Passive Neck Muscle Implementation in Finite Element Models Influences Upper Cervical Spine Kinematics in Low-Speed Rear Impacts

Matheus A. Correia, Prasannaah Hadagali, Stewart D. McLachlin, Duane S. Cronin

I. INTRODUCTION

Finite element (FE) human body models (HBMs) can provide insight into the response of vehicle occupants in impact scenarios, but biofidelity of tissue-level response is required to predict injury risk [1]. Head and neck FE models (NMs), for example, have been developed with the aim of predicting tissue-level injury [2]. Although many NMs provide representative head kinematics, even when using simplified tissue representations [2-3], studies have shown that intervertebral (IV) kinematics are important for tissue response prediction [4]. In addition, IV kinematics are affected by the muscle tissue, especially in low-severity impacts [3]. For low-impact severities (≤4 g), a study by Stemper [5] reported the IV kinematics for rear-impact sled tests with cadaveric head and neck specimens, while Sato [6] presented human volunteer data for low rear-impact severities (≤4 g). Several FE NMs use one-dimensional (1D) Hill-type elements to represent the line-of-action between the origin and insertion points of the muscles [3][7-9]. To enable the modelling of seat-belt interaction, these NMs often represent the neck tissue volume as a homogeneous hyperelastic adipose tissue, but do not discretise the individual muscle volumes. Other models have proposed a hybrid implementation using hyper-viscoelastic 3D elements to represent the passive response of the muscle volume [10], combined with discretised 1D Hill elements representing the active response [11]. Although there is a recent trend towards simplified implementations for improved computation efficiency, there has been no comparison in the literature between the commonly used 1D elements implementation and hybrid elements implementation of the neck muscles with respect to localised responses, such as IV kinematics. Understanding the differences in these implementations is essential for future analysis of tissue-level injury prediction. The current study aimed to compare the 1D and hybrid muscle approaches using the Global Human Body Models Consortium (GHBMC) FE NM response and existing experimental data for IV kinematics in low-speed rear impacts. The NM allowed the 3D elements to be removed while maintaining the 1D elements, enabling a direct comparison between the two muscle implementations.

II. METHODS

The head and neck were extracted from the GHBMC full-body model (M50-O v5-1). This Hybrid neck model (H_{NM}) included the osteoligamentous spine (T1 to C1), skin, adipose tissue and 3D passive muscles [11] (Fig. 1a). For a direct comparison, only passive muscle properties were considered in the present study, without any muscle activation. A simplified neck model (S_{NM}) (Fig. 1b) was then created by removing the 3D muscles and representing the passive response with the parallel elastic element in the 1D Hill-type elements, similar to existing neck models in the literature [8], with the muscle line-of-action constrained by attachments to the vertebra. A limitation of the passive muscle representations in many current 1D implementations [3][7-9] is using only one stress-strain curve without consideration of the effect of tissue deformation rate. Therefore, two passive muscle properties were simulated with the S_{NM} model to capture a range of experimental muscle tissue properties. Quasi-static passive muscle tissue properties [8], which are used in many existing neck models, were implemented for a low-

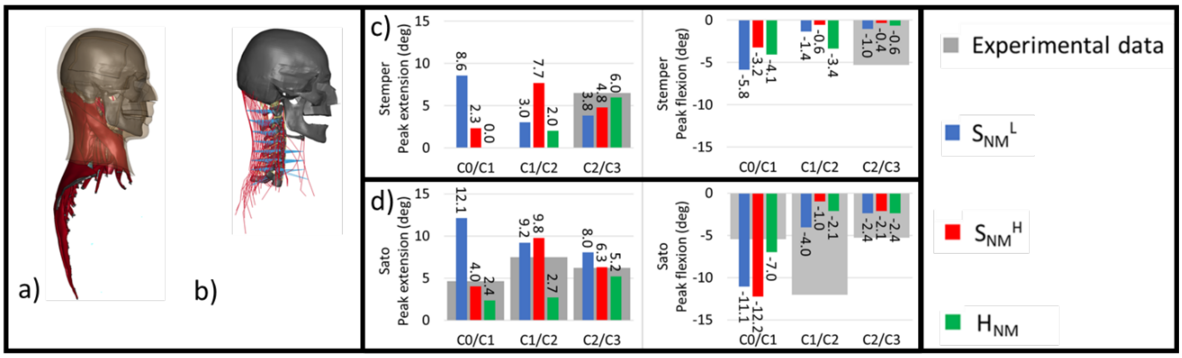

Fig. 1. a) H_{NM} and b) S_{NM} models. The peak extension (positive) and flexion (negative) are shown for the boundary conditions from c) cadaveric specimens [5] and d) volunteers [6]. The experimental peaks represented one SD and were omitted when not available in the literature.

M. A. Correia is a Ph.D. student (+1 519-888-4567 x32682, macorrei@uwaterloo.ca), P. Hadagali is a Ph.D. student, S. D. McLachlin is a Professor and D. S. Cronin is a Professor in the Mechanical & Mechatronics Department at the University of Waterloo in Canada

rate version of the NM (S_{NM}^{L}). The passive properties representing a higher strain rate (25 s^{-1}) were implemented as an upper bound (S_{NM}^{H}) [10]. The first thoracic vertebra (T1) kinematics from rear-impact experiments (Stemper [5], Sato [6]) were applied as boundary conditions to the T1 vertebra in the three models (S_{NM}^{L}, S_{NM}^{H} and H_{NM}). The simulation termination times were limited by the duration of the experiments, resulting in 110 ms for the cadaveric boundary conditions and 200 ms for the boundary conditions of the volunteers. The IV peak rotations in the model were defined as the maximum rotation of the vertebra in relation to the inferior vertebra in extension and flexion. The percent difference was calculated using the maximum rotations from each model.

III. INITIAL FINDINGS

The IV kinematics were similar for the models (S_{NM}^{L}, S_{NM}^{H} and H_{NM}), except for the C0/C1 joint, which presented a maximum difference of 9.7° (80%) in peak vertebral extension and 5.2° (43%) in peak vertebral flexion among the models (Fig. 1). The cadaveric case presented a maximum difference of 8.6° (100%) in peak vertebral extension and 2.6° (45%) in peak vertebral flexion for the C0/C1 joint. The absolute differences were smaller for the cadaveric case, but these differences increased with time. In addition, only the H_{NM} presented flexion for all the 110 ms at the C0/C1 level for the cadaveric simulations, better approximating the S-shape trend identified in the experiments [5] (Fig. 2). Comparing the model to the experimental corridors from the study with volunteers, the H_{NM} was, at most, 1.5° (28%) outside the corridor for the C0/C1. In contrast, this value was up to 7.5° (162%) and up to 6.8° (124%) for the S_{NM}^{L} and the S_{NM}^{H}.

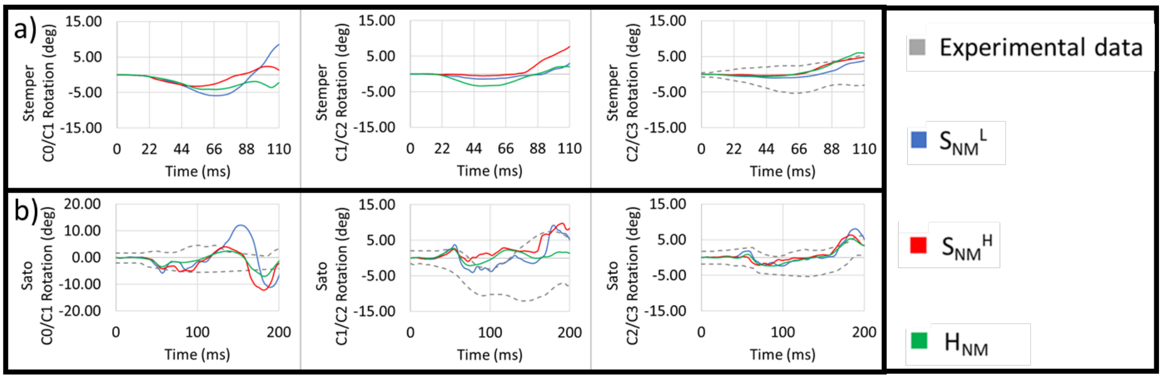

Fig. 2. IV kinematics for the impact simulations for the boundary conditions from a) cadaveric specimens [5] and b) volunteers [6]. The corridors represent one SD and were omitted when not available in the literature.

IV. DISCUSSION

The hybrid muscle implementation improved the response for the upper cervical spine kinematics relative to reported experimental data. This assertion is reinforced by comparing the C0/C1 rotation to in vivo range of motion (ROM) in combined flexion and extension (6.3°±1.6°) [12]. The C0/C1 rotation was more than 90% of the average ROM for the S_{NM}^{L} and S_{NM}^{H} cases and within one standard deviation (SD) of the ROM for the H_{NM}. Therefore, the rotations of the C0/C1 joints in the simplified NMs were the only ones exceeding reported physiologic levels for the simulated low-severity scenario. Although all models behaved similarly in the early stages of the simulations, the differences were larger beyond about 150 ms (Fig. 2). The contrast in response is due to the discretisation of the 3D muscles, creating more biofidelic lines-of-action compared to the 1D elements, and also due to the material implementation presenting rate effects or only a single stress-strain curve. A series of factors [9][13] can influence the neck response in rear impacts, but this study highlights the necessity of also replicating muscle volumetric effects for assessing tissue-level response, especially of tissues connected to the upper cervical spine. Muscle activation, not considered in the current study, is known to be important for low-severity impact with volunteers and should be investigated further. Although simplified muscle implementations are becoming more common due to their reduced computational cost, the prediction of tissue-level injury requires consideration of the 3D muscles and lines-of-action.

V. REFERENCES

[1] Yang, K., *et al.*, AP, 2018. [6] Sato, F., *et al.*, *IRCOBI*, 2016. [11] Corrales, M. A., *et al.*, *IRCOBI*, 2021.
[2] Putra, I., *et al.*, BMBB, 2022. [7] Kato, D., *et al.*, *IRCOBI*, 2018. [12] Zhou, C., *et al.*, *J Biomech*, 2020.
[3] Devane, K., *et al.*, *Traf Inj Prev*, 2019. [8] Fice, J., *et al.*, *A Biom Eng*, 2011. [13] Fice, J., *et al.*, *A Biom Eng*, 2021.
[4] Bumberger, R., *et al.*, *Int J Crashworthiness*, 2020. [9] John, J., *et al.*, *J Biomech Eng*, 2019.
[5] Stemper, B., *et al.*, AAAM, 2004. [10] Gierczycka, D., *et al.*, *Traffic Inj Prev*, 2021.