INJURY BIOMECHANICS RESEARCH
Proceedings of the Thirty-Ninth International Workshop

Development of a Detailed Finite Element Neck Model for Automotive Safety Research

J.B. Fice, J.A. Moulton, D.S. Cronin

Department of Mechanical and Mechatronics Engineering, University of Waterloo

ABSTRACT

This study outlines the development and validation of a detailed finite element model of a 50th percentile male cervical spine. This work is part of a larger collaborative effort to produce a full human body model for the automotive community. The goal of this work is to produce a detailed, biofidelic and frangible human body model that will provide a valuable tool in automotive safety research.

The geometry of the vertebrae and muscle tissue in the model were derived from CT scans of a volunteer and the cervical vertebrae were positioned according to available literature for intervertebral disc spacing, the relative positions of vertebral landmarks, and spinal curvature reported for a seated 18-24 year old male. All of the relevant tissues were incorporated in the model, including the vertebrae (cancellous and cortical bone), the discs with representations of the annulus fibrosus, ground substance and nucleus pulposus, the facet joints, and the ligaments. The muscle in the model included 3D solid elements to represent the passive tissue properties, with embedded axial elements using a Hill relationship for the active properties. All of the material properties in the model were derived from the literature, while the ligament properties were measured experimentally at deformation rates relevant to automotive crash scenarios.

The model was validated using a hierarchical approach, beginning at the segment level with physiological loads. Subsequently the full spine was validated using 15g frontal, 7g lateral, and 4g rear impacts. The model was found to generally agree with the cadaver studies at the segment level and the volunteer studies at the full spine level without calibration of the material properties to improve the model response.

INTRODUCTION

Neck injuries resulting from automotive collisions continue to have a significant impact on society in terms of cost and disability. Automotive collisions are the most common source of spinal cord injuries resulting from a loss of stability in the cervical spine (Cusick & Yoganandan, 2002). Neck strain or sprains, often referred to as whiplash, account for 28% of all injuries from automotive collisions, resulting in over 900,000 victims in the United States in 2000 (Quinlan et al., 2004).

The goal of this work was to outline the development and validation of a 50% percentile male cervical spine finite element model, which was part of a larger collaborative effort to produce a full human body model for the automotive safety community. This model will provide automotive designers a detailed, biofidelic, and frangible model of the human neck which can assist in the design of safety systems to help reduce the incidence of neck injuries in automotive collisions.

The geometry for the model came from CT scans of a representative 50% male volunteer, and was assembled using information from the scans and supplemented with literature data to obtain an accurate seated posture. The ligament material properties were experimentally determined (Mattucci et al. 2012), and all other properties were taken from literature and were not calibrated to match a specific dataset (Panzer et al., 2011; Hedenstierna et al., 2008). Validation was carried out in a step-wise fashion, beginning with segments extracted from the full model under physiologic loads, and then full neck model simulations of impact events in frontal, lateral, and rear impact.

METHODS

The finite element model represents a 50th percentile male cervical spine with a simplified head and first thoracic vertebrae for appropriate boundary conditions. The model was developed in LS-DYNA R4.2.1 (LSTC, Livermore, CA), and it was meshed using Hypermesh (Altair, Troy, MI) and LS-PrePost (LSTC, Livermore, CA). The model contains 304,385 elements including 204,180 hexahedral solids, 95,630 shells, and 4,575 1D axial elements.

Geometry and Posture

The CAD geometry was developed from CT scans of a representative 50% percentile male (Gayzik, et al., 2011). The posture of the seated occupant was developed using a combination of the CT scan data and several studies reported in the literature (Klinich et al., 2004, Reed et al., 2002, Lu et al., 1999). The relative positions of the vertebrae were adjusted to fit a Bezier spline curve (with a superior angle of 11.5° and an inferior angle of 9°) through landmarks on the posterior surface of the vertebral bodies as per the technique presented by Klinich et al. (2004). This was supplemented with intervertebral disc spacing reported by Lu et al. (1999). The curvature of the spine was measured using the curvature index technique and compared to literature (Klinich et al. 2004).

Model Assembly

The Neck Model (NM) consisted of seven cervical vertebrae, composite intervertebral discs, detailed facet joints, non-linear rate dependent ligaments, 3D passive muscle, and 1D active muscle (Figure 1). In addition to these main components, the model also incorporated a representation of the spinal cord and a shell structure on the anterior of the neck that represented the volume filled by the trachea, thyroid cartilage, and cricoids cartilage etc. (Figure 1). To simulate the boundary and loading conditions on the neck, the NM included a simplified head with representative inertial properties and the first thoracic vertebrae (T1). The intervertebral discs were constructed with solid elements for the annulus fibrosis ground substance and nucleus pulposus, and layers of shell elements representing the fiber lamina (Figure 2A). The facet joints were modeled with a superior and inferior layer of solid elements for the articular cartilage and a squeeze-film model to simulate the synovial fluid (Figure 2B). Ligaments were represented using multiple 1D nonlinear rate dependent tension-only beam elements (Figure 2C). A total of 26 neck muscles were modeled using solid elements for the passive response with embedded Hill-type axial elements to simulate the active response (Figure 3). The axial active muscle shared nodes with the 3D passive muscle elements, and their relative position was maintained using additional 1D beam elements attached to the vertebrae (Figure 3). All of the material properties in the model were derived from the literature, while the ligament properties were measured experimentally at deformation rates relevant to automotive crash scenarios (Panzer et al., 2011; Hedenstierna et al., 2008; Mattucci et al. 2012).

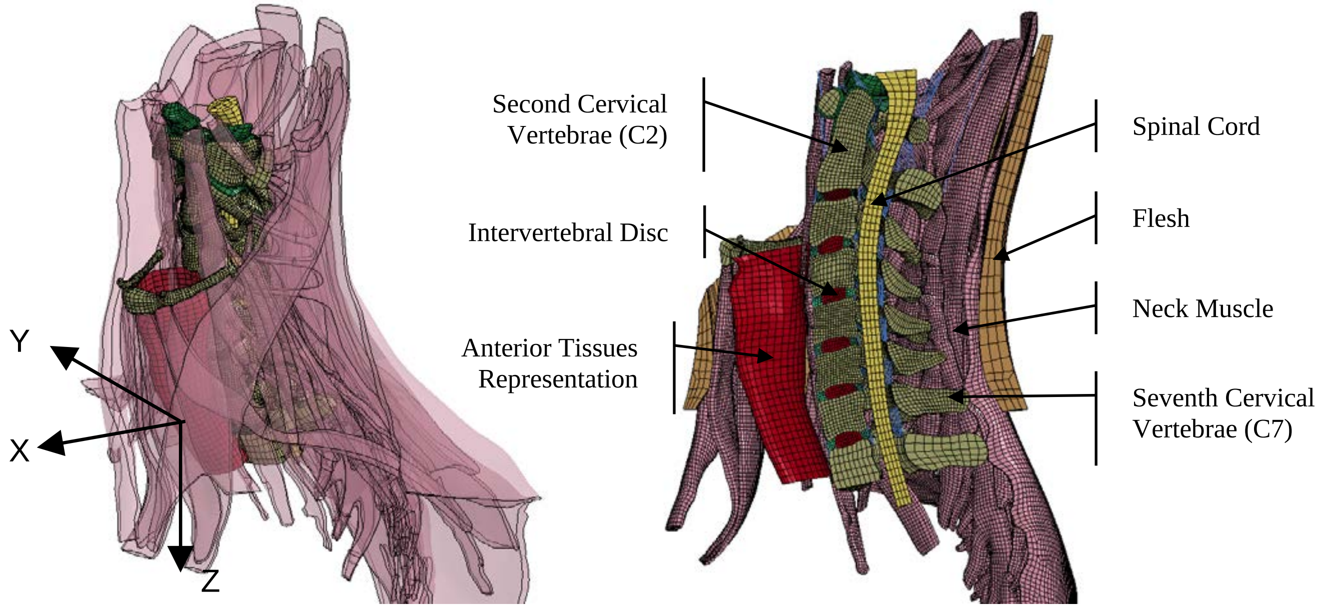


Figure 1: Isometric view of the neck model with the muscle translucent (left) and sagittal plane section (right).

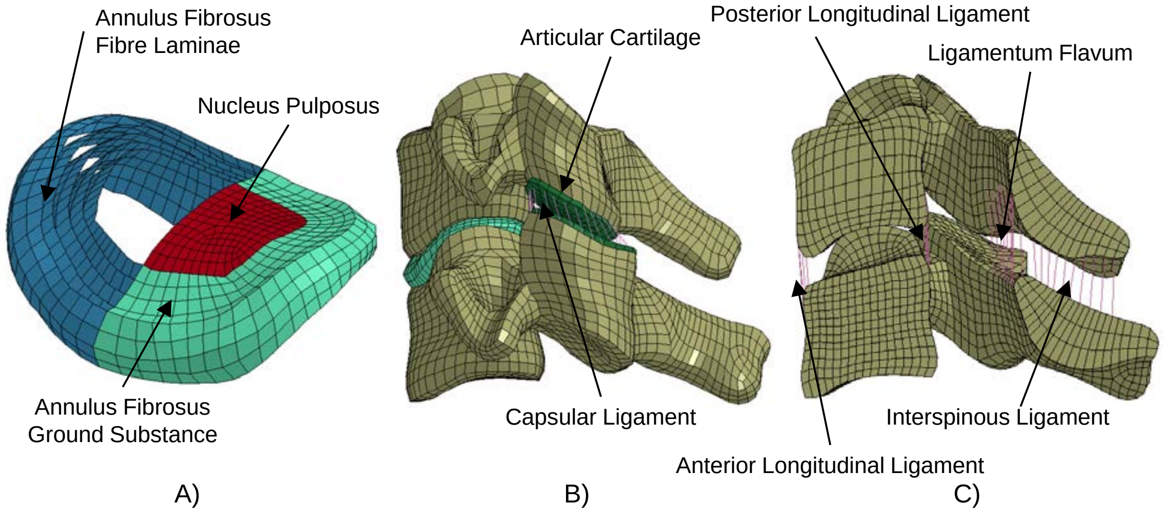
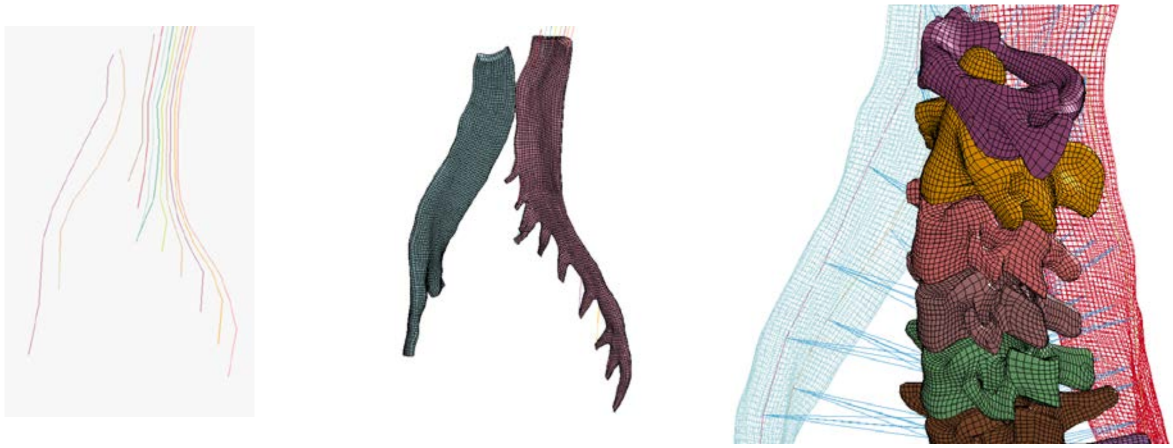


Figure 2: From the C4-C5 segment: A) Disc Sectioned to show the solid elements (right) and the embedded shell elements (left), B) Right view showing the facet joint, C) Sectioned view showing the ligaments.



A) B) C)

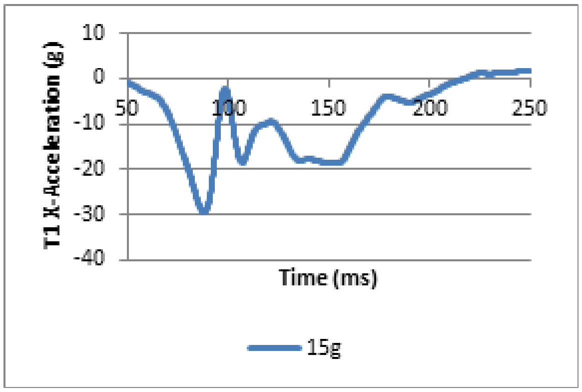
Figure 3: The sternocleidomastoid and the semispinalis capitis are shown: A) Discrete elements are used for active properties, B) Solid volumes for the passive properties, C) Additional elements are used to maintain the relative position of the active muscle elements.

To model volunteers in impact, the flexor and extensor muscles were contracted at a time of 74ms following impact and remained active for 100ms, which was supported by EMG measurements of volunteers in impacts (Siegmund, et al. 2003a). The activation level as a function of time was determined using active state dynamics (Happee et al., 1994), with the peak activation level not exceeding 87.1%. In rear impact, the extensor muscle activation was set to 70% of the flexor muscles (Siegmund, et al. 2003a, Siegmund et al., 2003b; Brault et al., 2000).

Model Validation

The cervical spine model was isolated into functional spinal units (vertebra-disc-vertebra) to validate the model with experimental tests conducted at the same level (Nightingale et al, 2002, Nightingale et al., 2007, Camacho et al., 2007, Wheeldon et al., 2006). Rotational displacements were applied about the center of mass of the superior vertebral body to model the physiological range of motion in flexion and extension. Similarly, the upper cervical spine (C0-C2) was validated in flexion and extension against data by Nightingale et al. (2007) and in axial rotation against data from Goel et al. (1990).

The Naval Biodynamics Laboratory (NBDL) performed a series of 46 frontal impacts on eight volunteers and 31 lateral impacts on nine volunteers (Wismans et al., 1986; Thunnissen et al., 1995). The NBDL tests were simulated by applying the experimental T1 motions in the X, Z, and Y-rotational directions to the T1 of the neck model (Figure 4). The T1 was constrained in all other directions and the head was free. Muscle activation was included to model the live volunteers. The resulting linear and angular accelerations were measured at the C.G. of the head and compared to response corridors representing the average volunteer response plus and minus a standard deviation.



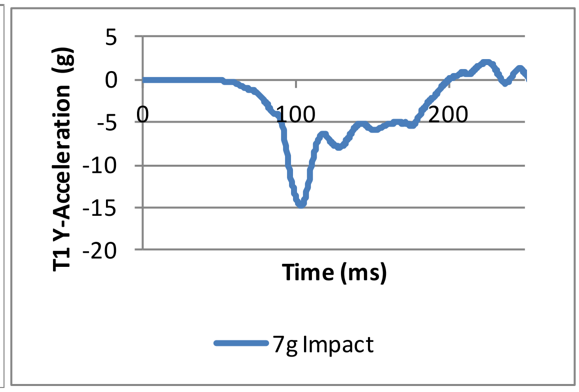


Figure 4: T1 inputs for NBDL 15g frontal impact (left) and 7g lateral impact (right).

Davidsson et al. (1998) performed 28 rear impacts on thirteen human volunteers at speeds between 5 and 7kph with an average peak acceleration of 3.6g. To model these rear impacts, the average T1 accelerations in the X and Z directions, along with the Y rotation were input onto the cervical spine model T1 as prescribed motion constraints (Figure 5). The T1 of the model was constrained in all other directions, and the head was not constrained. To model the headrest, the average sled anterior acceleration was input as prescribed motion. The headrest material properties were based on automotive seat cushion material, and the stiff backing was assumed to be pine (Green et al., 1999; Campbell & Cronin, 2007). The mass of the headrest and stiffness of the springs were input using values provided by Davidsson et al. (1998) and headrest was initially positioned 86mm away from the skull, which was the average value used in the experiments. Muscle activation was included to mimic the behavior of volunteers. The global kinematic response of the model was compared to corridors presented by Hynd et al. (2007), which represent the average volunteer response plus and minus one standard deviation.

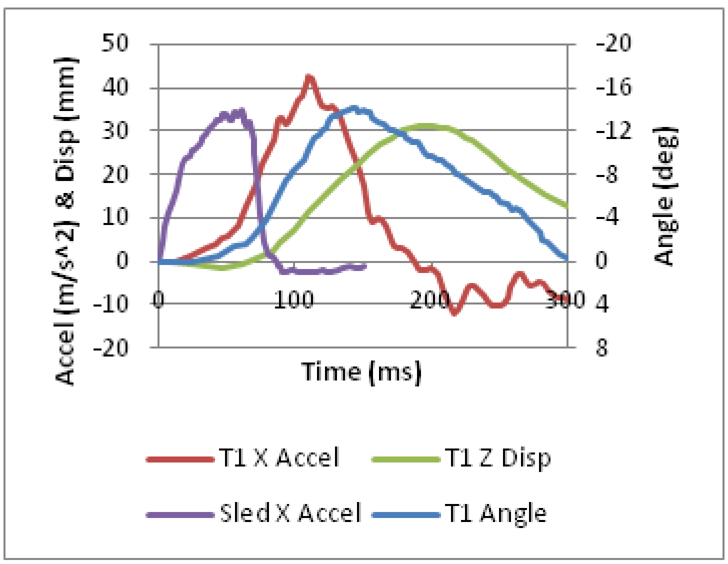


Figure 5: T1 and headrest inputs for the 4g rear impact simulation.

RESULTS

The curvature index in the model was measured to be 0.8%, which is within the low end range of the experimental data for a young (18-24 years) or mid aged (35-44 years) male (Figure 6). The cord angle measured from the inferior portion of the C7 vertebrae to the superior surface of the odontoid process was found to be 92° which compares well to 91° reported in a study of the seated posture by Reed et al., (2002). The disc geometry resulting from the vertebral body dimensions and positioning were within a standard deviation of the experimental average at each vertebral level (Figure 7). The resulting cervical spine column posture was in good agreement with the literature for a seated occupant and with the scan data produced by Gayzik et al., 2011.

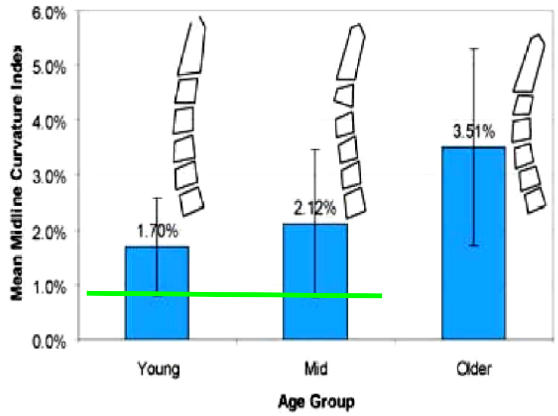


Figure 6: Green line shows the curvature index for the model compared to data from Klinich et al., 2004.

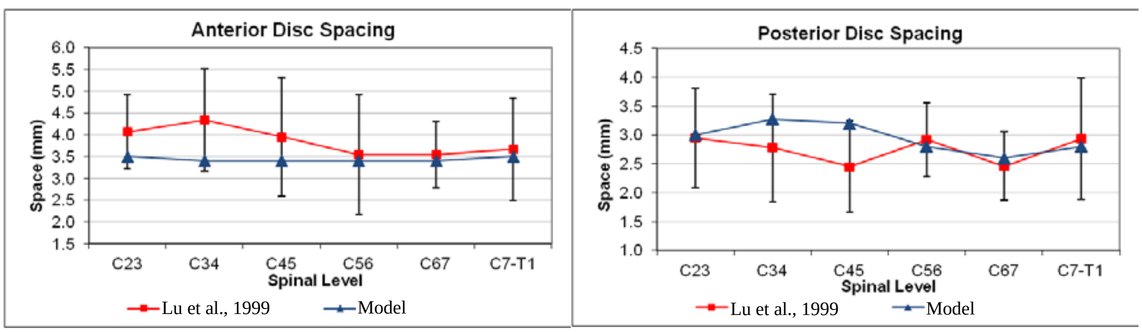


Figure 7: Anterior and posterior disc spacing for the intervertebral disc of the model compared to data from Lu et al., 1999.

The extracted segment models exposed to extension moments showed favourable agreement with values reported in the literature over the range of physiologic moments tested (Figure 8). The model was shown to be generally stiffer when compared to the experiments, but well within a standard deviation of the averages for the C23 and C7T1 segments. In flexion, each of the segments simulated showed a close agreement with one of the experimental studies, but there was more variation of the experimental averages between experiments for the flexion tests (Figure 9). The upper cervical spine segment simulations, which consisted of the occipital condyles, C1, and C2, showed a close agreement with the experimental studies up to approximately 2Nm in flexion, extension and rotation, but were lax for higher applied moments (Figure 10).

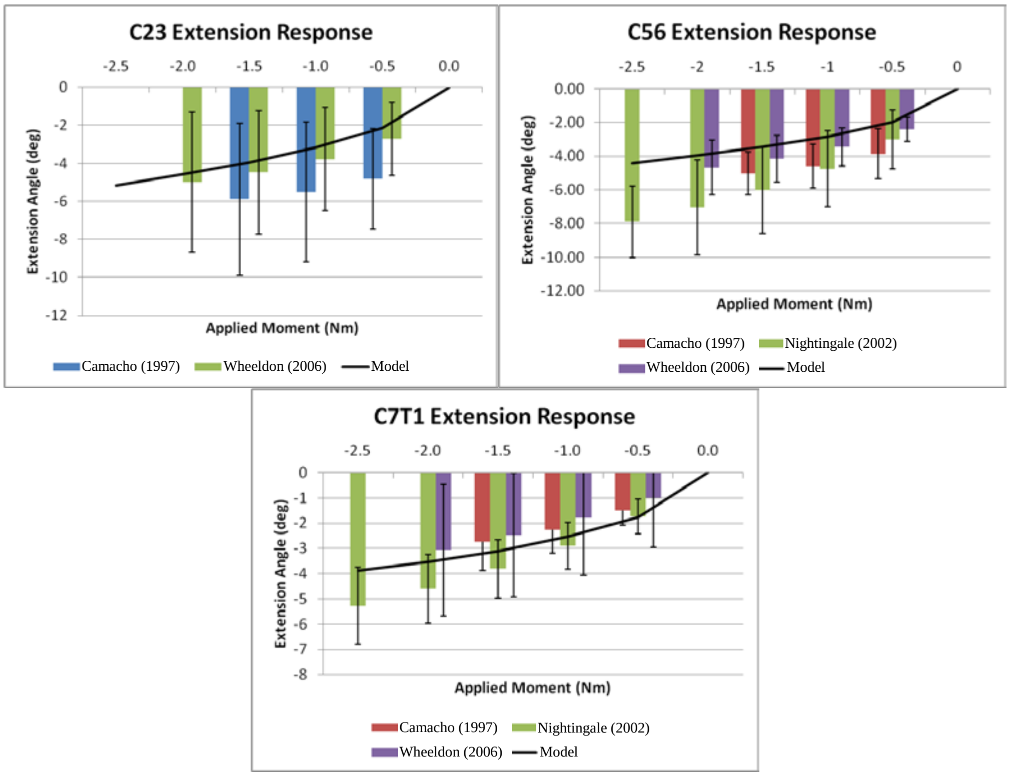


Figure 8: Extracted vertebrae - disc - vertebrae segment responses to extension moments.

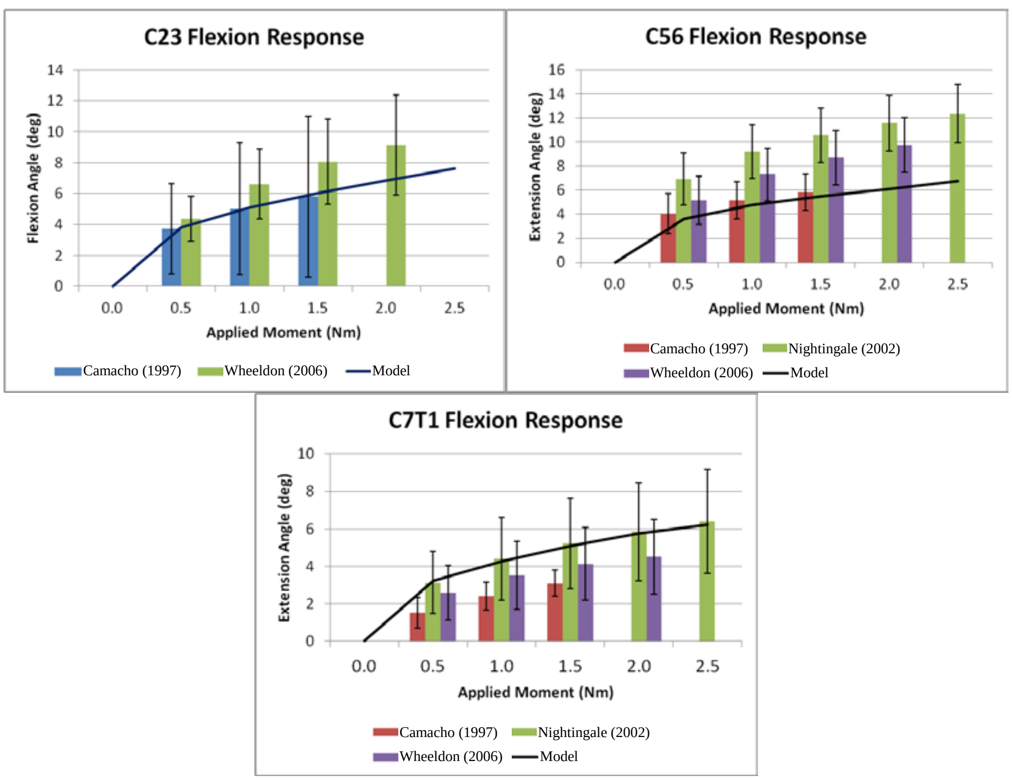


Figure 9: Extracted vertebrae - disc - vertebrae segment responses to extension moments.

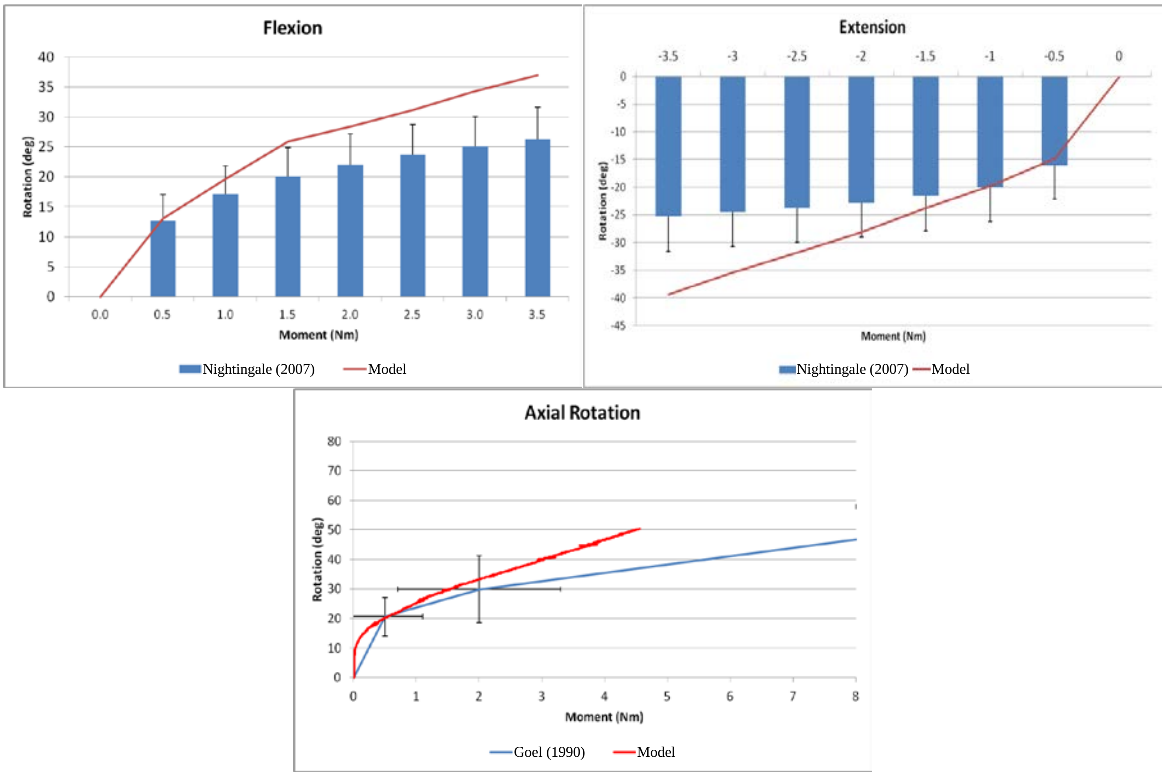


Figure 10: Occipital Condyles - C1 - C2 segment response in flexion, extension, & axial rotation.

The response of the full neck model in a frontal 15g impact fell mostly within one standard deviation of the average volunteer response (Figure 11). The neck model exhibits a representative shape and appropriate peak values, but it experiences additional oscillations when compared to the volunteers. In a 7g lateral impact, the neck model response was a good match to the volunteer data (Figure 12). The peak lateral acceleration of the head overshot the response corridor (plus and minus a standard deviation of the mean) by 1g, but the timing of the response was very close to the volunteers. The peak rotational acceleration and the fore-aft acceleration of the head were both within a standard deviation of the average response, but the peaks occurred approximately 10 to 20ms early. In rear impact, the fore-aft displacement of the head fell within a standard deviation of the volunteer average except for a slight dip between 100 and 150ms (Figure 13). Both the head rotation and occipital condyle upwards displacement with respect to the T1 were exaggerated between 85ms to 200ms when compared to the volunteer response (Figure 13).

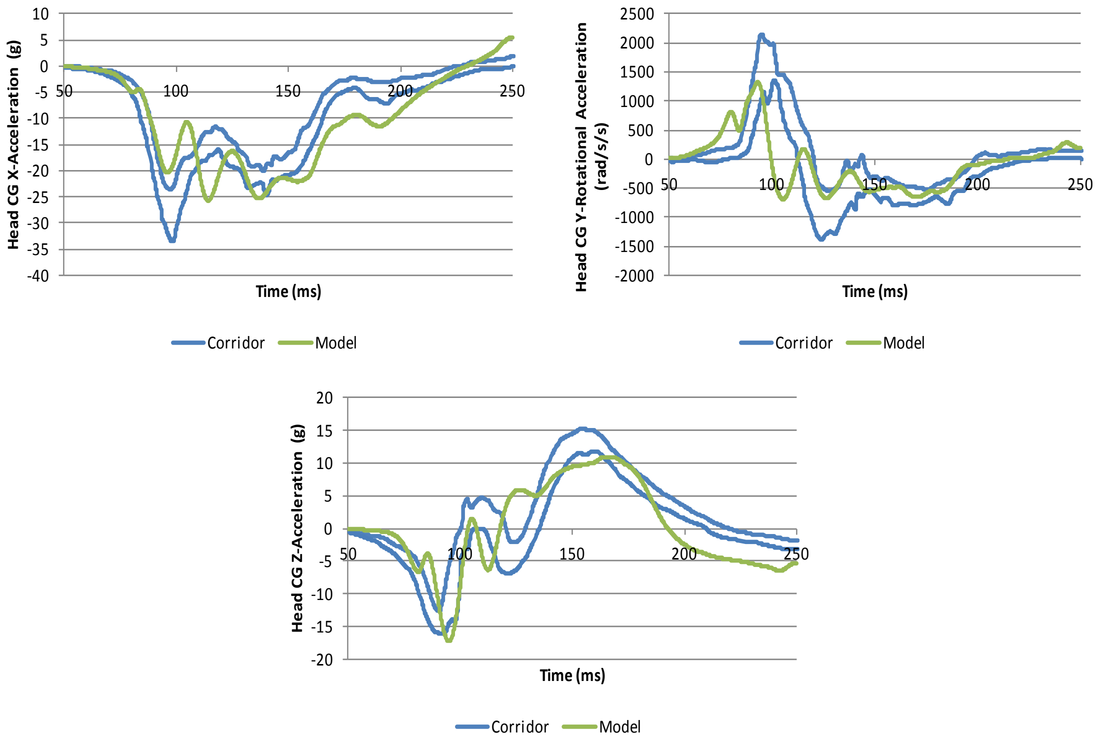


Figure 11: Full neck model response to NBDL 15g frontal impact.

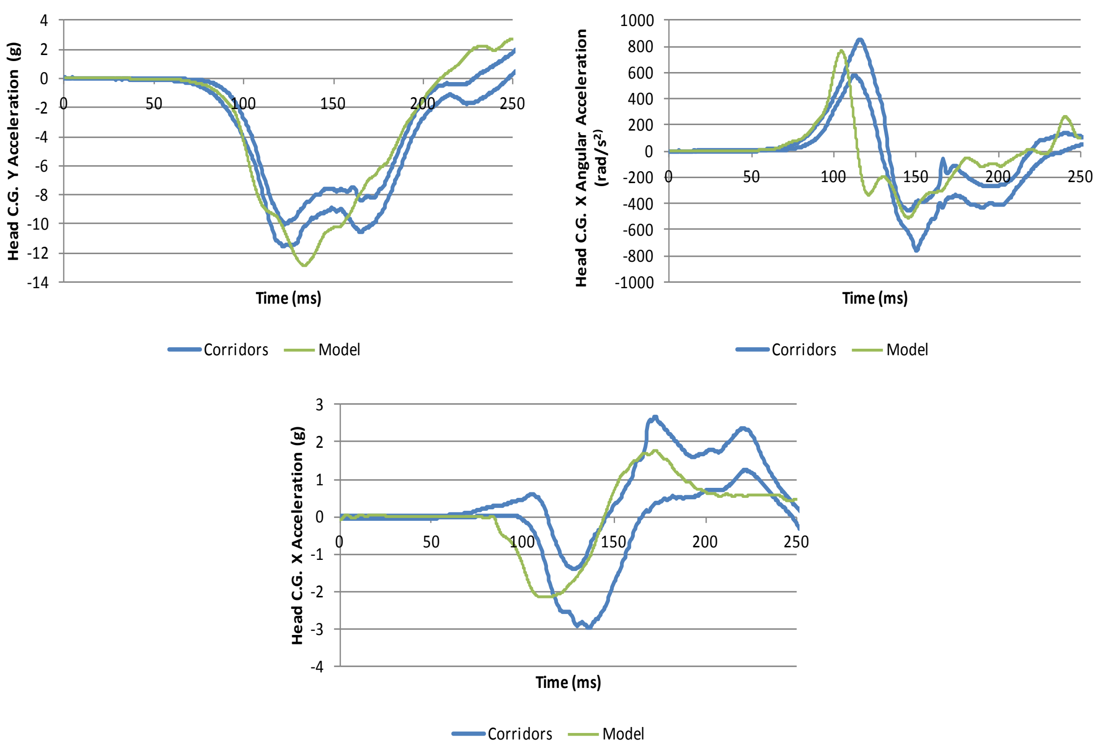


Figure 12: Full neck model response to NBDL 7g lateral impact.

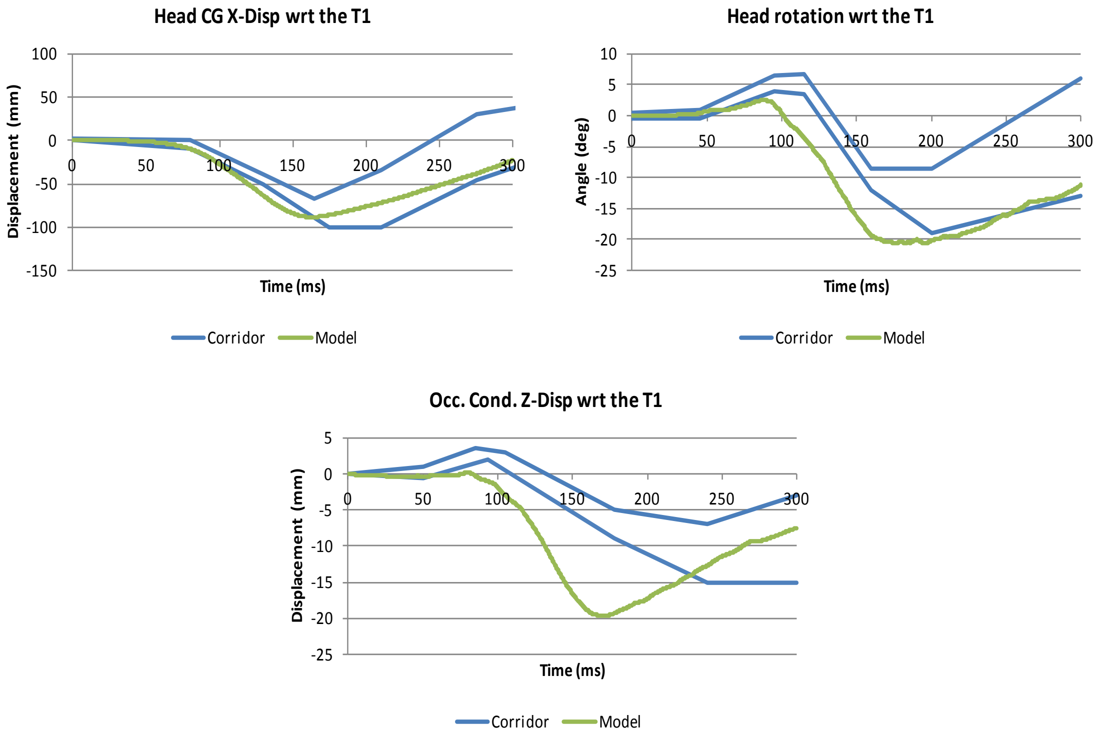


Figure 13: Full neck model response to Davidsson et al., (1998) 4g rear impact.

CONCLUSIONS

The development of a detailed 50th percentile male cervical spine model was presented. The model was validated at the segment level, assembled into a full neck model, and validated against a range of impact cases without calibration of the material properties. Ongoing work will include the implementation of tissue damage and injury prediction.

ACKNOWLEDGEMENTS

The authors gratefully acknowledge support from the Global Human Body Models Consortium.

REFERENCES

CAMACHO, D.L.A., NIGHTINGALE, R.W., ROBINETTE, J.J., VANGURI, S.K., COATES, D.J., AND MYERS, B.S., 1997. Experimental Flexibility Measurements for the Development of a Computational Head-Neck Model Validated for Near-Vertex Head Impact. Proceedings from the 41st Stapp Car Crash Conference, 473 – 486. SAE 973345.

CUSICK JF, YOGANANDAN N., Biomechanics of the cervical spine 4 major injuries. Clinical Biomechanics 2002;17(1):1–20.

DAVIDSSON, J., DEUTSCHER, C., HELL, W., LINDER, A., LOVSUND, P., SVENSSON, M., ET AL. (1998). Human volunteer kinematics in rear-end sled collisions. In Proceedings of IRCOBI (pp. 289-301). Goteborg.

DEWIT, J. A., AND CRONIN, D.S., 2010. Cervical spine segment finite element model validation and verification at traumatic loading levels for injury prediction. In Proceeding of IRCOBI. Hanover, Germany.

FICE , J. B., CRONIN, D.S., PANZER, M.B. 2011. Cerivcal model to predict capsular ligament response in rear impact. Annals of Biomedical Engineering 39(8), 2152-62.

GAYZIK, F.S., MORENO, D.P., GEER, C.P., WUERTZER, S.D., MARTIN, R.S., STITZEL, J.D., 2011. Development of a Full Body CAD Dataset for Computational Modeling: A Multi-modality Approach. Annals of Biomedical Engineering 39(10), 2568-83.

GOEL, V.K., WINTERBOTTOM, J.M., SCHULTE, K., CHANG, H., GILBERTSON, L.G., PUDGIL, A.G., GWON, J.K., 1990. Ligamentous laxity across C0-C1-C2 complex: Axial torqure-rotation characteristics until failure. Spine 15(10), 990-996.

HEDENSTIERNA, S., HALLDIN, P., BROLIN, K., 2008. Evaluation of a Combination of Continuum and Truss Finite Elements in a Model of Passive and Active Muscle Tissue. Computer Methods in Biomechanics and Biomedical Engineering 11(6), 627-639.

KLINICH, K.D., EBERT, S.M., VAN EE, C.A., FLANNAGAN, C.A.C., PRASAD, M., REED, M.P., AND SCHNEIDER, L.W., 2004. Cervical Spine Geometry in the Automotive Seated Posture: Variations with Age, Stature, and Gender. Proceedings from the 48th Stapp Car Crash Conference, 301 – 330. SAE 2004-22-014.

LU, J., EBRAHEIM, N.A., YANG, H., ROLLINS, J., YEASTING, R.A., 1999. Anatomic bases for anterior spinal surgery: surgical anatomy of the cervical vertebral body and disc space. Journal of Clinical Anatomy 21, 235-239.

MATTUCCI, S., MOULTON, J. CHANDRASHEKAR, N., CRONIN, D.S., “Strain Rate Dependent Properties of Younger Human Cervical Spine Ligaments”, Journal of the Mechanical Behavior of Biomedical Materials, Accepted Feb 8, 2012.

MATTUCCI, S., 2011. Strain Rate Dependent Properties of Younger Human Cervical Spine Ligaments. MS Thesis, University of Waterloo.

NIGHTINGALE, R.W., WINKELSTEIN, B.A., KNAUB, K.E., RICHARDSON, W.J., LUCK, J.F., AND MYERS, B.S., 2002. Comparative Strengths and Structural Properties of the Upper and Lower Cervical Spine in Flexion and Extension. Journal of Biomechanics 35, 725 – 732.

NIGHTINGALE, R.W., CHANCEY, V.C., OTTAVIANO, D., LUCK, J.F., TRAN, L., PRANGE, M., MYERS, B.S., 2007. Flexion and extension structural properties and strengths for male cervical spine segments. Journal of Biomechanics 40, 535-542.

QUINLAN KP, ANNEST JL, MYERS BS, RYAN G, HILL H. Neck strains and sprains among motor vehicle occupants – United States, 2000. Accident Analysis & Prevention, 2004;36(1):21–7.

REED, M.P., MANARY, M.A., FLANNAGAN, C.A.C., SCHNEIDER, L.W., 2002. A statistical method for predicting automobile driving posture. Human Factors 44 (4), 557-568.

ROBBINS, D.H., 1983. Anthropometric specifications for mid-sized male dummy, volume 2. University of Michigan Transportation Research Institute, UMTRI-83-53-2.

SIEGMUND, G. P., SANDERSON, D. J., MYERS, B. S., & INGLIS, J. T., 2003a. Rapid neck muscle adaptation alters the head kinematics of aware and unaware subjects undergoing multiple whiplash-like perturbations. Journal of biomechanics, 36(4), 473-82.

SIEGMUND, G. P., SANDERSON, D. J., MYERS, B. S., & INGLIS, J. T., 2003b. Awareness affects the response of human subjects exposed to a single whiplash-like perturbation. Spine, 28(7), 671-9.Brault, J. R., Siegmund, G. P., & Wheeler, J. B., 2000. Cervical muscle response during whiplash: evidence of a lengthening muscle contraction. Clinical biomechanics, 15(6), 426-35.

THUNNISSEN, J., WISMANS, J., EWING, C.L., AND THOMAS, D.J., 1995. Human Volunteer Head-Neck Response in Frontal Flexion: A New Analysis. Proceedings from the 39th Stapp Car Crash Conference 439 – 460. SAE 952721.

VAN DER HORST, M.J., 2002. Human Head Neck Response in Frontal, Lateral and Rear End Impact. PhD Thesis, Eindhoven University of Technology.

WHEELDON, J.A., PINTAR, F.A., KNOWLES, S., AND YOGANANDAN, N., 2006. Experimental Flexion/Extension Data Corridors for Validation of Finite Element Models of the Young, Normal Cervical Spine. Journal of Biomechanics 39, 375 – 380.

WISMANS, J., VAN OORASHOT, H., AND WOLTRING, H.J., 1986. Omni-Directional Human Head-Neck Response. Proceedings from the 30th Stapp Car Crash Conference 313 – 331. SAE 861893.