Development of a Multibody Human Leg Model based on Beam Approximation

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I. INTRODUCTION

Traditionally in multibody human models, a leg has been modelled with multiple bodies connected together with spherical joints and non-linear torsional springs (M1 – joint-based model). The existing leg models were validated to match the moment vs. deflection response in three-point bending mode under dynamic [1] or quasi-static conditions [2]. Using limited responses, such as the moment vs deflection, could over-define the system and lead to non-physical, highly non-linear joint restraint characteristics. A literature review revealed a drop in the area moment of inertia from the proximal to the distal end of the tibia, and the joint stiffness distribution between the proximal and distal ends did not match this behaviour in the existing multibody leg models. Such models could show non-physical behaviour when used under conditions for which they were not specifically validated. To overcome the issue of over-defining the system, a beam-based approximation of the long bones of the tibia has been presented in the current study, which would constrain the design space available to model the long bone by considering the area moment of inertia distribution along the bone.

II. METHODS

The leg model (M2) developed in the current study (Fig. 1(a)) was based on the multibody leg model (M1) developed by Kerrigan *et al*. [1] in MADYMO v7.6 [4]. The geometry and dimensions were based on the 50th percentile UVA-GM FE human model developed by Untaroiu *et al*. [3] (Fig. 1(b)). The tibia was modelled using four beam elements (five nodes) joined along the axis of the tibia (Fig. 1(a)) using the BODY.FLEXIBLE_BEAM in MADYMO [4]. The exterior flesh of the leg was divided into five parts so that the location of center of gravity (CG) of each body (inclusive of flesh and bone) approximately matched with that of the FE model. Nodes 2, 3 and 4 divided the tibia into four equal segments in length. Each node was rigidly supported by a corresponding body. An in-house programme was developed to calculate the geometric properties, including the area (A) and area moment of inertia (AMOI) of the tibia about the anterior-posterior (AP), mediolateral (ML) and superior-inferior (SI) axes, at pre-defined cross-sections through the length (at sections 15%, 25%, 50%, 75% and 85% of tibial length) from the CT scan data of 11 specimens tested in three-point bending by *Kerrigan et al*. [5]. The geometric properties from all of the specimens at the chosen normalised lengths were averaged to represent the mean geometric properties of the specimens and were incorporated into the model. The Young's modulus of cortical bone and the flesh contact characteristics from the five body regions were then optimised to validate the model against the PMHS responses under the three-point dynamic leg bending conditions (proximal, distal and mid loading at 1.5 m/s) [5]. The test and the model responses were compared against each other for the following time histories during the three-point bending: impactor contact force (I. F.); proximal reaction force (P. F.); distal reaction force (D. F.); proximal cup angle (P. A.); and distal cup angle (D.A.) (Fig. 1(c)). The sum of the normalised (using the standard deviation of mean PMHS response) root mean square error between the test and model responses was used as the cost function (C), and minimised during model validation using a global optimisation technique provided by Matlab (Mathworks, Natick, MA).

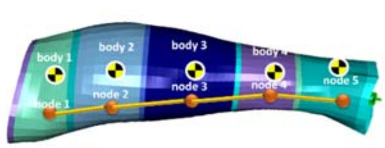

Fig. 1(a). Madymo leg model.

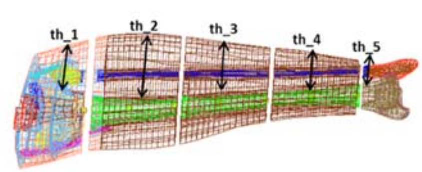

Fig. 1(b). UVA GM leg model.

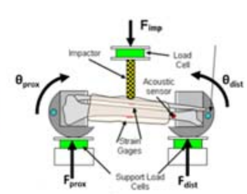

Fig. 1(c). Three-point bending set-up.

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Flesh was modelled using a polynomial-based, nonlinear, stress-strain curve (Eq. 1):

$$\sigma = \begin{cases} a_l\lambda, \lambda < \varepsilon_l \\ a\lambda + b(\lambda-\varepsilon_l)^3 + c(\lambda-\varepsilon_l)^5, \lambda > \varepsilon_l \end{cases} \quad (1)$$

where, λ represents the strain (penetration/thickness of facet) in flesh and σ represents the corresponding stress (contact force/contact area) developed in the material. The ε_l represents the strain at which flesh bottoms out and shows non-linear behaviour. The flesh belonging to different leg bodies were assigned different flesh thickness values. The flesh thickness for each of the five bodies in the model was taken from the General Motors (GM)/University of Virginia (UVA) 50th percentile male FE model (Fig. 1(b)), with distances measured from the tibia to the lateral flesh (th_1=46.9 mm, th_2=71.0 mm, th_3=70.1 mm, th_4=45.5 mm, th_5=35.8 mm). The fibula was not modelled, owing to its low bending stiffness compared to that of tibia. However, it plays a role in the bottoming out of flesh while the leg is being loaded by an impactor. Therefore, the flesh thickness (th_j) in each of the five body regions was modified following Eq. (2):

$$th_j = th\_j * scf_i - b_f, where \text{ (j=1\&2, i=1: Prox); (j=3, i=2: Mid) \& (j=4\&5, i=3: Dist),} \quad (2)$$

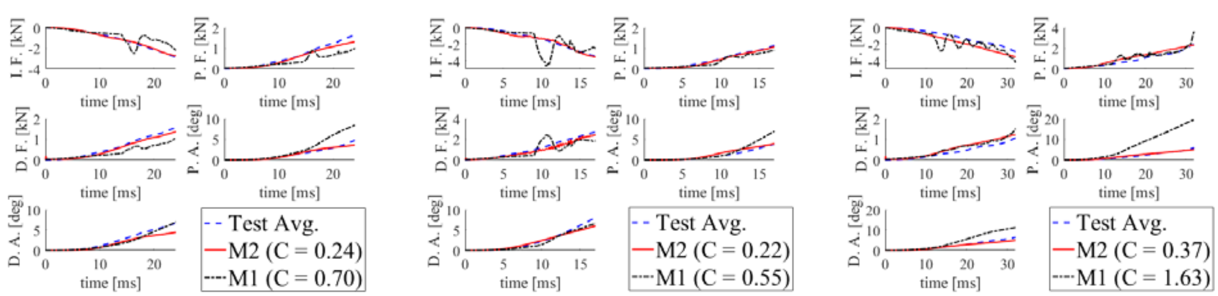
where scf_i represents the scale factor for each region, b_f represents the reduction in thickness to account for fibula-flesh contact, and *th_j* is the tibia-flesh thickness measured from the UVA-GM FE model. A stress proportional damping proposed by Anderson *et al*. [6] was employed to model the flesh damping, and the damping constant (c_f) was one of the other parameters optimised to match the test response. The range of each parameter used in optimisation is summarised along with their optimal values (Table I).

III. INITIAL FINDINGS

The optimised leg model (M2) (Table I) matched well with the test data under all three test modes (Fig. 2) and showed better biofidelity (lower C values) when compared with the joint-based model (M1).

TABLE I
OPTIMISED PARAMETERS OF THE BEAM-BASED LEG MODEL

Parameter	Value (Range)	Parameter	Value (Range)
Young's modulus	15.0 GPa (5–20)	scf_1	1.59 (0.7–1.6)
a	2.74e6 Pa (7e4–4.4e6)	scf_2	1.11 (0.7–1.6)
b	2.83e7 Pa (2e5–6e7)	scf_3	0.70 (0.7–1.6)
c	3.59e7 Pa (7.5e6–2.25e8)	b_f	7.58 mm (0–30)
c_f	0.028 (0–0.05)	ε_l	0.144 (0.05–0.25)



(a) Mid loading. (b) Distal loading. (c) Proximal loading.

Fig. 2. Comparison of three-point bending responses between PMHS and Madymo models (C is the cost).

IV. DISCUSSION

Model M1 was validated to average bending moment, and therefore did not show good correlation to other responses (impactor force and cup rotation angles). M1 model may be improved by considering other responses during model validation. Nevertheless, the M2 (or beam-based) model has an inherent advantage over the M1 model because it uses a smaller number of parameters due to the consideration of biomechanical information (A and AMOI). Model M2 showed better match in proximal and distal rotational angles (Fig. 2) as it had the appropriate stiffness (Young's modulus) for the long bone. The average Young's modulus of the cortical bone obtained through optimisation of model M2 falls within the experimentally observed range of 18.6±3.5 GPa [7]. Although only tibial geometric properties were used in building the model (M2), it is validated against the PMHS test data, which included fibular response and is a limitation of the current approach in modelling the leg using multibody models.

V. REFERENCES

1 [1] Kerrigan, J. *et al.*, TIP, 2009.
2 [2] Madymo Human models manual, v7.6, 2015.
3 [3] Untaroiu, C. *et al.*, *Stapp Car Crash J.*, 2005.
4 [4] MADYMO Reference Manual, release 7.6, 2015.
5 [5] Kerrigan, J. *et al.*, *I J Crash*, 2004.
6 [6] Anderson, R. W. G., *Non Linear Dynamics*, 2008.
7 [7] Rho, J. Y. *et al.*, JBME, 1993.