How are loads in the pedestrian struck side leg affected by boundary conditions in Human Body Model simulations?

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I. INTRODUCTION

Several developments of biofidelic subsystem impactors for pedestrian testing have been presented in the past, with the purpose of minimising the lower extremity injury severity in pedestrian accidents. The flexible pedestrian legform impactor (FlexPLI) shows higher biofidelity compared to the European Enhanced Vehicle-Safety Committee (EEVC) pedestrian legform impactor [1]. Due to the missing upper body mass the FlexPLI cannot adequately simulate the bending load generated on a lower limb of pedestrians for specific vehicle geometries [2]. The advanced pedestrian legform impactor (aPLI) with a supplementary upper body mass (SUBP) was developed to address this issue [2]. The aPLI was developed (and thus correlates well) with the JSAE Human Body Model (HBM). It would appear sensible to re-evaluate the approach by employing another HBM and another generic test buck and by applying varying boundary conditions. The current study aims to answer the following questions: (1) What is the effect of varying boundary conditions (like impact height, leg length and mass, contact with non-struck side leg, Isolated leg with simplified upper body mass, etc.) on the pedestrian leg loads? (2) Is the SUBP developed by [3] transferable to another HBM?

II. METHODS

Simulations were performed in Ls-Dyna with the 50th percentile male pedestrian model of the Total Human Models for Safety (THUMS) version 4.02 without fracture. THUMS was propelled with 40 km/h against the stationary and fully constrained Generic Vehicle Testing Rig (GVTR) representative of a contemporary European sedan, [4-5]. All numerical simulations were performed up to 80 ms.

In the THUMS 20 cross-sectional (crs) force transducers were introduced along the upper leg (femur) and lower leg (tibia and fibula). Each crs was assigned an individual coordinate system, shown in Figure 1 that was re-adjusted at each time-step [6]. The lower leg moment output in this study considers tibia moments only and disregards those in the fibula. The crs selected as per [7], remained unchanged throughout the study, i.e., remained the same when impact locations were varied.

Consistent with the approach taken in [3], the struck side (right) leg of the THUMS was isolated and equipped with an upper body mass (Figure 1). The technical specifications of SUBP were extracted from [3]. The SUBP consists of the rigid V-shaped pelvic bone to which a compliant flesh-like part was adhered. The THUMS isolated leg + SUBP have a total mass of 22.8 kg. The SUBP mass, inertia and centre of gravity relative to the hip joint are as per [3]. The hip joint (constrained joint revolute allowing rotation, only 103° in medial direction to 70° in lateral around local x-axis pointing anteriorly) connecting SUBP with the isolated leg was located at the centre node of femur head.

The THUMS was positioned in the gait posture according to Euro NCAP TB024 [8, 9] in a pre-simulation with the right leg being on the struck-side and trailing (Figure 2). The same leg posture was used in the isolated leg simulations. Static and dynamic friction between GVTR and THUMS was set to 0.3 [8-9].

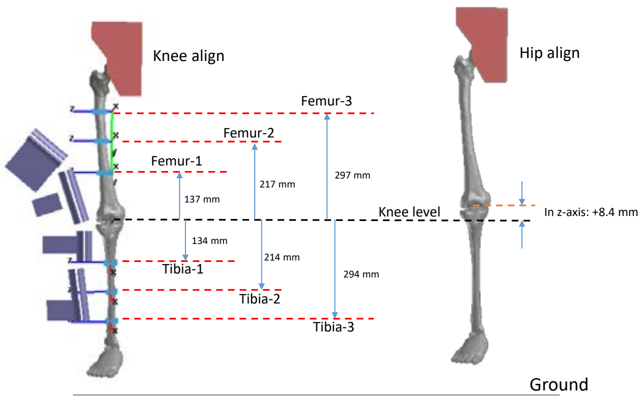

Figure 1. The isolated THUMS leg with SUBP and crs.

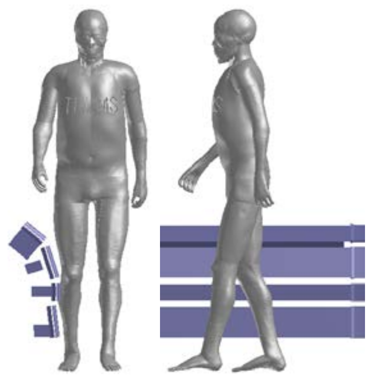

Figure 2. The full THUMS in gait posture.

In this study, the following alignments were investigated: (1) knee aligned to [3] i.e. the femoral epicondyle was 520 mm above the ground, (2) the hip aligned to [3] i.e. the mid-point of femur head (same as hip joint centre) was placed at 958 mm above the ground. Simulations were performed with active and disabled contact between the two legs, as described in [3]. Additionally, simulations were performed with an isolated THUMS leg with a SUBP. To study the effect of anthropometry, the THUMS leg length was scaled from 901 to 933 mm to match aPLI and the mass was scaled, i.e. additional nodal masses were included to increase the weight by 1.1 kg to match the 12.1 kg of aPLI. Furthermore, simulations with the isolated leg with SUBP without feet were performed.

III. INITIAL FINDINGS

Lower extremity bending loads (x-moment) for the full THUMS are shown in Fig. 3. In the femur sections 1 and 2 (distal), the moments were 9-12% lower (compared to the THUMS with contact) for the THUMS without (indicated as w/o) leg contact, whereas they were 22-29% higher in femur-3 (proximal). Knee and hip alignment had negligible effects (3-9% increase in hip aligned) on femur moments. The tibia peak bending moments showed minor variations in knee and hip aligned cases and w/o leg contact. The effect of disabling the contact between the legs was greater than the effect of the different impact heights.

Fig. 4 shows the variations of isolated THUMS leg in comparison to the simulations of full THUMS w/o leg contact (for the knee alignment). In contrast to full THUMS w/o leg contact, the isolated leg showed higher moments for different boundary conditions, except for Tibia-3. The largest effects were observed when the leg length was scaled.

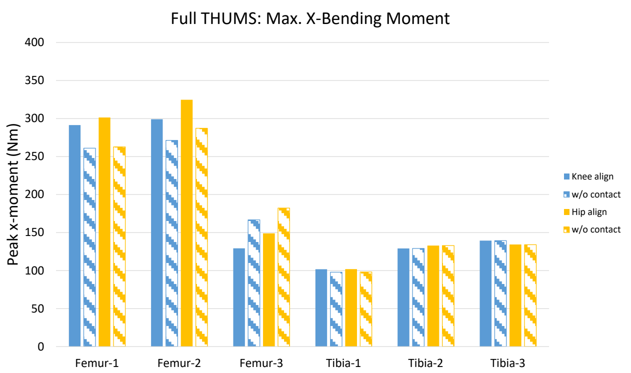

Fig. 3. Peak bending moments for the full THUMS with and w/o contact and two different impact heights

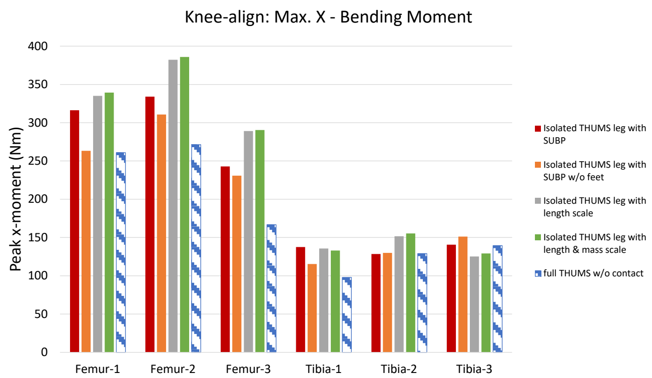

Fig. 4. Peak bending moments for various boundary conditions (knee alignment).

IV. DISCUSSION

The current study shows that the sensitivity of lower leg loads on varying boundary conditions was relatively small. The measurement in the uppermost femur cross-sections (crs-3) seemed to be less robust and might need to be considered with care when used as an assessment criterion. The THUMS isolated leg with SUBP showed similar peak bending moments to the full THUMS for the tibia, but returned higher moments in the femur, particularly in the two uppermost crs. These crs were also highly affected by the length of the femur, which indicated that the results might be influenced by the detailed anthropometry of the applied HBM. In full THUMS, the absence of a contact between the legs underestimated the moments for femur crs 1 and 2, while it overestimated moments in the uppermost femur, which should be considered when defining tolerances for assessment procedures using aPLI.

V. ACKNOWLEDGEMENT

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VI. REFERENCES

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