A multibody dynamics model of the MiL-Lx surrogate for under-body blast

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I. INTRODUCTION

Anti-vehicle (AV) mines and improvised explosive devices (IED) are playing an increasingly important part in modern warfare [1]. In recent conflicts of such nature, lower extremity injuries featured highly in mounted troops whose vehicles were attacked by a mine [1]. This injury mechanism is caused by the transfer of the blast load through the vehicle's floor and into the subject's lower limbs [2]. All experiments to date with post mortem human subjects (PMHS) have been conducted with the lower extremity in a nominal posture, with the foot flat, the tibia vertical and the knee joint at 90° flexion (the 90-90-90 posture). Mounted troops do not always sit in that configuration, however, and the injury risk for a posture other than the nominal in unknown. Therefore, the aim of this project is to develop a simple, fast-running, multibody model of the MiL-Lx lower-extremity surrogate to investigate the response to simulated under-body blast for different postures, as it is the most biofidelic surrogate to date for under-body blast [3].

II. METHODS

A multibody model of the MiL-Lx was developed using MSC Adams (v2013, MSC.Software, Santa Ana, USA). The components were modelled as cylinders, with the length and mass matching the corresponding ones in the physical surrogate (Fig. 1). Two point masses of 40 kg and 2.38 kg were added at the hip and knee joints to account for half of the upper body weight and the metallic knee joint of the MiL-Lx, respectively. All of the components, with the exception of the compliant element, were modelled as rigid bodies. The compliant element was given stiffness and coefficients to account for its behaviour in the physical model (Table 1). Hip and knee joints were modelled as hinge joints, with contact defined between the loading plate and the foot and between the femur and the seat. The hip joint was also given a translational freedom in the sagittal plane. The displacement of the plate was controlled to simulate underbody blast and match the experimental design presented in [4].

A sensitivity study was performed to identify the parameters to which the model was most sensitive. The step size, material properties, compliant component parameters, initial and boundary conditions were varied by ±20% and ±40% using the one-at-a-time approach.

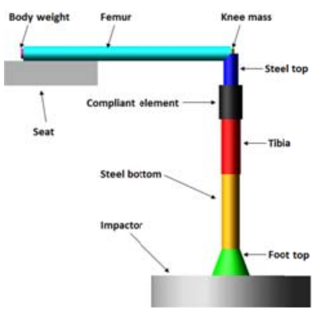

Fig. 1. Geometry of the multi-body model representing the MiL-Lx surrogate limb.

(a) A (b) B

Fig. 2. Postures of the lower limb obtained by varying: (a) A – the knee angle and the ankle angle outwards; (b) B – the knee angle and the loading plate angle;

Component	Mass (kg)	Stiffness coefficient (kN/m)	Damping coefficient (Ns/m)
Steel top	0.350	Rigid	Rigid
Compliant element	0.120	220-250	10
Tibia	0.367	Rigid	Rigid
Steel bottom	1.249	Rigid	Rigid
Foot top	1.111	10000	10
Knee mass	2.316	Rigid	Rigid

Table 1. Mass, stiffness and damping coefficients (where applicable) of MiL-Lx model components.

The model was then used to investigate the response to a simulated under-body blast for four different postures, obtained by varying the hip angle, knee angle, ankle angle and loading plate angle in increments of

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15° (Fig. 2). The load at the upper tibia and the compression of the compliant element were compared to the experiment, which is shown in Fig. 3 in [4].

III. INITIAL FINDINGS

Stiffness and damping coefficients of the compliant element were calibrated in order to match the experimental data. For stiffness values between 220 and 250 kN/m and a damping coefficient between 10 and 100 Ns/m, the model gives a peak force and compliant element compression which match the experimental values, (Fig. 3 (a), (b)). The sensitivity study showed that the stiffness of the compliant element had the greatest effect on the output impulse. The average model runtime is 10 s on a typical home computer.

(a) (b)

Fig. 3. Comparison between experimental (black line) and simulation results for different stiffness values of the compliant element for (a) axial force, and (b) compliant element compression histories.

(a) (b)

Fig. 4. Peak tibia force for different knee joint angles of (a) posture A, and (b) posture B.

IV. DISCUSSION

This fast-running model is able to predict the load through the MiL-Lx in underbody blast, and therefore give an estimate of injury risk to the lower extremity for various loading conditions. Important limitations of this model are that it represents the extremity with simple geometries and surrogates all the compliance of the leg at one place, similar to the physical MiL-Lx. In addition, the model exhibits a fairly linear response and is not able to capture aspects of the more complex, nonlinear compliance response of the physical surrogate.

For postures A and B, the peak tibia forces decreased with an increase in the knee angle, as shown in Fig. 4 (a), (b). Comparing these two postures, the peak tibia force for the second posture was consistently higher than for the first posture. Similar results were obtained using a MADYMO model of the Hybrid III leg for this set of postures [6]. The model allows for an inexpensive sweep of the effect of different postures on injury risk, it can be used to assess mitigation systems at the foot–floor interface, and can be implemented in larger systems (e.g. vehicle). A criterion to assess out-of-position vertical loading scenarios might be the Tibia Index [5], as it takes into account the bending moment of the tibia.

In addition, the model can give an estimate of the load going up the skeleton, allowing for boundary conditions for FE models of the pelvis and the spine. The model can be extended to include the proximal skeleton. Care should be taken when interpreting these results with regard to injury risk, as the physical MiL-Lx is not validated for out-of-position configurations.

V. REFERENCES

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