Trajectory-Based Muscle Activation in a Finite Element Neck Model for Frontal Impact Scenarios

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I. INTRODUCTION

Computational Human Body Models (HBMs) are essential for prediction of vehicle occupant response in impact events [1]. Currently, the role of neck musculature activation during these events is not well understood in the context of injury risk prediction (e.g. whiplash associated disorders) and pre-crash intervention kinematic response (e.g. autonomous braking). As such, there is a need to incorporate active muscle response in HBMs to better simulate impact response [2-3]. However, contemporary HBMs must be assessed with relevant experimental data. A human volunteer dataset [4] reports the kinematic response of the head and first thoracic vertebra in frontal impact scenarios for a range of impact severities (2–15 g), providing important data to assess HBMs with active neck musculature. Investigation of this dataset may inform muscle activation schemes that will permit the development and assessment of muscle control algorithms. In the current study, prior to implementing complex closed-loop activation schemes, open-loop control of the neck musculature activation levels was investigated over a range of impact severities.

II. METHODS

The head and neck were extracted from a current HBM (M50-O v4.5) from the Global Human Body Models Consortium (GHBMC), including detailed models of the cervical vertebrae, first thoracic vertebra (T1), intervertebral discs, cervical spine ligaments, 3D passive musculature, Hill-type 1D active musculature, and head (Fig. 1). The linear acceleration and rotational displacement boundary conditions from the volunteer experiments were applied to the model T1 for different frontal impact severities ranging from 2 g to 15 g for a total duration of 250 ms (Fig. 2) [5]. The muscles were activated in two groups: extensors and flexors [5]. A series of initial simulations was investigated to determine the sensitivity of the finite element (FE) model to the input parameters and to establish a range of boundary conditions for the optimisation study: the default activation in the model [6]; no muscle activation (i.e. lower bound); and a startle activation scheme, corresponding to maintaining the head in a neutral posture.

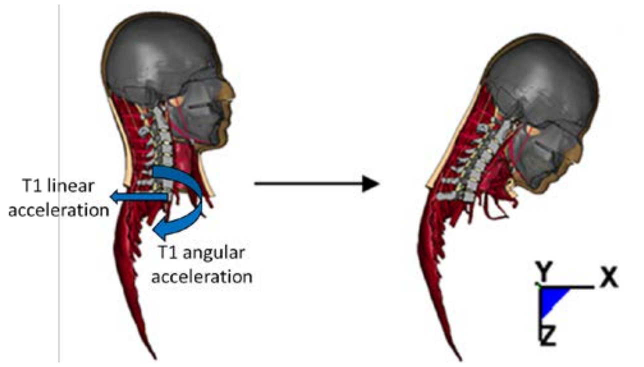

Fig. 1. GHBMC model and cross-section views of the initial (0 ms) and final (250 ms) steps of an 8 g frontal impact.

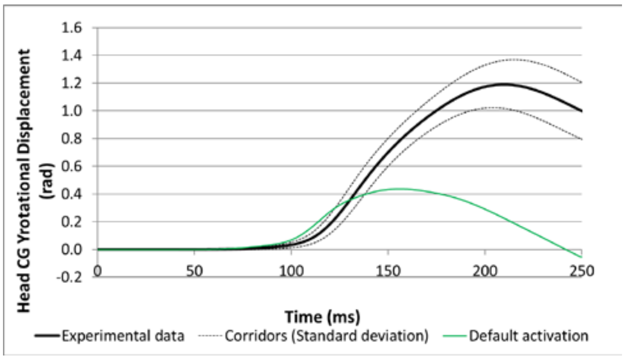

Fig. 2. Model response with the experimental data boundary conditions and default activation.

An optimisation method (linear polynomial with D-optimal point selection and domain reduction) was applied, using commercial software (LS-OPT, LSTC, Livermore, CA), to identify the activation scheme for each impact severity [7-8]. The simulations were undertaken with a commercial FE program (LS-DYNA, LSTC, Livermore, CA), focusing on two parameters: activation onset time; and activation level. The activation time

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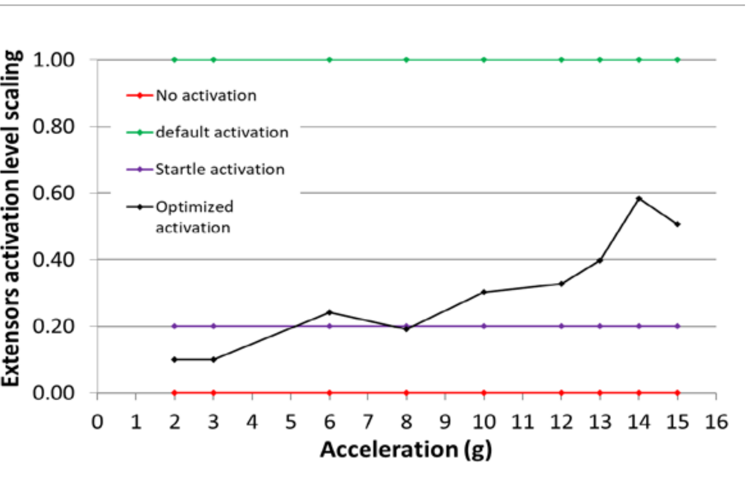
range (55–99 ms) was based on reported values for the trapezius and sternocleidomastoid [9-15]. The variation for the activation was based on the results from the initial parametric study. The model was assessed based on Y-rotational displacement, Y-rotational acceleration, X-linear displacement and X-acceleration for each of the simulations inside each optimisation iteration, and the results were compared with the respective kinematics from the experimental data using the mean square root method. The optimisation was considered converged when the average of the mean square root values varied less than 1%.

III. INITIAL FINDINGS

Varying the activation of the flexor muscles did not strongly influence the response for impact severities above 4 g and, as such, were set to the default activation of the model. The optimised muscle activation level scaling (Fig. 3) shows a trend of increasing with increasing impact severity. The correlation of the head centre of mass kinematics using the optimised scheme with the volunteer data was higher than all other schemes for all severities tested (Fig. 4), and this is also reflected in the average cross-correlation for each scheme (Table I).



ig. 3. Extensors activation scaling.

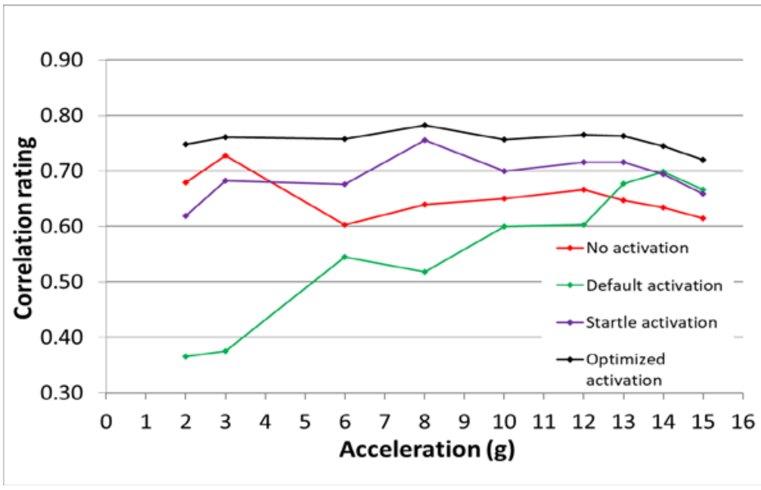


Fig. 4. Correlations with the experimental data.

TABLE I
AVERAGE CROSS-CORRELATION WITH EXPERIMENTAL DATA

Activation scheme	None	Default	Startle	Optimised
Average correlation rating	0.651	0.561	0.691	0.755

IV. DISCUSSION

On average, the model response improved by 35%, based on cross-correlation ratings, with the optimised muscle activation strategy. This activation scheme could be used to provide an improved kinematic response for the current model, while also guiding future research on activation strategies for impact scenarios. This trajectory-based activation scheme can identify strengths and areas for improvement within existing models and ultimately inform more accurate activation curves that could, along with respective kinematics datasets, validate or calibrate closed-loop muscle activation schemes. The strongest correlation was obtained with no activation for lower severity impacts in the parametric study, suggesting some of the soft tissues may be overly stiff in the model. Future research will investigate the effect of soft tissue on low severity impact response.

V. REFERENCES

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