TABLE III
SENSITIVITY OF CORRELATION DATA TO STRAIN THRESHOLD

Strain Threshold (%)	CORA Rating
1	0.578
2	0.587
3	0.619
4	0.630
5	0.639
6	0.662
7	0.659
8	0.647
9	0.651
10	0.639

TABLE IV
SENSITIVITY OF CORRELATION DATA TO TIME DELAY

Neuromuscular Delay (ms)	CORA Rating
18.1	0.653
21.3	0.637
24.5	0.639
27.7	0.663
30.9	0.656

Head Fall Test

The average male experimental response during the head fall tests can be compared with responses seen using the GHBMC in Fig. 6. Initially, the vertebrae of the GHBMC were constrained from T1 downward. With these constraints the head COG displacement of the passive GHBMC is lower than that of the middle 50th percentile of experimental subjects. While male subjects experience an average maximum displacement of 79.7 mm that occurs 191 ms after head release before the force produced by the muscles can overcome that of gravity, the GHBMC without active muscle contributions has 31.3 mm of displacement occurring over a time period of 162.5 ms. Subsequently the constrains were changed to constrain all movement from T3 downward, allowing translational and rotational movement of T1 and T2. This is likely a more realistic representation of the motion occurring and results in a further displacement of the head to 45.3 mm. The model with vertebral constraints beginning at T3 is then used for the remaining simulations, as further reduction of the constraints was deemed unrealistic in the experimental setup.

To determine which components are absorbing additional energy and increasing the stiffness of the model, the internal energy of the tissues found in the neck are plotted in Fig. 7. As would be expected by their relatively large volume in the neck and the deformation seen in simulation, the 3D muscles absorb a large amount of energy. The 1D ligamentous structure elements absorb relatively little energy in comparison. Unexpectedly, the parts comprising the skin absorb more energy than any of the other components studied. In order to determine the contributions to neck stiffness added by the skin, all three layers were removed from the model. When the skin is removed, the maximum displacement of the head COG is 91.5 mm. Simulation results of the modified neck now fall within one standard deviation (51.6−107.8 mm) of the male experimental responses, and the passive head and neck model falls farther than the average volunteer with active muscle responses.

When muscle activation dynamics are added to the model without skin, maximum displacement is reduced and occurs earlier. The first muscles from the flexor group begin to activate at 69.4 ms. As more muscle elements pass the strain threshold and are recruited, the neck flexes back upward after reaching its maximal displacement of 65.7 mm at 186 ms. The recruitment of muscles within the flexor and extensor groups is shown in Fig. 8 and 9, with the contributions of the flexors outweighing that of the extensors. Of the 250 flexor beam elements, 28 were activated during the simulation. Of the 666 extensor elements, only six were activated.

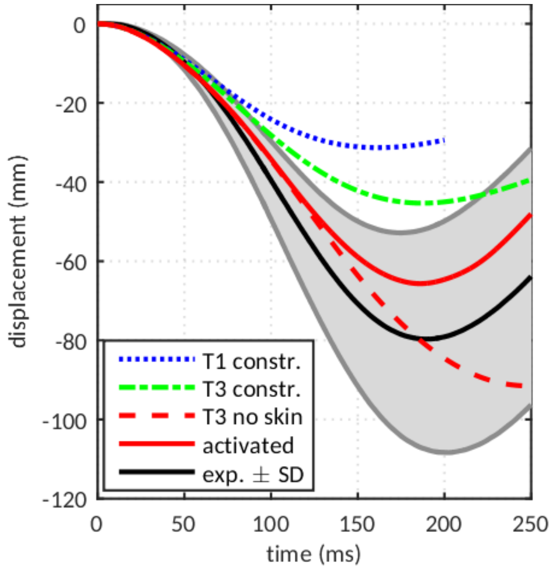

Fig. 6. Displacement resulting from head fall test when modifications to the GHBMC are made to alter neck stiffness.

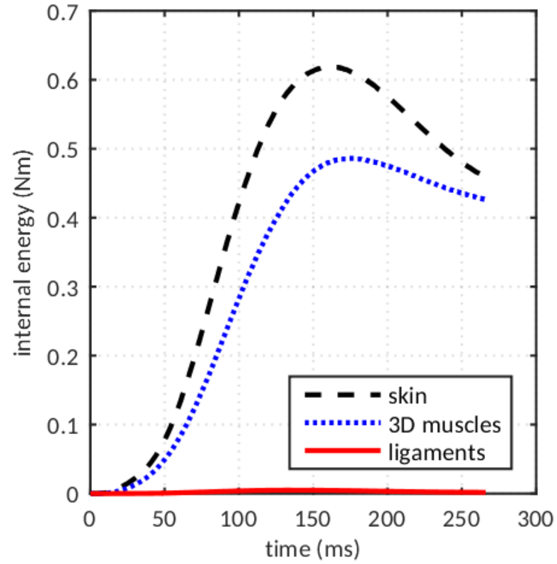

Fig. 7. Internal energy in different parts of the neck during head fall test with vertebral constraints beginning at T1.

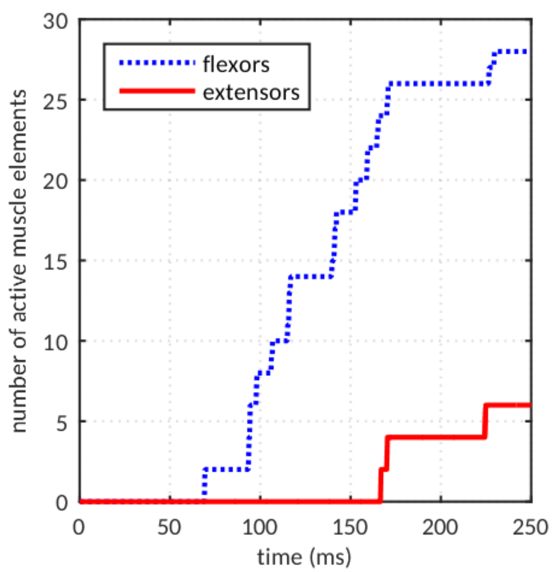

Fig. 8. Comparison of the number of flexors versus extensor beam elements activated during the simulated head fall tests.

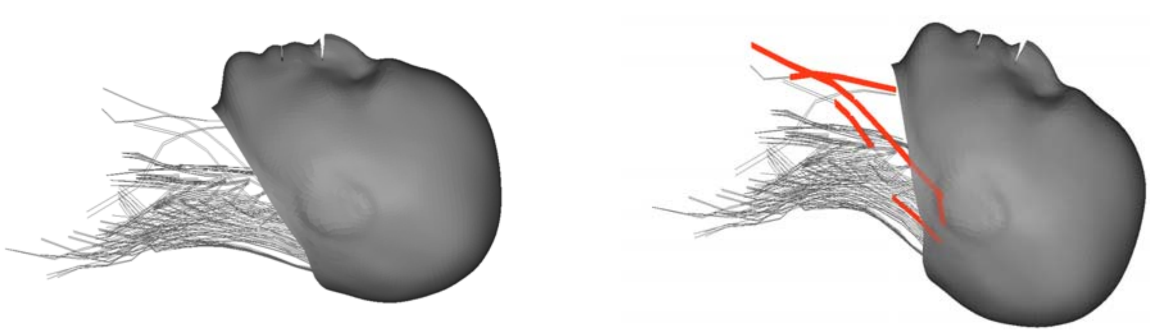

Fig. 9. Activated muscle elements during a head fall test are highlighted in red, while inactive elements remain grey. Shown on the right is the end state of the model (time = 250 ms), which can be compared to the initial state on the left.

IV. DISCUSSION

Validation of the activation dynamics subroutine and resulting muscle model was performed by replicating porcine contraction tests. The general agreement between experimental and simulation results in both contraction velocity magnitude and timing indicate that MAT_156 is a sufficiently complex model for modelling active and passive properties on the full muscle level. Peak shortening velocity in simulation occurs between 40 and 70 ms, with an average time of 52 ms after muscle activity onset. If peak velocity is assumed to correspond to peak activity, the data is well aligned with literature that has found a 55 ms delay between the onset of muscle activity and peak activation level [6].

Greater impact velocities have been shown to result in a decreased time to onset of muscle activity in occupants. When volunteer subjects in rear-speed collisions were subjected to changes in velocities of 4 km/h and 8 km/h, muscle activation occurred earlier in subjects at the higher impact speed [41]. As the strain threshold is reached more quickly in our model with higher loading scenarios, the muscle activation time decreases with increased loading. Strain rate dependencies have not yet been fully included in the model, despite their potentially large effects. These effects will be included in the neural excitation factor. Neural excitation has been shown to vary based on the load, from 0.54 to 0.99 within the loading range of interest, with larger loads resulting in higher excitation values [37].

Though the initial values for neural delay and strain were determined through a parameter study, a sensitivity analysis reveals that more optimal values may still exist. While the influence of a small change in time delay does not largely affect the correlation results, the model is more sensitive to the values chosen for the strain threshold. Though the initial subroutine begins with what the authors believe to be reasonable values for strain activation threshold and muscle latency, further work must be done to determine the optimal values for an average male subject, optimising based on a full human model rather than a single muscle test. Due to the 74 hour run time of the simulation across 20 CPUs, an MPP compatible version of the subroutine should be implemented before further parameter modification is tested.

Muscle activity can significantly affect the movement of the head and neck in low-speed collisions [42]. If the GHBMC is to be used in these scenarios, it is important to begin with a passive neck stiffness that resembles a relaxed occupant. Though the GHBMC neck has been validated in frontal, rear and side impact, validation scenarios found in [28–31] represent more severe impacts than is suitable for studying muscle activity in the head fall test. While neck displacement timing of the model agrees very well with experimental results, the magnitude of the displacement, though within one standard deviation of experimental data, is still below the one of an average male volunteer. Reducing the stiffness of the neck that results from the skin offers one possible mechanism to improve kinematics, but the model may still need further stiffness reduction.

The activation of flexor muscles in response to head extension in Fig. 8 and 9 indicate that overall, the correct muscles are activated by the subroutine. The results of the sensitivity analysis and the activation of elements belonging to the extensor group indicate that improvements can still be made to the model. The first of which may be increasing the strain threshold required to begin the muscle activation process. This would allow the head to fall farther and fewer extensors would be likely to reach the strain threshold. There also exists the question of whether to activate single beam elements or the entire muscle. In the initial implementation, beam elements are activated individually which is based on the physiology of a muscle spindle. They are embedded in muscle fibres, which are bundled into fascicles with an average length of 1.5-18.2 cm in the neck [43]. This is close to the length of a beam element in the GHBMC model, which fall in the range of 0.49-7.66 cm. This approach justifies the use of muscle spindle delay and strain threshold parameters for the activation scheme. A more physiologically representative approach to be tested in future iterations may be to continue to analyse the strain in individual beam elements, but activate the entire muscle when the strain threshold is surpassed. This would allow for a more straightforward comparison of the cross-sectional area of the activated muscles across different muscle groups. Controlling the whole muscle would also be necessary in order include occupant bracing and other voluntary movement in future iterations.

The present work is most relevant when considered in the context of a relaxed volunteer, as no pre-impact bracing response or other voluntary responses are included, but protective muscle reflex activity is considered. The control approach used will also allow for the inclusion of more advanced activation dynamic models, such as the model proposed by Hatze [44], or an intentional occupant manoeuvre. This approach also allows for the development of a hierarchical approach where the neural stimulation either stems from a stretch mediated response or from an intentional movement initiated in the brain. The conscious movement could be calculated by the highest hierarchy of the control loop and could be composed of a feedforward control in combination with inverse dynamics similar to the control used for robotic manipulators.

V. CONCLUSIONS

In this study, strain-dependent muscle activation dynamics were developed for use in a standard muscle material model used in FE-HBMs through a subroutine in LS-DYNA. When the activation dynamics were tested under a set of loading conditions, timing and magnitude of shortening velocity of the muscles agreed well with literature and experimental results. Initial parameters used to calculate activation and excitation levels resulted in CORA ratings of “good” and “fair” when compared to experimental data, though correlation can be improved by further study of the critical parameters. To determine its suitability for low-impact loading scenarios, the GHBMC was subjected to a head fall test in both active and passive modes. After modifications to reduce neck stiffness, magnitude and timing of the displacement are in reasonable agreement with experimental results. These results suggest feasibility of the activation method to describe reflexive muscle activation dynamics within the FE-HBM for use in predicting head and neck kinematics in low-severity impact scenarios.

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VII. REFERENCES

[1] Jarawan, E., Mohan, D., *et al.* 2004 World report on road traffic injury prevention.
[2] Commision of the European Communities. (2001) White Paper European Transport Policy for 2010: time to decide.
[3] Quinlan, K.P., Annest, J.L., Myers, B., Ryan, G., and Hill, H. (2004) Neck strains and sprains among motor vehicle occupants—United States, 2000. *Accident Analysis & Prevention*. **36**(1): pp. 21-27.
[4] Chappuis, G. and Soltermann, B. (2008) Number and cost of claims linked to minor cervical trauma in Europe: results from the comparative study by CEA, AREDOC and CEREDOC. *European Spine Journal*, 17(10): pp. 1350-1357
[5] Carroll, L.J., Holm, L.W., *et al*. (2009) Course and prognostic factors for neck pain in whiplash-associated disorders (WAD): results of the bone and joint decade 2000–2010 task force on neck pain and its associated disorders. *Journal of manipulative and physiological therapeutics*, **32**(2): pp. S97-S107.
[6] Van der Horst, M.J., Thunnissen, J., Happee, R., Van Haaster, R., and Wismans, J. The influence of muscle activity on head-neck response during impact. *SAE Conference Proceedings*, 1997.
[7] De Jager, M., Sauren, A., Thunnissen, J., and Wismans, J. Global and a detailed mathematical model for head-neck dynamics. *Proceedings of the Stapp Car Crash Conference*, 1996.
[8] Khodaei, H., Mostofizadeh, S., Brolin, K., Johansson, H., and Östh, J. (2013) Simulation of active skeletal muscle tissue with a transversely isotropic viscohyperelastic continuum material model. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, **227**(5): pp. 571-580
[9] Begeman, P., King, A., Levine, R., and Viano, D.C. (1980) Biodynamic response of the musculoskeletal system to impact acceleration, SAE Technical Paper.
[10] Brolin, K., Hedenstierna, S., Halldin, P., Bass, C., and Alem, N. (2008) The importance of muscle tension on the outcome of impacts with a major vertical component. *International Journal of Crashworthiness*, **13**(5): pp. 487-498.
[11] Brolin, K., Halldin, P., and Leijonhufvud, I. (2005) The effect of muscle activation on neck response. *Traffic injury prevention*, **6**(1): pp. 67-76.
[12] Meijer, R., Van Hassel, E., *et al*. Development of a multi-body human model that predicts active and passive human behaviour. *Proceedings of the International Conference on Biomechanics of Impact IRCOBI, Dublin-Ireland*, 2012.
[13] Östh, J., Brolin, K., and Happee, R. (2012) Active muscle response using feedback control of a finite element human arm model. *Computer methods in biomechanics and biomedical engineering*, **15**(4): pp. 347-361.

[14] Kistemaker, D.A., Van Soest, A.K.J., and Bobbert, M.F. (2006) Is equilibrium point control feasible for fast goal-directed single-joint movements? *Journal of Neurophysiology*, **95**(5): pp. 2898-2912.

[15] Oliveira, A.R., Gonçalves, S.B., de Carvalho, M., and Silva, M.T. (2016) Development of a musculotendon model within the framework of multibody systems dynamics. *Computational Methods in Applied Sciences*, **42**: pp. 213-237.

[16] Ambrósio, J.A. and Kecskeméthy, A. Multibody dynamics of biomechanical models for human motion via optimization. *Multibody Dynamics*. pp. 245-272. *Springer*, 2007

[17] User Manual: M50 Occupant Version 4.3 for LS-DYNA.

[18] Gayzik, F.S., Moreno, D.P., Vavalle, N.A., Rhyne, A.C., and Stitzel, J.D. Development of the Global Human Body Models Consortium mid-sized male full body model. *Proceedings of the Injury Biomechanics Research Thirty-Ninth International Workshop*, 2011.

[19] Cronin, D.S. (2014) Finite element modeling of potential cervical spine pain sources in neutral position low speed rear impact. *Journal of the mechanical behavior of biomedical materials*, **33**: pp. 55-66.

[20] Fice, J.B. and Cronin, D.S. (2012) Investigation of whiplash injuries in the upper cervical spine using a detailed neck model. *Journal of biomechanics*,. **45**(6): pp. 1098-1102.

[21] Panzer, M.B. and Cronin, D.S. (2009) C4–C5 segment finite element model development, validation, and load-sharing investigation. *Journal of biomechanics*, **42**(4): pp. 480-490.

[22] Panzer, M.B., Fice, J.B., and Cronin, D.S. (2011) Cervical spine response in frontal crash. *Medical engineering & physics*, **33**(9): pp. 1147-1159.

[23] Ivancic, P., Panjabi, M.M., Ito, S., Cripton, P., and Wang, J. (2005) Biofidelic whole cervical spine model with muscle force replication for whiplash simulation. *European spine journal*, **14**(4): pp. 346-355.

[24] Panjabi, M.M., Ito, S., Pearson, A.M., and Ivancic, P.C. (2004) Injury mechanisms of the cervical intervertebral disc during simulated whiplash. *Spine*, **29**(11): pp. 1217-1225.

[25] Panjabi, M.M., Pearson, A.M., *et al*. Cervical spine ligament injury during simulated frontal impact. *Spine*, 2004. **29**(21): pp. 2395-2403.

[26] Pearson, A.M., Ivancic, P.C., Ito, S., and Panjabi, M.M. (2004) Facet joint kinematics and injury mechanisms during simulated whiplash. *Spine*, **29**(4): pp. 390-397.

[27] Ito, S., Ivancic, P.C., *et al*. (2005) Cervical intervertebral disc injury during simulated frontal impact. *European Spine Journal*, **14**(4): pp. 356-365.

[28] Davidsson, J., Deutscher, C., Hell, W., Lövsund, P., and Svensson, M.Y. (2001) Human volunteer kinematics in rear-end sled collisions. *Crash prevention and injury control*, **2**(4): pp. 319-333.

[29] Hynd, D., Svensson, M., Trosseille, X., van Ratingen, M., and Davidsson, J. (2007) Dummy requirements and injury criteria for a low-speed rear impact whiplash dummy. *European Enhanced Vehicle-Safety Committee, EEVC WG12 A*, **505**: pp. 2007.

[30] Thunnissen, J., Wismans, J., Ewing, C., and Thomas, D. Human volunteer head-neck response in frontal flexion: a new analysis. *Proceedings of Stapp Car Crash Conference*, 1995.

[31] Wismans, J., Van Oorschot, H., and Woltring, H. (1986) Omni-directional human head-neck response. SAE Technical Paper.

[32] Winters, J.M. (1995) An improved muscle-reflex actuator for use in large-scale neuromusculoskeletal models. *Annals of biomedical engineering*, **23**(4): pp. 359-374.

[33] Zajac, F.E. (1988) Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Critical reviews in biomedical engineering*, **17**(4): pp. 359-411.

[34] Winters, J.M. and Woo, S.L. (1990) *Multiple muscle systems: Biomechanics and Movement Organization*.

[35] Ito, Y., Corna, S., von Brevern, M., Bronstein, A., and Gresty, M. (1997) The functional effectiveness of neck muscle reflexes for head-righting in response to sudden fall. *Experimental brain research*, **117**(2): pp. 266-272.

[36] Iwamoto, M., Nakahira, Y., and Kimpara, H. (2015) Development and validation of the Total HUman Model for Safety (THUMS) toward further understanding of occupant injury mechanisms in precrash and during crash. *Traffic injury prevention*, **16**(sup1): pp. S36-S48.

[37] Günther, M., Schmitt, S., and Wank, V. (2007) High-frequency oscillations as a consequence of neglected serial damping in Hill-type muscle models. *Biological Cybernetics*, **97**(1): pp. 63-79.

[38] Garrett, W.E., Safran, M.R., Seaber, A.V., Glisson, R.R., and Ribbeck, B.M. (1987) Biomechanical comparison of stimulated and nonstimulated skeletal muscle pulled to failure. *The American journal of sports medicine*, **15**(5): pp. 448-454.

[39] Lieber, R.L. and Friden, J. (1993) Muscle damage is not a function of muscle force but active muscle strain. *Journal of Applied Physiology*, **74**(2): pp. 520-526.

[40] Gehre, C., Gades, H., and Wernicke, P. Objective rating of signals using test and simulation responses. *Proceedings of 21st International Technical Conference on the Enhanced Safety of Vehicles Conference (ESV),* 2009.

[41] Brault, J.R., Siegmund, G.P., and Wheeler, J.B. (2000) Cervical muscle response during whiplash: evidence of a lengthening muscle contraction. *Clinical biomechanics*, **15**(6): pp. 426-435.

[42] Beeman, S.M., Kemper, A.R., Madigan, M.L., Franck, C.T., and Loftus, S.C. (2012) Occupant kinematics in low-speed frontal sled tests: Human volunteers, Hybrid III ATD, and PMHS. *Accident Analysis & Prevention*, **47**: pp. 128-139.

[43] Kamibayashi, L.K. and Richmond, F.J. (1998) Morphometry of human neck muscles. *Spine*, **23**(12): pp. 1314-1323.

[44] Hatze, H. (1978) A general myocybernetic control model of skeletal muscle. *Biological cybernetics*, **28**(3): pp. 143-157.