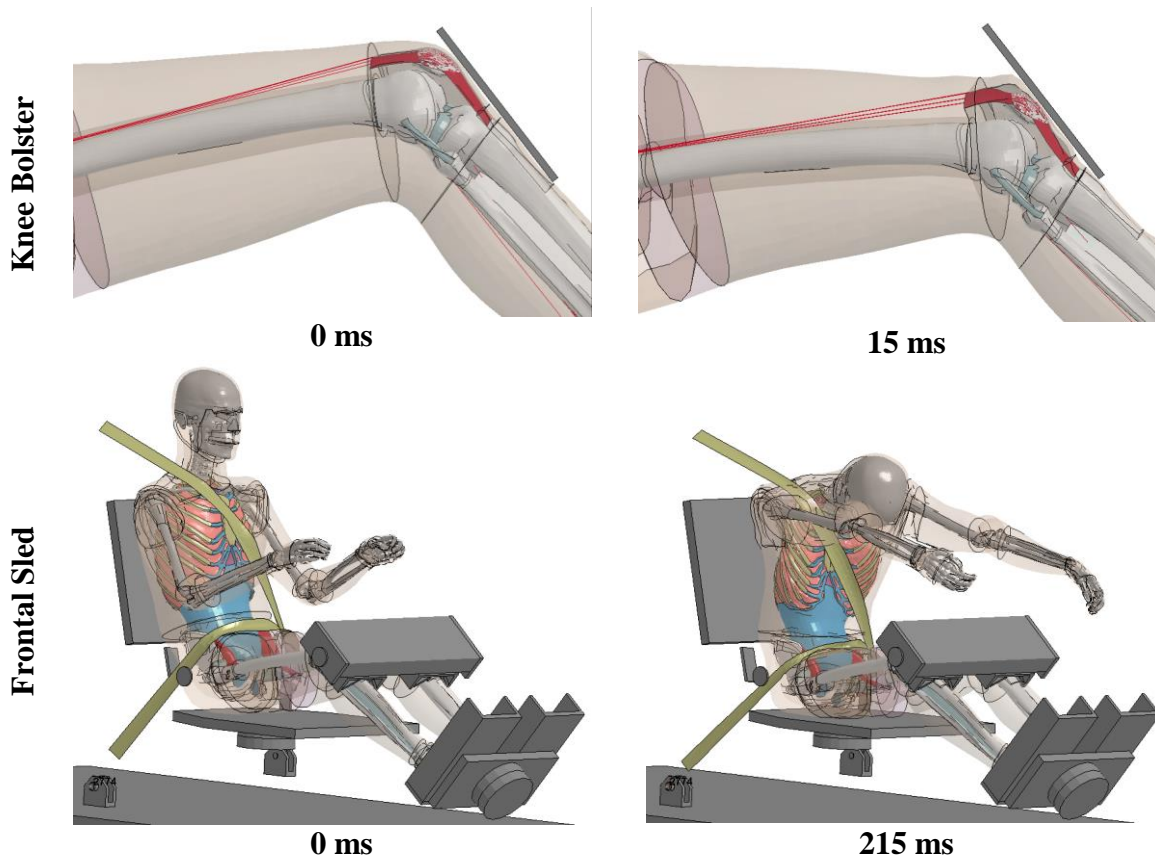

Figure 3: Time history of the M50-OS+LEx during the knee bolster impact and frontal sled simulations compared at the start and peak of the event.

The M50-OS+LEx showed a 71% reduction in run-time, with a normalized runtime of 2.3 min/ms compared to 7.9 min/ms of the M50-O (Figure 4). This comparison is of minutes per real time to calculate one millisecond of simulation.
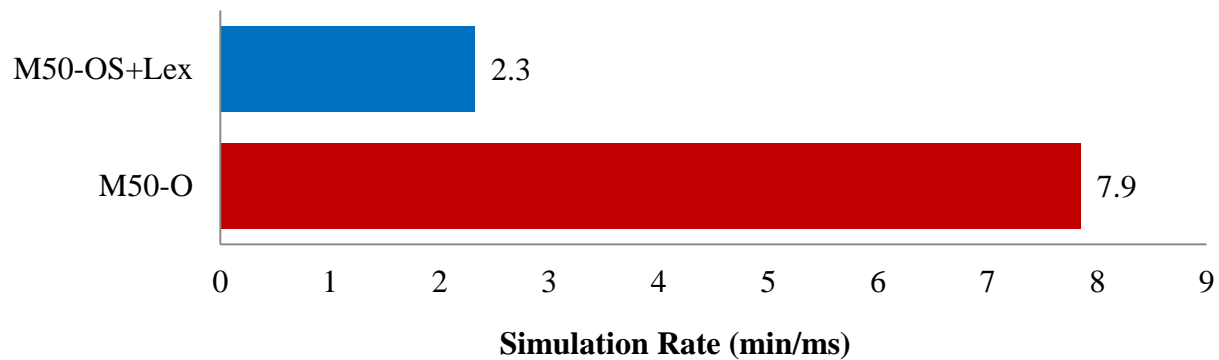
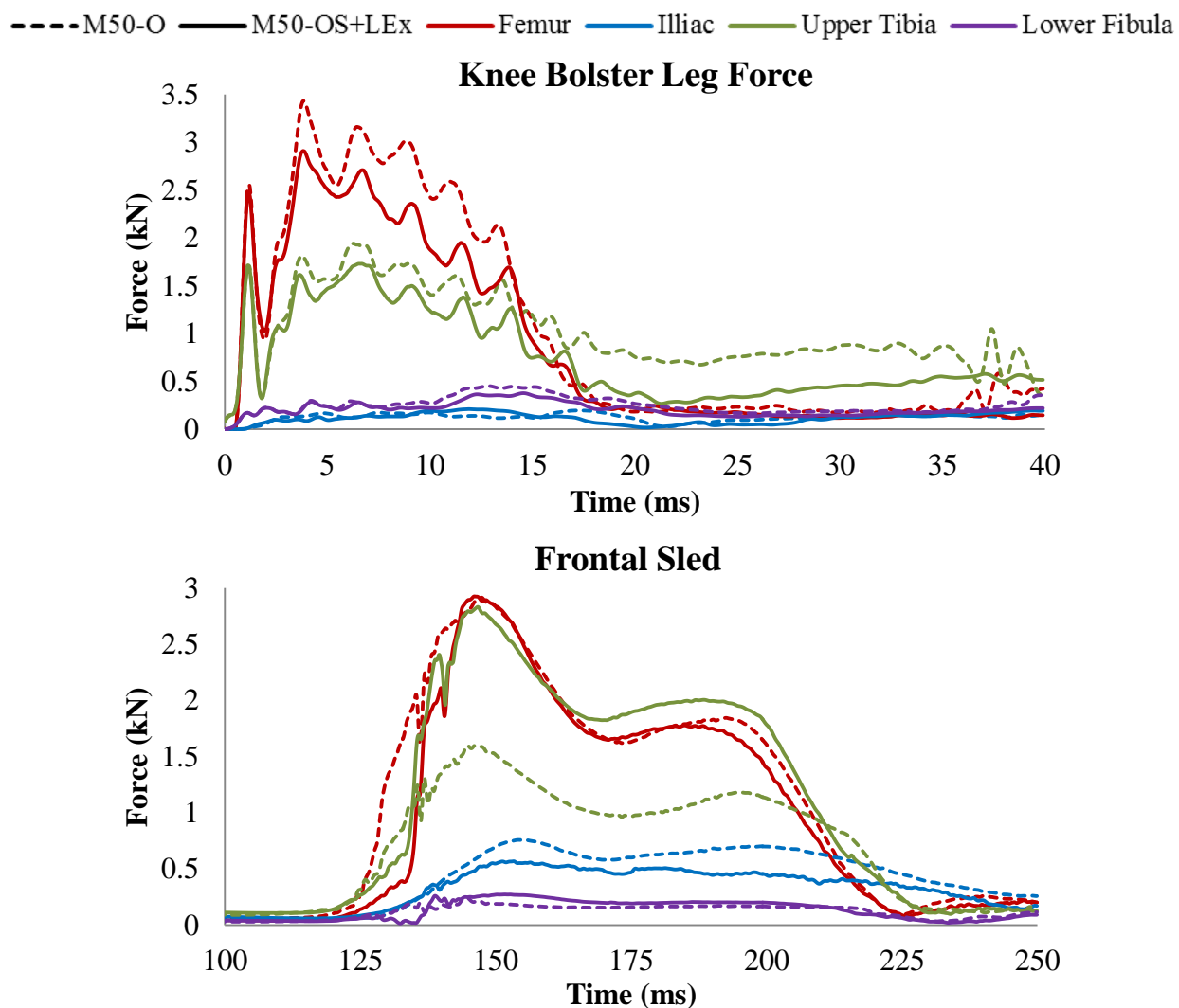Figure 4: Simulation rate comparison of the M50-O and M50-OS+LEx.

Figure 5: Force time history comparisons between the M50-O and M50-OS+LEx for the knee bolster and frontal sled simulations. An average of the left and right leg force was compared for the knee bolster impact, where only the right leg was compared for the frontal sled. The initial 100 ms settling period during the frontal sled is not shown.

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DISCUSSION

The incorporation of the detailed lower extremity into the M50-OS has shown the ability to yield reasonable force response in comparison to the M50-O, but with a runtime reduction of ~71%. As we demonstrated, the modular lower extremity model had a runtime ratio of 2.3 min/ms vs 7.9 min/ms of the detailed model. We achieved this time savings by modularly incorporating only the lower extremity into the simplified model, which allowed us to utilize the remaining time saving mesh/modeling in the M50-OS.

The knee bolster impact shows that the M50-OS+LEx underestimates force response consistently across the measured load cells by approximately 15%. The shape and time alignment of the response compares well with the M50-O. The underestimation of peak force may be due to the more simplified abdominal region of the M50-OS, which is still present in the M50-OS+LEx. This region has a single layer of coarse flesh, where the M50-O has abdominal and posterior muscles directly meshed to the pelvic bones.

The frontal sled force response shows the best correlation between the two models in the femur and fibula forces. The M50-OS+LEx accurately matched the femur force of the M50-O in terms of shape, magnitude, and time alignment. This may be a valuable attribute for this technique when modeling vehicle crashes, as femur fractures are among the most severe lower extremity injuries. The iliac force is noticeably lower in the M50-OS+LEx which is consistent with the knee bolster impact and can be attributed to the existing differences in the abdominal flesh mentioned previously. However, the tibia force is much higher in the M50-OS+LEx compared to the M50-O which is opposite to what was seen in the knee bolster impact. This may be due to differences in how the two models settled prior to the frontal sled simulation. Care was taken to settle the two models for the same amount of time, but the differences in flesh density of the buttocks may have led to the M50-OS+LEx knees settling further away from the knee bolster of the sled. This could lead to a higher severity impact of the tibia to the knee bolster which may explain this localized difference in this load cell.

CONCLUSIONS

Lower extremity injuries are common in vehicular crash incidents yet there is a limited selection of tools to study them. Computational tools such as HBMs can be used to address this need but detailed models can carry a high computational cost. The method introduced here, allows researchers to obtain similar force data at a much reduced computational cost. The implementation of the detailed lower extremity provided force time histories that tend to agree with the overall shape and magnitude of the data from the M50-O. Further investigation into the validation of this technique may include full vehicle buck simulations and optimization of included detailed components to ensure the best balance between time savings and performance.

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ACKNOWLEDGEMENTS

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