The Role of Non-linear Stiffness in Modelling High-Rate Axial Loading of the Spine

Lucas Low, Robert Salzar, Nicolas Newell, Spyros Masouros

I. INTRODUCTION

Injury to the spine is a common result of high-rate axial loading, such as underbody blast events (UBB) to seated vehicle occupants, with lumbar spine fractures present in 35% of troops injured by UBB [1]. The blast pulse has been reported to reach the pelvis and lumbar spine within 30 ms, with a peak upwards acceleration at the seat exceeding 100 g [2]. Insight into the load transfer and kinematics of the spine during high-rate loading is key to developing targeted and effective injury-mitigating strategies.

Human body models are common in the automotive industry for predicting injury and designing injury mitigation technologies. However, these models have not been validated for high-rate axial loading as seen in UBB. Furthermore, most are finite element models, which are complex to modify and computationally intensive. A simple, quick-running multibody model of the spine that is able to simulate UBB could offer a practical aid to mitigation efforts in UBB. The vision is for the model to be easy to modify and run almost 'real time', allowing the assessment of various injury-mitigation measures, such as seat cushions, and the effect of posture, gender and patient-specific geometries on spinal injury.

This study aims to compare the use of linear or non-linear stiffness to model the intervertebral disc in a multibody model of the spine in UBB.

II. METHODS

A multibody model (Fig. 1) has been developed that represents the head and torso (represented by the skull and spinal column). Each vertebral body (VB) is modelled as a rigid body and represents both the VB and surrounding soft tissue. The model is generated from a CT scan using the method described in [3]. The CT scan is derived from a full-body cadaveric specimen used in Test 1.6 of [2]. Test 1.6 is a lab-simulated cadaveric UBB event, which will be simulated for this study using the model and then the results compared.

Adjacent VBs are connected by a 6 degree-of-freedom spring-damper system at the centre of rotation on top of the bottom VB [4]. The stiffness and damping properties represent the combined effects of the intervertebral discs, ligaments and soft tissue in the torso. The initial stiffness and damping values are acquired from values reported in experimental studies of functional spinal units (FSUs) [5-11]. Stiffness is input as a force-displacement or moment-rotation relationship, either linear or by fitting force-displacement and moment-rotation plots reported in the literature using a Mooney-Rivlin formulation (Fig. 2). Damping is implemented as a linear equation due to lack of force- and moment-velocity curves.

The model input is the acceleration recorded at the sacrum during Test 1.6, as described in [2]. This model was run using both linear and non-linear models of FSU stiffness. Accelerations at the skull and T1 VB, which were recorded in the experiment, are compared to the model response at the skull and T1, and the CORA ISO 18571 score is calculated.

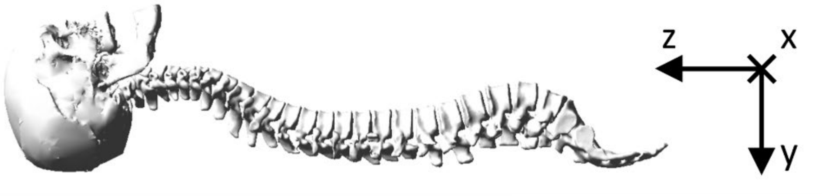

Fig. 1. The multibody model is composed of rigid bodies of the skull, vertebral bodies and sacrum. Adjacent vertebral bodies are connected by a 6 degree-of-freedom spring-damper element. Acceleration is input at the sacrum. The 3D representations of the skull and spine are for illustrative purposes and are not involved in the simulation. The coordinate system for the model is shown on the right.

L. Low (e-mail: lucas.low15@imperial.ac.uk) is a PhD student in the Centre for Blast Injury Studies in the Department of Bioengineering, S. Masouros is Reader in the Department of Bioengineering, N. Newell is Research Fellow in the Department of Mechanical Engineering, all at Imperial College London, UK. R. Salzar is Associate Research Professor at the University of Virginia, USA..

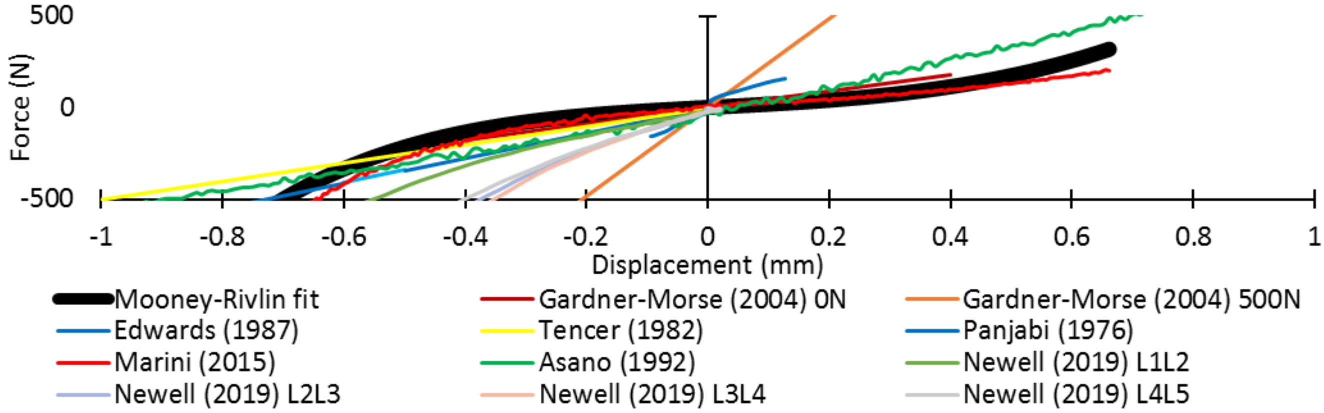

Fig. 2. The linear and non-linear force-displacement curves for axial translation of functional spinal units (caudal-to-cranial; negative displacement indicates compression) reported in experimental studies [5-11]. Gardner-Morse *et al.* [5] report stiffnesses at 0 N and 500 N axial compressive preload. The Newell *et al.* [11] tests were conducted at a strain rate of 1/s.

III. INITIAL FINDINGS

The acceleration responses of the skull and T1 VB from the model and experiment are shown in Fig. 3. The experiment shows a peak of 128 g at 13.9 ms at T1. The model with non-linear stiffness shows a peak of 125 g at 13.4 ms, with a CORA score of 0.543 at T1 and a CORA score of 0.555 for the skull. The model with linear stiffness shows a peak acceleration of 126 g at 17.3 ms, with a CORA score of 0.447 at T1 and a CORA score of 0.114 for the skull. In both the model with non-linear stiffness and the experiment, the pulse at the skull and T1 is smoother than that of the sacrum, with a single peak at the skull and T1. The model with linear stiffness shows multiple peaks.

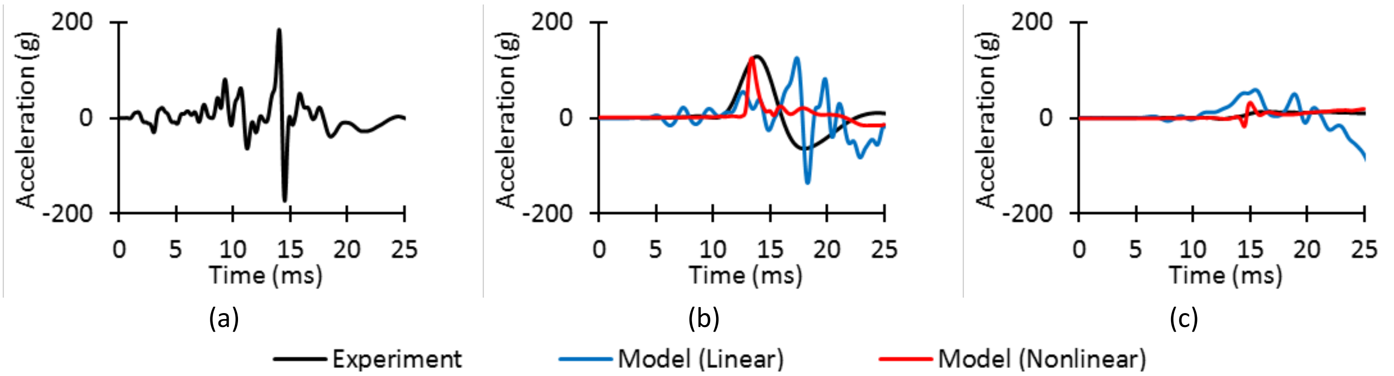

Fig. 3. Axial acceleration from the experiment and model with linear or non-linear stiffness. (a) Sacrum (experimental data are the input to the model), (b) T1, (c) Skull.

IV. DISCUSSION

A multibody model of the spine was developed and used to simulate a lab-simulated UBB test; use of either linear or non-linear stiffness for the FSUs was explored. Both linear and non-linear stiffness options captured the magnitude and timing of the peak acceleration at the skull and T1 measured experimentally. However, only the model with non-linear stiffness captured the shape of the pulse adequately. This indicates that non-linear stiffness is required for the model to simulate high-rate axial loading accurately. The next immediate step is to explore the effect on model response of the parameters of the non-linear stiffness across cervical, thoracic and lumbar parts of the spine.

V. REFERENCES

[1] Vasquez, K. B., *et al., Military Medicine*, 2018.
[2] Bailey, A. M., *et al., Ann Biomed Eng*, 2015.
[3] Low, L., *et al., IRCOBI,* 2020.
[4] Pearcy, M., *et al., Spine*, 1988.
[5] Gardner-Morse, M. G., *et al., J Biomech*, 2004.
[6] Edwards, W. T., *et al., J Biomech,* 1987.
[7] Tencer, A. F., *et al., J Biomech,* 1982.
[8] Panjabi, M. M., *et al., J Biomech*, 1976.
[9] Marini, G., *et al., J Biomech*, 2015.
[10] Asano, S., *et al., Spine*, 1991.
[11] Newell, N., *et al., Spine*, 2019.