Simulation of Average Female Rear-End Volunteer Tests using the Active ViVA OpenHBM

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I. INTRODUCTION

Whiplash associated disorders (WAD) may occur in rear-end vehicle crashes that are not particularly fast in terms of impact velocity nor extreme in terms of acceleration measured. Consequently, the injuries occurring in the neck are classified as minor and there is currently no consensus on injury mechanics. A recent model developed to study rear-end impacts is the ViVA OpenHBM, which is a 50th percentile female finite element human body model (HBM) [1-2]. It was developed because several studies have shown that women have a 1.5–3 times higher risk of suffering from long-term WAD [3]. In this study, the ViVA OpenHBM was enhanced with muscle activity. In addition, to change initial posture and spinal curvature, PIPER metadata were developed for the ViVA OpenHBM. It was then used to simulate two sets of tests that were conducted with female volunteers. Initial findings indicate that the model was able to reproduce the volunteer displacements.

II. METHODS

A finite element HBM with muscle activity, representing the average female, was compared to volunteer tests relevant for rear-impact scenarios. The ViVA OpenHBM [1-2] was enhanced to account for muscle activity in the torso and neck. For pre-positioning the model, the PIPER software [4] was used. In the first set-up, the model was compared to volunteer tests [5], where two females were subjected to a pulse on a mini-sled using a rigid seat. In a second, more complex and realistic seat set-up, the model's kinematics were compared to volunteer sled tests with eight subjects [6] (see Fig. 3).

Computational Modeling

As previously published [1-2], the detailed version of the ViVA OpenHBM was modeled with all important neck muscles using 1D beam elements and the LS-DYNA Hill-type muscle material *MAT_156 (see Fig. 1). The LS-DYNA option *PART_AVERAGED was used for muscle routing. In this study, ideal torque actuators (see Fig. 2) were implemented between all thoracic and lumbar vertebrae, instead of modelling the complex anatomy of the torso muscles. However, the main contribution of this study was the calculation of muscle activity for all the muscles included in the model. Muscle activity in the neck was determined based on the stretch reflexes of the individual muscles. The torques, representing muscle activity in the torso, were determined separately for each actuator using PID-controllers via the *DEFINE_CURVE_FUNCTION keyword. All muscle controllers took neural delay and activation dynamics into account. The controllers were not active during pre-positioning and pre-simulation, and the resulting posture was used as the initial posture and reference in the controller.

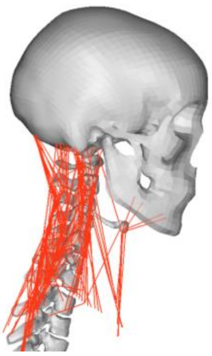

Fig. 1. Neck muscles of the ViVA OpenHBM.

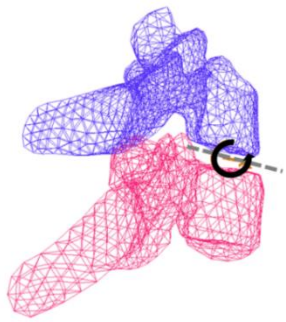

Fig. 2. Torque actuator between two vertebrae.

Fig. 3. Initial posture for simulation of sled tests conducted in [5].

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Pre-positioning using PIPER

A model's initial position is important in low-g scenarios, such as rear-impact scenarios, where spinal curvature and seat interaction play a major role. For finite element HBMs it is not a simple task to change the initial position due to, e.g. the absence of ideal joints or continuously meshed surrounding soft tissue. The PIPER software framework [4] offers various methods for morphing and scaling HBMs. Here, the ViVA OpenHBM was pre-positioned to fit onto the lab seats using PIPER software. To achieve a realistic spinal curvature, the spine predictor module was used. The model was subsequently positioned onto the lab seats in a pre-simulation under gravity before the pulse was applied.

III. INITIAL FINDINGS

Comparing head, T1 and pelvis displacements of the model to those of the volunteers, good agreement was found for both test set-ups. The realistic pre-positioning in the PIPER software played an essential role in achieving a good agreement. The parameters for the stretch reflex controller were identical to the ones used in [7]. Unfortunately, the angular displacement of the head was found to be too large in the model compared to the volunteers. In Fig. 4, the x-displacement and the angle of the head in the second set-up [6] are shown exemplarily.

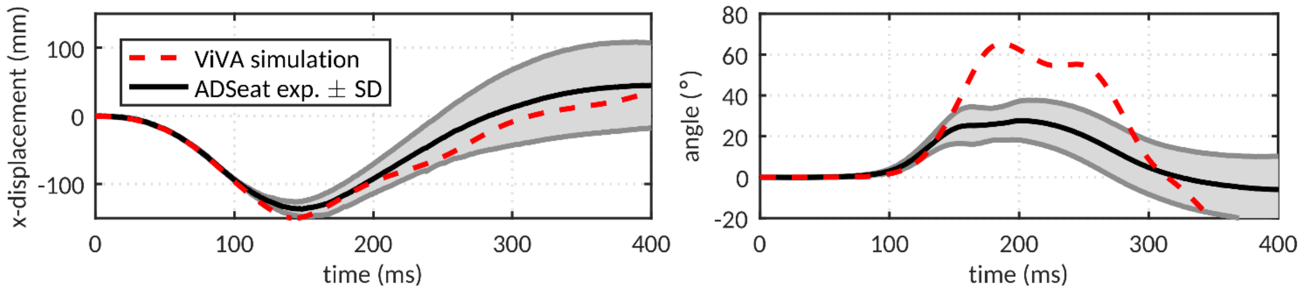

Fig. 4. x-displacement and angle of the head in the sled coordinate system in the ADSeat set-up [**Error! Reference source not found.**].

IV. DISCUSSION

The simulated displacements of the active ViVA OpenHBM developed in this study looked promising, although the erroneous head rotation indicates that model improvements and muscle tuning are required before the model can be used to study WAD. For example, analysing traditional injury criteria for the head-neck region using simulated ligament strains would be misleading with the current model. Therefore, we conclude that a stretch reflex-based controller alone is not sufficient to represent activation of the neck muscles. To improve the angular motion of the head, the existing controller might be enhanced with a PID controller based on the head angle, similar to the approach of [7].

The PIPER metadata, along with an updated version of the ViVA OpenHBM model, are available for download at: https://gitlab.inria.fr/piper/viva.

V. FUNDING

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VI. REFERENCES

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