COMPARISON OF HEAD-NECK KINEMATICS BETWEEN ISOLATED FINITE ELEMENT (FE) HEAD-NECK MODEL AND FULL-BODY MODEL IN LOW SEVERITY REAR-END IMPACT

I Putu A., Putra[1]
Johan, Iraeus[1]
Fusako, Sato[1,2]
Mats Y., Svensson[1]
Astrid, Linder[1,3]
Robert, Thomson[1]

[1]Division of Vehicle Safety, Department of Mechanics and Maritime Sciences, Chalmers University of Technology, Gothenburg, Sweden
[2]Vehicle Safety Research Group, Safety Research Division, Japan Automobile Research Institute
2530 Karima, Tsukuba, Ibaraki, Japan
[3]The Swedish National Road and Transport Research Institute, Gothenburg, Sweden

Paper Number 23-0276

ABSTRACT

The objective of the present study was to analyze whether the kinematics of an isolated head-neck model can replicate those observed on a whole body model in order to reduce simulation time in development or optimization tasks. Previous studies have shown how muscle controllers improved head-neck kinematics responses over a passive neck muscle implementation. These studies used volunteer T1 displacement time histories prescribed on the model T1 as the loading input to develop the neck controller characteristics. It was not clear whether the implementation of a controller based on volunteer kinematics with an isolated head-neck model was directly transferable to a full-body model. The current study shows that the head-neck model produced almost identical responses as the full body model for the first 200ms of the event for most kinematic variables. The head rotational displacement corresponded well during the first 150ms. The isolated head-neck model predicted more displacement and rotations than when mounted on a full-body model. The current simplification of a head-neck model still produced reasonable kinematic responses during the critical time period to assess soft tissue neck injuries, making it suitable for developing and tuning neck muscle controllers.

INTRODUCTION

Finite Element (FE) Human Body Models (HBMs) have been a powerful and essential tool when studying road user safety. However, until recently, FE HBMs that represented average female anthropometry did not exist. To fill this gap, an open-source HBM of the 50th percentile female stature called VIVA+ 50F (John et al. 2022a, 2022b) was developed (John et al. 2022a, 2022b) and validated against Post-Mortem Human Subject (PHMS) responses in a rear impact (John et al. 2022b). It was further developed by adding active reflexive neck muscle controllers (Putra et al. 2022). Neck muscle activities have been shown to influence the head-neck kinematics during rear-impact volunteer tests (Brault et al. 2000, Siegmund et al. 2003, Blouin et al. 2006, G.P Siegmund, 2011, Dehner et al. 2013, Mang et al. 2015).

With added muscle controllers, the VIVA+ 50F head and neck kinematics was improved over a passive neck implementation compared to volunteer responses (Putra et al. 2022). The model used in previous studies consisted of a head-neck model with the volunteer T1 displacement time histories prescribed on the model T1. Therefore, it was not clear whether the implementation and optimization of the active muscle controller using the isolated head-neck model could also be used in the full-body model. In addition, only a female model was available at the time of the implementing an active neck muscle controller. Consequently, a direct comparison between female and male models could not be conducted.

Based on identified limitations, the objective of the present study was to analyze whether the isolated head-neck model can replicate the full-body model head-neck kinematics and be used to develop an active muscle controller strategy.

MATERIAL AND METHODS

The overall flowchart of the study is presented in Figure 1 below. Similar processes for both the average female and male VIVA+ models, VIVA+ 50F and 50M, were conducted, but only the female model is shown in the flowchart. First, the full-body model was run, using the boundary conditions matching the volunteers’ test setup to generate the T1 linear and rotational accelerations. The generated T1 accelerations were then prescribed to the isolated VIVA+ head-neck models. This was conducted to ensure that both models had similar T1 accelerations. Finally, the head C.G. and cervical vertebra C.G. kinematics of the VIVA+ full-body models and VIVA+ head-neck models were compared and analyzed.

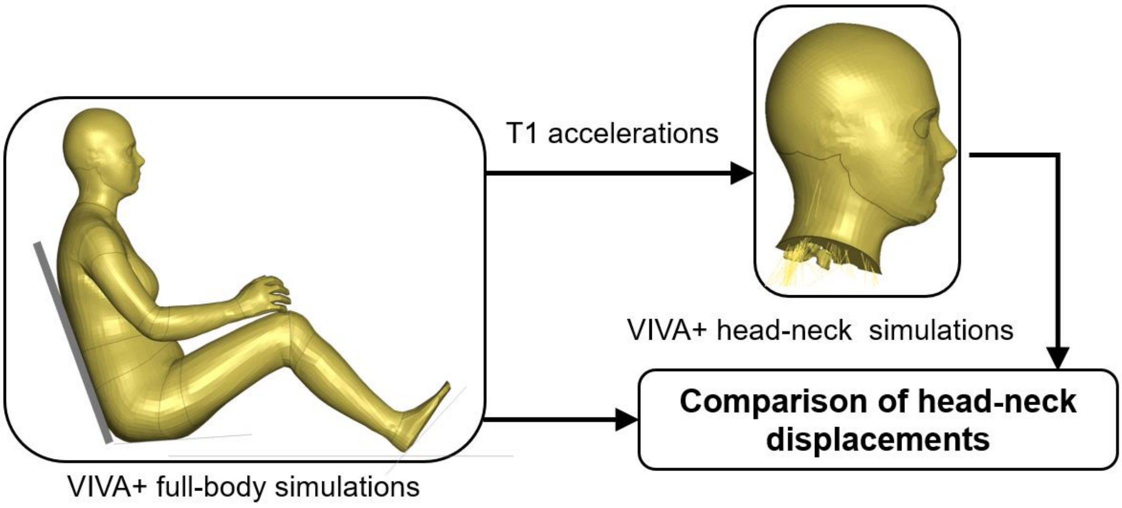

Figure. 1 Flow Chart of the Present Study's Methods

VIVA+ Female and Male Occupant Human Body Model

The baseline of VIVA+ FE HBM is an average 50th percentile female model, which is morphed to create an average 50th percentile male (Figure 2). The male and female models have identical elements definitions with the gender specific nodal coordinates defined by several statistical shape models describing the outer body shape, ribcage, femur, tibia and pelvis (John et al. 2022a). Besides geometry changes, differences in male and female head mass, soft tissues densities, knee ligaments characteristics, and quadriceps muscle stiffness were included (John et al. 2022a). In the present study, sub-models that consist only of head-neck were created by cutting both average female and male models below the first thoracic vertebrae (T1).

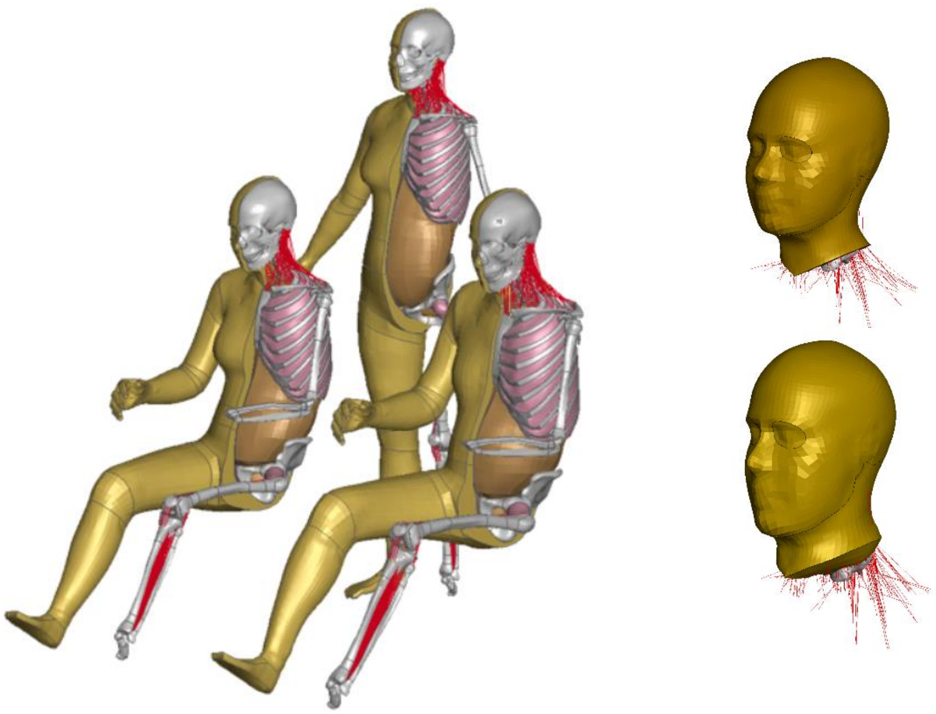

Figure. 2 VIVA+ Finite Element Human Body Models and Isolated VIVA+ 50F and 50M Head-Neck Models

Rear end Impact Volunteer Test based on Sato et al. 2014

Volunteer kinematic data was derived from the volunteer test series of Sato et al. (2014), which was a low-speed, rear-impact sled test with a delta velocity of 5.8km/h and peak acceleration of $42m/s^2$. Four male volunteers and two female volunteers were seated in a rigid seat (seatback angle 20 degrees from vertical) without a head restraint. Head C.G displacements and accelerations, T1 C.G displacements, and C1-C7 rotational displacements (recorded with a high-speed X-ray camera) data were used in the present study to compare the VIVA+ models kinematics.

Simulation Set-up

Comparisons of head and neck kinematics responses between the VIVA+ head-neck models and the full-body models in their passive neck muscle configurations were conducted. The aim was to evaluate whether the head-neck system of the VIVA+ model would generate similar head and neck kinematics in the isolated head-neck compared to the full-body model. This would demonstrate that the isolated model was a suitable basis for active muscle controller optimizations to identify the controller characteristics (Putra et al. 2020).

Full HBM model simulations of the VIVA+ 50F and 50M models were done following the Sato et al. (2014) set-up (Figure 3) to generate T1 accelerations. Sled acceleration from the experiment was prescribed on the seat model. The duration of each simulation was 650ms, with the first 450ms used to settle the model under gravitational acceleration. The T1 accelerations produced by the full-body model simulations were then used to prescribe the T1 motion for the isolated head-neck models. In the head-neck model, the lower nodes of the skin and several nodes of the soft tissues were constrained to move with the T1 (Figure 4).

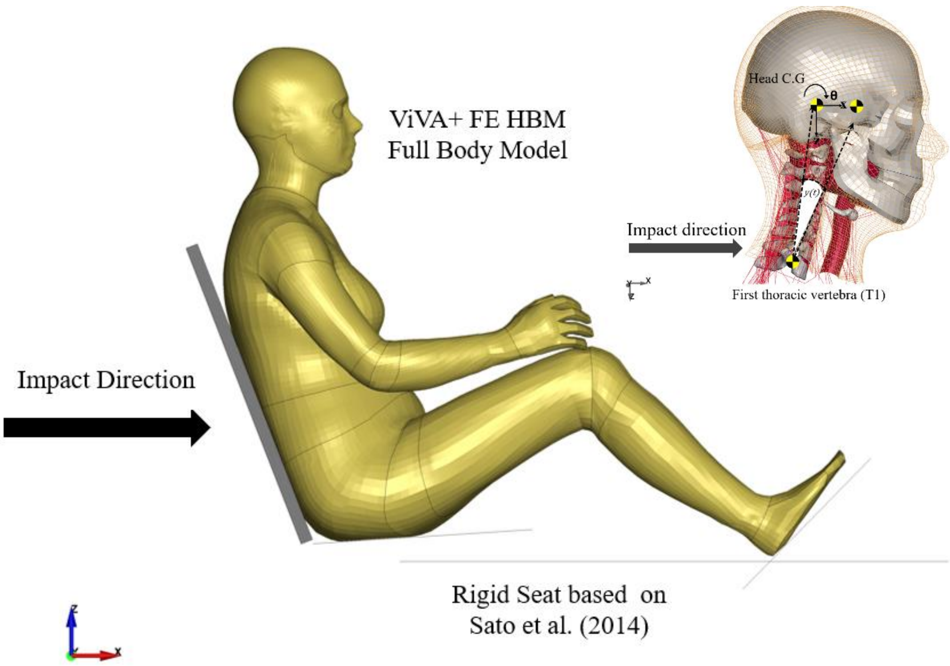

Figure. 3 Simulations set-up of VIVA+ 50F and 50M Head-Neck and Full Body Model based on Sato et al. (2014).

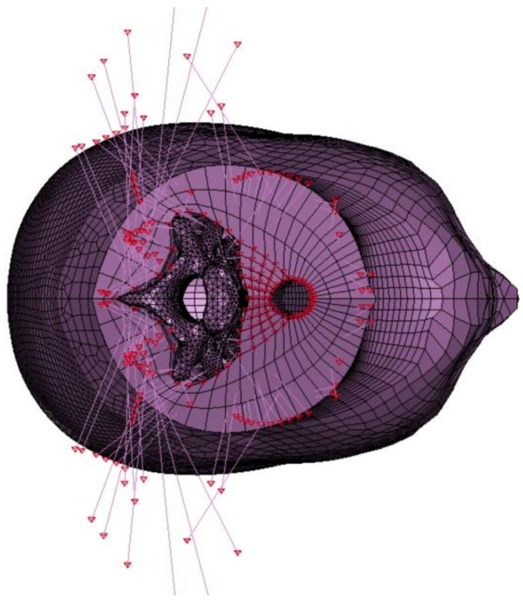

Figure. 4 List of Constrained Nodes to Allow the Isolated Head-Neck Model Moves with the T1

RESULTS

The T1 C.G x- and z-linear displacement and T1 rotational y-displacement of the VIVA+ 50F and 50M are presented with the volunteer kinematics in Figure 5. The goal of this study was to compare the isolated head-neck kinematics to those produced by a full HBM simulation. The full HBM calibration and validation to the sled test was beyond the scope of this study and not necessary for the purpose of model comparison. Although the responses are not identical, the results of the simulations are similar to those of the volunteers.

The T1 C.G. displacements (x- and z-) and T1 C.G. y-rotation from full-body simulations (Figure 5) were then prescribed to the T1 C.G. of the head-neck models. Figure 6 show that the female head-neck model could replicate the full-size female model's responses for the first 200ms except for head C.G. rotational displacement 7) when both models have identical T1 kinematics. The head-neck model's head C.G. rotations only followed the full-body model's head rotations for 150ms. These responses were observed for both female (Figure 6) and male (Figure 7).

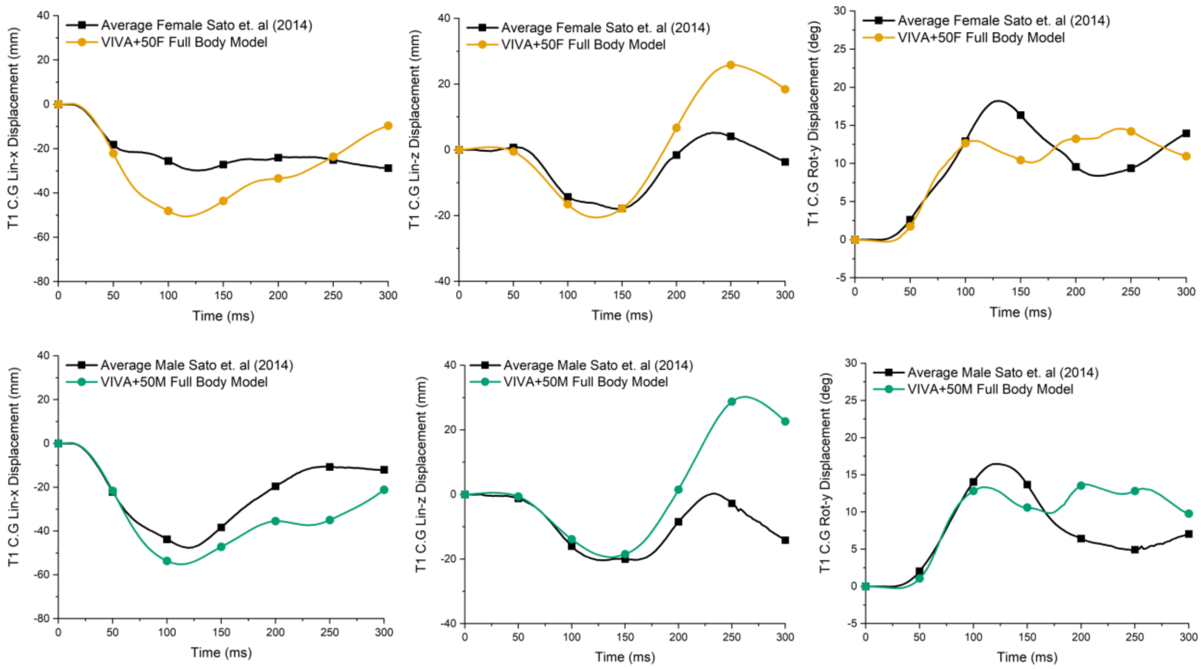

Figure 5. Comparison of T1 C.G Displacements and Rotation based on Full-Body Model Simulation with VIVA + 50F and VIVA+ 50M and volunteer tests from Sato et al. (2014) at 5.8km/h

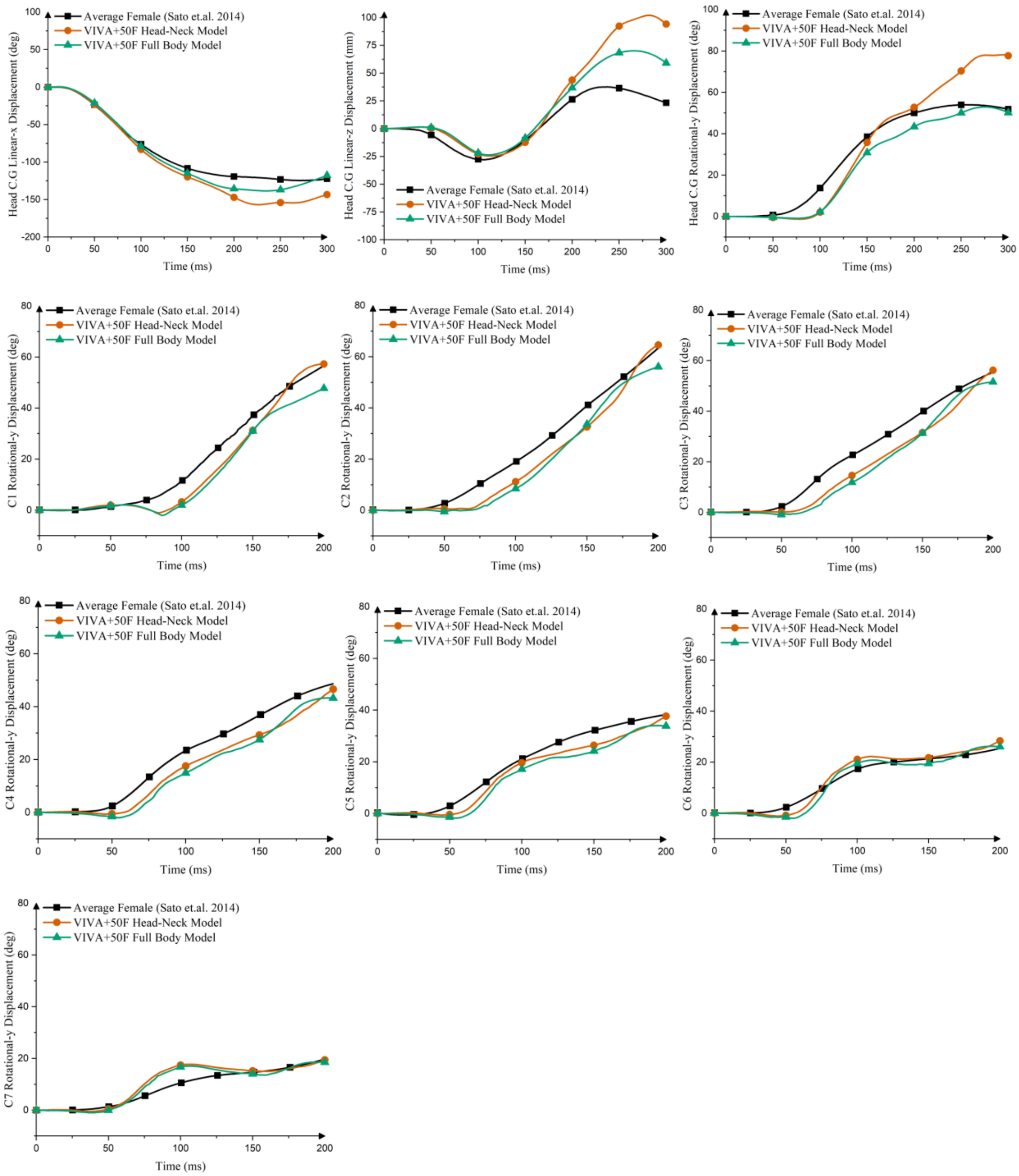

Figure 6. Comparison of Head-Neck Kinematics of VIVA+ Head-Neck Female Model and Full-Body Female Model with Similar T1 Displacement and Rotations as Input.

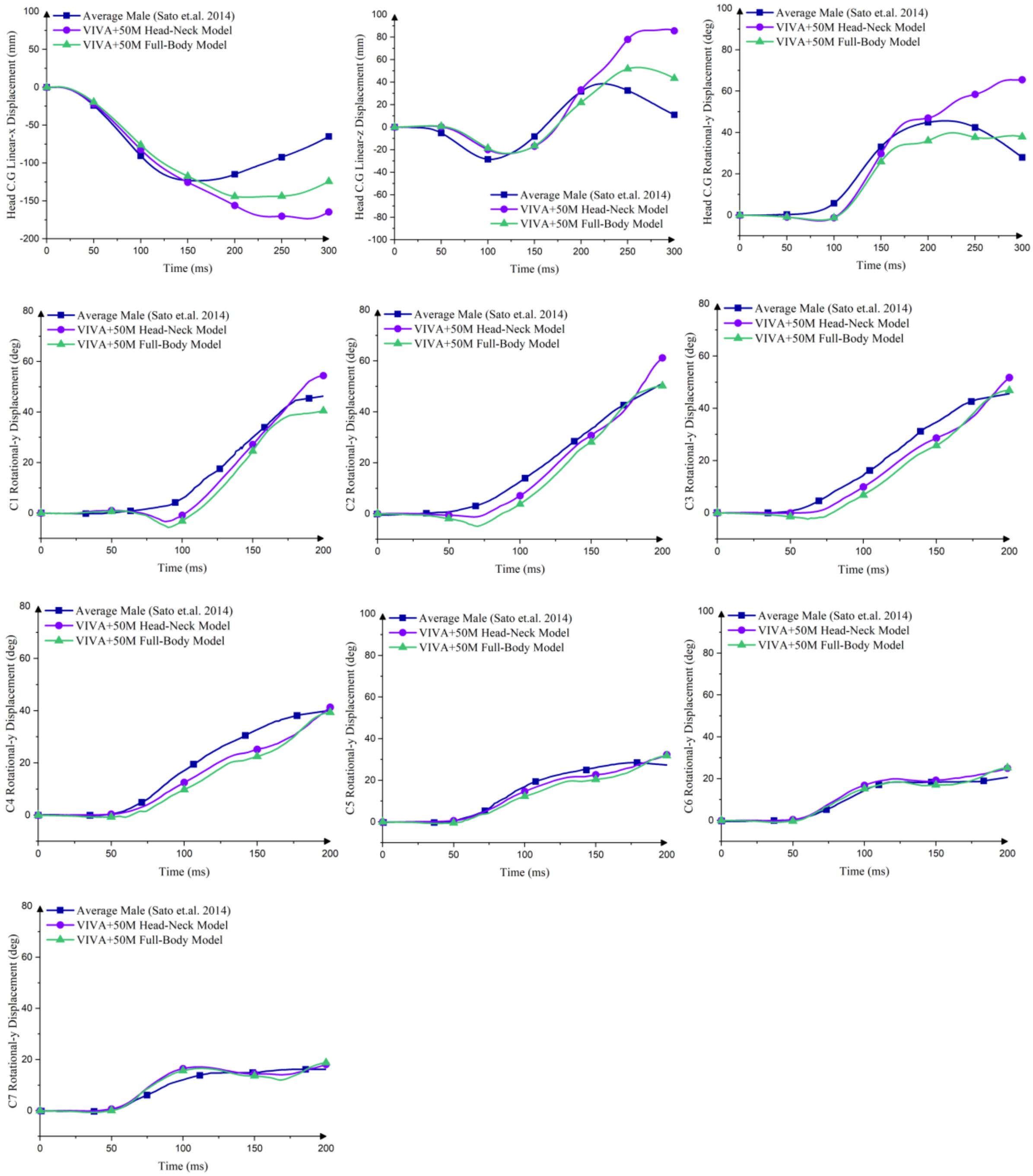

Figure 7. Comparison of Head-Neck Kinematics of VIVA+ Head-Neck Male Model and Full-Body Male Model with Similar T1 Displacement and Rotations as Input.

DISCUSSION

The objective of the current study was to assess whether a head-neck model with prescribed T1 displacements and rotations can be used to develop an active muscle controller strategy. The main reason was that running active muscle controller optimizations using the full-body model is computationally very expensive. However, running an isolated head-neck model under similar loading conditions reduces computational time by almost 75%.

The comparison between head-neck kinematics of a head-neck model and full-body model for both female and male models revealed that the head-neck model produced almost identical responses up until 200ms for most kinematic variables but only until 150ms for the head rotational displacement. Currently, most of the whiplash injuries hypotheses are associated with the retraction phase of the neck (Svensson et al. 1993, Yoganandan et al. 2002, Ono et al. 2006), which occurs in the first 150ms after impact. Based on these hypotheses and the simulation results, it was suggested that the isolated head-neck model could be used to develop an active muscle controller that produces similar head-neck kinematics responses as a full-body model for the duration of the event where soft tissue neck injuries are hypotheses to occur.

The main differences in kinematics between the head-neck model and full-body of the VIVA+ 50F and 50M models occurred at 150ms (in head rotational y-displacement) and at 200ms (in head linear x- and z-displacement). The head-neck model most often produced more displacements and rotations than the full-body model. The main reason for the less stiff response of the head-neck model can be explained by the difference in modelling the neck's soft tissue and skin. In the full-body model, the neck's soft tissue and skin are “seamlessly” connected to the soft tissues and skin of the upper torso with the lower muscle attachments moving with their anatomical structures. However, in the head-neck model the lower nodes of the skin and several nodes of the soft tissues were constrained to move with the T1 vertebra. The lower muscle attachments were also fixed to a plane that only followed the motions of T1 and not their true anatomical motions.

Figure 4 shows that not all nodes on the lower neck surface were constrained. Simulations conducted with this definition resulted in an overly stiff response. The interaction of structures that cross this intersection between the neck and upper torso vary across the cross-section and the constraints illustrated in Figure 4 gave the best results in the study. Further investigations of the boundary conditions could produce better results if the sub-model should agree with the full model over a longer time period. This would require even further investigation of the neck muscle attachments below the section plane as these affect the muscle forces and resulting influence on the response.

CONCLUSIONS

Computer simulations of an isolated head-neck complex can reasonably duplicate the associated kinematics observed in a full-body model. Reducing the model to a sub-model with only the relevant body segments reduced simulation times by 75%. This model reduction facilitates optimization studies to obtain neck control characteristics. These optimizations require 150+ simulations and more than one optimization run are typically needed to explore the model parameter solution space. The boundary conditions for the sub-model are important to define and the existing implementation is suitable for the study of head-neck kinematics in low severity rear impacts.

ACKNOWLEDGEMENT

This study was conducted within the project VIRTUAL and was received funding from the European Union Horizon 2020 Research and Innovation Programme under Grant Agreement No. 768960.
EU Trademark Application No. 01880153

REFERENCES

Blouin, J. S., Inglis, J. T., and Siegmund, G. P. (2006). Auditory startle alters the response of human subjects exposed to a single whiplash-like perturbation. *Spine (Phila. Pa. 1976).* 31, 146–154. doi: 10.1097/01.brs.0000195157.75056.df.

Brault, J. R., Siegmund, G. P., and Wheeler, J. B. (2000). Cervical muscle response during whiplash: Evidence of a lengthening muscle contraction. *Clin. Biomech.* 15, 426–435. doi: 10.1016/S0268-0033(99)00097-2.

Dehner, C., Schick, S., Kraus, M., Scola, A., Hell, W., and Kramer, M. (2013). Muscle Activity Influence on the Kinematics of the Cervical Spine in Frontal Tests. *Traffic Inj. Prev.* 14, 607–613. doi: 10.1080/15389588.2012.734937.

John J, Klug C, Kranjec M, Svenning E and Iraeus J (2022a) Hello, world! VIVA+: A human body model lineup to evaluate sex-differences in crash protection. Front. Bioeng. Biotechnol. 10:918904. doi:10.3389/fbioe.2022.918904

John, J., Putra, I. P. A., and Iraeus, J. (2022b). Finite element human body models to study sex-differences in whiplash: Validation of VIVA+ passive responses in rear-impact. IRCOBI Conference. IRC-22-36, Porto, Portugal

Mang, D. W. H., Siegmund, G. P., Brown, H. J., Goonetilleke, S. C., and Blouin, J. S. (2015). Loud preimpact tones reduce the cervical multifidus muscle response during rear-end collisions: A potential method for reducing whiplash injuries. *Spine J.* 15, 153–161. doi: 10.1016/j.spinee.2014.08.002.

Ono, K., Ejima, S., Suzuki, Y., Kaneoka, K., Fukushima, M., and Ujihashi, S. (2006). Prediction of Neck Injury Risk Based on the Analysis of Localized Cervical Vertebral Motion of Human Volunteers During Low-Speed Rear-end impacts. IRCOBI Conf. Proc., 103–113

Putra, I. P. A., Iraeus, J., Sato, F., Svensson, M. Y., & Thomson, R. (2022). Finite element human body models with active reflexive muscles suitable for sex based whiplash injury prediction. Frontiers in bioengineering and biotechnology, 10, 968939. https://doi.org/10.3389/fbioe.2022.968939

Putra, I. P. A., Iraeus, J., Sato, F., Svensson, M. Y., Linder, A., and Thomson, R. (2020). Optimization of Female Head–Neck Model with Active Reflexive Cervical Muscles in Low Severity Rear Impact Collisions. *Ann. Biomed. Eng.*, 1–14.

Sato, F., Nakajima, T., Ono, K., and Svensson, M. (2014). Dynamic Cervical Vertebral Motion of Female and Male Volunteers and Analysis of its Interaction with Head/Neck/Torso Behavior during Low-Speed Rear. *Ircobi.Org*, 227–249. Available at: http://www.ircobi.org/downloads/irc14/pdf_files/31.pdf.

Siegmund, G. P. (2011). What occupant kinematics and neuromuscular responses tell us about whiplash injury. *Spine (Phila. Pa. 1976).* 36, S175–S179. doi: 10.1097/BRS.0b013e3182387d71.

Siegmund, G. P., Sanderson, D. J., Myers, B. S., and Inglis, J. T. (2003). Awareness affects the response of human subjects exposed to a single whiplash-like perturbation. *Spine (Phila. Pa. 1976).* 28, 671–679. doi: 10.1097/00007632-200304010-00010.

Svensson, M. Y., Aldman, B., Hansson, H. a, Lövsund, P., Seeman, T., Suneson, a, et al. (1993). Pressure Effects in the Spinal Canal during Whiplash Extension Motion: A Possible Cause of Injury to the Cervical Spinal Ganglia. Int. IRCOBI Conf. Biomech. Impacts, 189–200.

Yoganandan, N., Pintar, F. A., and Cusick, J. F. (2002). Biomechanical analyses of whiplash injuries using an experimental model. Accid. Anal. Prev. 34, 663–671. doi: 10.1016/S0001-4575(01)00066-5.