A Biofidelic Surrogate Neck for Axial Impacts

[1]Tim S. Nelson and [1]Peter A. Cripton
[1]Injury Biomechanics Laboratory, Departments of Mechanical Engineering and Orthopaedics
and International Collaboration On Repair Discoveries, University of British Columbia
Vancouver, Canada

ABSTRACT

Axial compressive neck injuries occurring during head-first impacts in sports and transportation accidents devastate the lives of those affected and their families. We desired a physical surrogate head and neck to provide a sufficiently biofidelic and repeatable response to head-first impacts for the purposes of evaluating injury prevention strategies and testing the efficacy of devices intended to prevent or mitigate injury. A surrogate neck based upon biomechanical concepts specifically for head-first impacts has been designed and built for use with a surrogate head and custom drop tower already developed and being used in our lab. The design allows for both sagittal rotation and compression between adjacent vertebrae in the sagittal plane. Vertebrae are constrained to rotate about centers of rotation typically located on an adjacent inferior vertebrae. A spring loaded preload mechanism utilizing 4 cables, 2 on each lateral side, applies preload along the centers of rotation for each vertebral level. In this study, it was our objective to subject the surrogate head and neck to a variety of baseline mechanical tests. Flexion-extension rotation response testing and a series of head-first impacts were performed with and without a guided preload system. Full surrogate spine (C0-T1) flexion-extension flexibility tests have been performed on a custom spine machine at three preloads (0,78, and 104 N) that can apply pure dynamic moments at controlled loading rates. 12 impacts onto a rigid perpendicular surface were conducted, 6 with preload and 6 without. In both flexibility and drop testing, kinematics were determined by tracking markers and planar photogrammetry. Drop testing showed a repeatable and realistic decoupled response between head and neck loading. Significant differences ($\alpha=.05$) were found to exist for peak neck load, neck impulse duration, peak head load, initial head impulse magnitude, and the time lag between head and neck loading. Kinematics in drop testing were very repeatable with and without preload although the kinematic patterns and posture prior to impact were quite different. Flexibility testing showed a highly non linear flexion-extension response with a large neutral zone. The range of motion (ROM) was considerably smaller with each incremental preload magnitude tested. Subaxial (C2-C7) and intersegmental ROM without a preload were within an acceptable tolerance of published data for in vitro and in vivo data.

1

INTRODUCTION

Axial compressive neck injuries occurring during head-first impacts in sports and transportation accidents devastate the lives of those affected and their families. Head first impacts have been studied by many researchers with cadaveric specimens. The wide variability among specimens has led to a vast array of different injuries produced from impacts with very similar impact velocities and head and neck alignments with the impact surface. A repeatable surrogate head and neck would remove the cost, preparation time, ethical issues, and inherent variability associated with testing biological specimens allowing a much larger number of tests to be performed. We desired a physical surrogate head and neck to provide a sufficiently biofidelic and repeatable response to head-first impacts for the purposes of evaluating injury prevention strategies and testing the efficacy of devices intended to prevent or mitigate injury such as novel roof mounted air bags in automobiles. Although there are many commercially available anthropometric head and neck assemblies, to the authors' knowledge, none have been designed specifically for studying head-first impacts. As such, a surrogate neck specifically for head-first impacts has been designed and built for use with a surrogate head and custom drop tower already developed and being used in our lab. In this study it was our objective to design and build a suitable neck and subject it to a variety of baseline mechanical tests. Some of the features of the design will be highlighted and the results of the evaluation studies presented.

METHODS

Design Features

The human cervical spine is an amazingly complex structure. It supports the head and allows a wide range of motion through flexion-extension, axial rotation, and lateral bending. It must resist muscle and inertial loads and allow for the wide range of motion while protecting the delicate spinal cord. Its geometry and material properties combine to provide highly nonlinear responses to loading(Nightingale, Myers et al. 1991; Goertzen, Lane et al. 2004). It also demonstrates motion coupling between axial rotation and lateral bending. (Milne 1991) As such, the engineering challenges for this surrogate neck were in making the appropriate simplifications to allow construction while still providing a realistic response. Figure 1 shows the surrogate neck from two directions. The 'vertebra' attached to the head represents the occiput or C0. The protruding bolts serve two purposes: to guide the cables for a compressive pre-load mechanism and to locate a superior vertebra's center of rotation on a point on the inferior adjacent vertebra. At the remaining cervical spine levels (C1-T1), slots at the bolts allowed for both compression and sagittal rotation. At the upper cervical spine only rotation was allowed. Rubber serrated sheets were stacked offset as 'discs' to provide a nonlinear stiffness for both pure compression and flexion-extension rotation resistance.

Anatomy No attempt was made to replicate exact anatomy. Instead a sagittal-plane-only phenomenological model based upon anatomic dimensions and mechanical response to loading was envisaged. The anatomy of the vertebrae have been simplified into stacked aluminum vertebra separated by simulated discs. The justification for this is that the cervical spine's

response to head-first impacts without significant lateral bending or axial rotation at impact is predominantly in the sagittal plane.(Pintar, Yoganandan et al. 1989) From the perspective of developing injury prevention measures, the lack of lordosis and simplified geometry represents a worst case scenario for developing high axial compressive neck loads.(Torg, Guille et al. 2002)

Axial Stiffness The axial stiffness of the cervical spine is highly non linear. It has a very low initial stiffness that rapidly increases with further displacement. This low initial stiffness of the neck has been attributed to a bimodal head and neck response upon dynamic axial impact (Nightingale, McElhaney et al. 1996). The largest head loads are associated with the initial stopping of the head, while the torso is still moving towards the ground. Due to the low initial neck stiffness, there is a lag between head and neck load development.

Range of Motion It was desired that the surrogate neck match the range of motion observed in human beings. Several studies of both clinical and in vitro nature were sourced for this data. (Dvorak, Panjabi et al. 1991) It was intended that both the overall C0-T1 and intervertebral rotations be matched. There is much variability among people and throughout the range of motion literature. Thus a representative range close to the *in vivo* average response was chosen.

Center of Rotation The segmental mean centers of rotation were identified relative to specific markers on the posterior margin of the vertebral body.(Dvorak, Panjabi et al. 1991) Anatomical references were consulted(Francis 1955; Katz, Reynolds et al. 1975; Kandziora, Pflugmacher et al. 2001; Dong, Xia Hong et al. 2003) to determine quantitative anatomy for human vertebrae to determine the numerical placement point along an average sized vertebral body at each vertebral level. Each vertebra has an elongated "C" shape when viewed in the coronal plane. The short sides of the "C" protrude distally and contain a slot in which a bolt passes through locating the center of rotation. The slot allows for pure compression of each vertebra while still imposing the sagittal center of rotation location.

Flexion-Extension Stiffness The cervical spine has been shown to be stiffer in extension than flexion.(Camacho, Nightingale et al. 1997) The ratio between extension and flexion stiffnesses and the absolute stiffnesses were used as design guides in order to achieve biofidelic sagittal plane motion under quasi-static loading. Flexibility testing performed on cadaveric head and complete cervical spine specimens (Head to T2) was reference for this purpose.(Camacho, Nightingale et al. 1997) The slopes of the loading curves were estimated allowing the determination of the ratio of extension to flexion stiffness at each intervertebral level. This information combined with the center of rotation and vertebral body information discussed earlier to determine the sagittal dimension posterior to the center of rotation for each vertebra.

Follower Load The follower load concept was developed to successfully simulate the overall compressive effect of musculature in the lumbar and cervical spine. (Patwardhan, Meade et al. 1998; Patwardhan, Havey et al. 2000). Prior to this work, researchers had been unable to load *in vitro* specimens to compressive levels known to exist *in vivo* without buckling occurring. The follower load is applied through guides located at the segmental centers of rotation. Without

the guides, a vertical load as low as 10 N was shown to cause buckling using *in vitro* cadaveric cervical spine specimens.(Panjabi, Cholewicki et al. 1998) The amount of compression due to muscle forces and head weight has been estimated to range from approximately 75 N in a neutral posture up to 155 N in extension in the passive spine (Miura, Panjabi et al. 2002). A mathematical model predicted compressive loads as high as 1200 N in maximal muscle contraction efforts.(Moroney, Schultz et al. 1988).

A spring mounted follower load system was developed that applies a compressive follower load via 4 cables, 2 on each lateral side that criss-cross over bolts that locate the centers of rotation for each segment. Also, during flexion or extension rotation, the spring system increases the compressive load somewhat to simulate the postural effect observed clinically.(Hattori, Oda et al. 1981)

Evaluation Experiments

To assess the performance and repeatability of the design several tests have been performed. In particular, flexion-extension flexibility testing and a series of drop tests have been performed. Figure 1 shows the testing configurations.

Flexibility testing. Full surrogate spine (C0-T1) flexion-extension flexibility tests have been performed on a custom spine machine that can apply pure dynamic moments at controlled loading rates.(Goertzen, Lane et al. 2004) The surrogate spine was fixed to the table at the caudal end and moments were applied to T1. A load cell at the arm measured motor torque and was sampled at 20 Hz. 1 High speed camera (Phantom V9) aligned perpendicular to the lateral view of the surrogate spine recorded the experiment at a resolution of 1440 x 1080 pixels a 20 fps synchronously with the load cell. A protocol consisting of 6 degrees/second with a torque limit of 6 Nm and motor rotation limit of 100 degrees was selected. This protocol was applied rotation with 0, 78, and 104 N follower loads. Intervertebral angles at each segment and the relative rotation between C2-C7 in the sagittal plane were determined through planar photogrammetry.

Drop Testing. 12 drop tests were performed from a drop height of 0.5m onto a rigid perpendicular surface, 6 in the presence (104 N) and 6 without a follower load. The speed at impact was consistently near 3 m/s which has been shown to be the impact speed at which compressive cervical spine injuries develop in diving accidents.(McElhaney, Snyder et al. 1979) Strain guage based uniaxial load cells (Omega LC 402-5K) measured axial force under the impact platform and at T1 at 255 kHz. This data was low pass filtered through a 4th order Butterworth filter. 2 high speed cameras (Phantom V9) captured video at 1000 fps. One camera was located perpendicular to the impact and the other slightly oblique. This allowed for 2D or 3D photogrammetry for kinematic calculations. Markers on the head and vertebrae were tracked with special software (TEMA Lite) and the data smoothed using a 4th order Butterworth filter. In this paper, only 2D kinematics are reported. Descriptive statistics and a matched pairs t-test analysis was performed using a statistical software package (SPSS, V13 Student ed.)
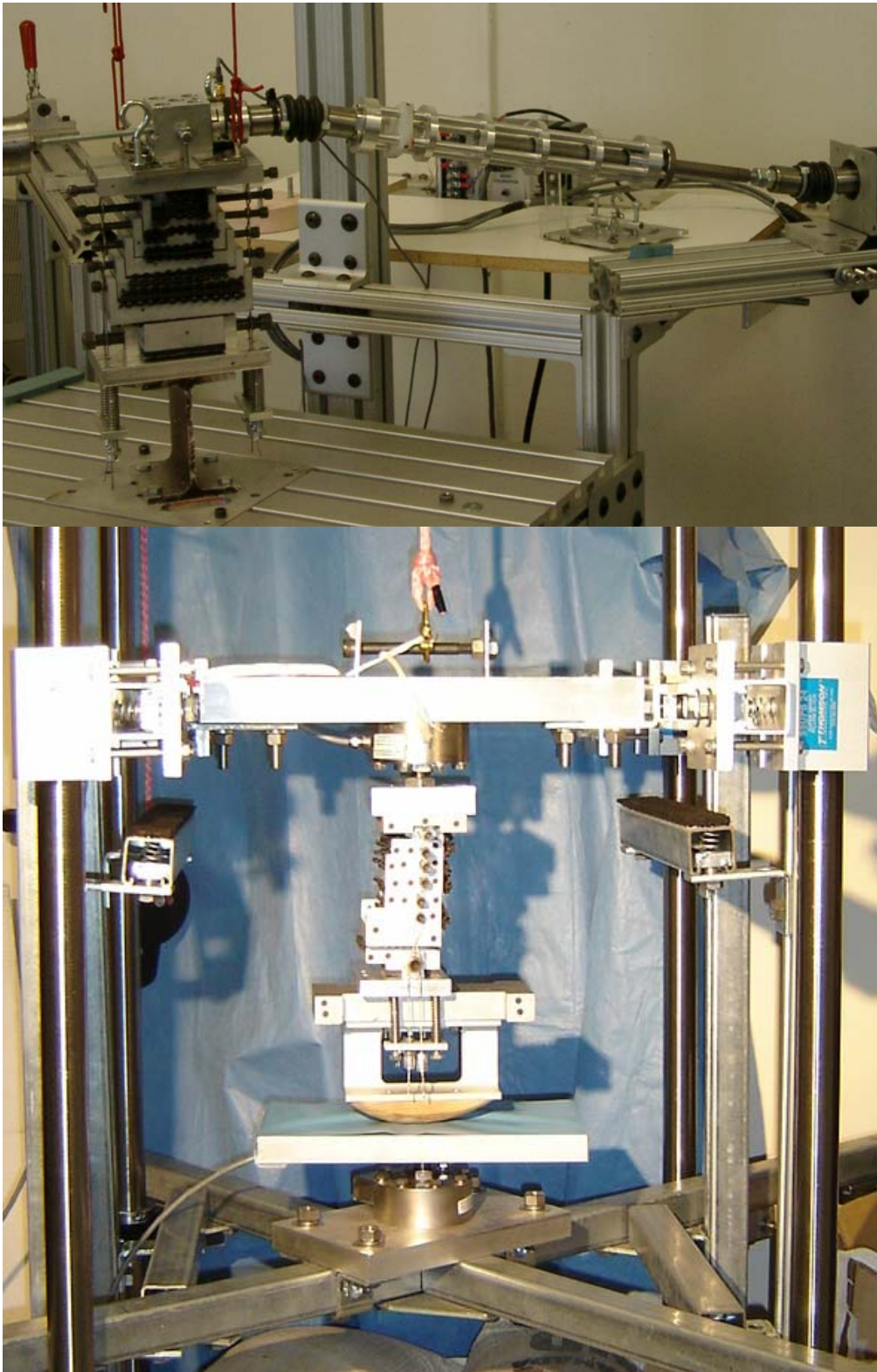
Figure 1. Top: Surrogate neck in spine machine. Bottom: Surrogate head and neck in drop tower.

RESULTS

Head-First Drop Tests

Drop testing revealed excellent repeatability for both kinematics and kinetics for both of the series of drops with and without a 104 N follower load. The head and neck demonstrate a realistic decoupled response in load development due to the low initial axial stiffness as shown in Figure 2.
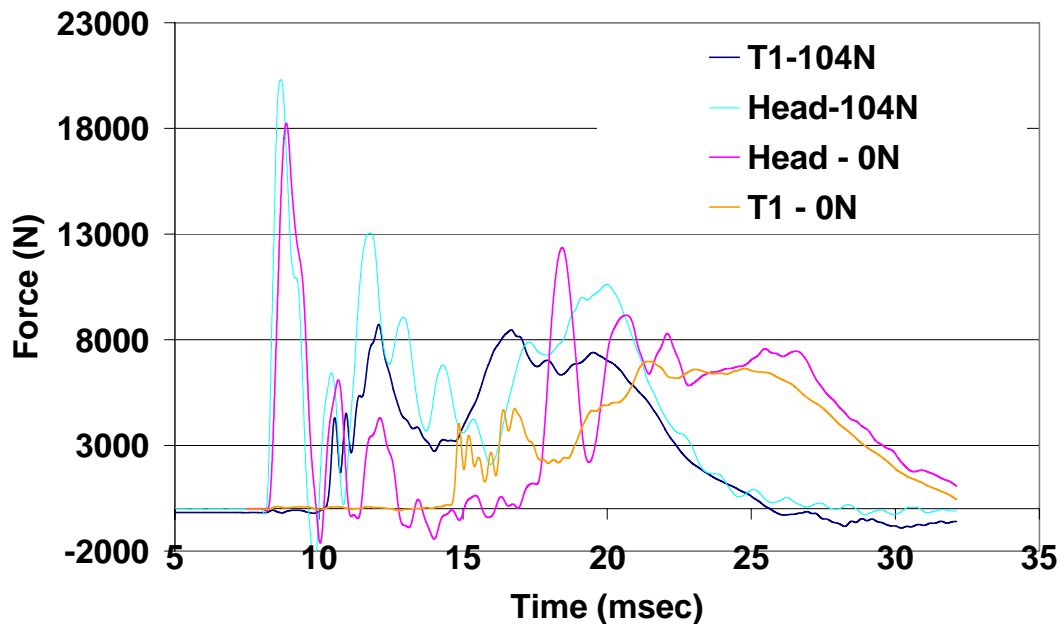
Head & Neck Force vs. Time

Figure 2: Two Impacts, one with 104 N follower load and one without.

The presence or absence of a 104 N follower load had no effect upon neck impulse magnitude but strongly affected impulse duration, peak neck force, and the time lag between head contact and neck load development (Table 1). The tests also show a significant difference for peak head load and initial head impulse with or without a follower load. Referring to figure 2, the effect of pre-load can be seen on two typical impacts. In the absence of follower load, the vertebrae can compress farther before they "bottom out" the low stiffness region of the rubber vertebrae. This has the effect of delaying the onset of neck load after head contact is made. In both cases, the first head impulse is associated with stopping of the head, and the 2nd impulse to the head is coupled with the neck loading and is a result of the momentum of the torso being transferred through the neck to produce head loading.

6

Table 1: Mean Impact Parameters ± 1 Standard Deviation (* indicates p<.05)

Follower Load (N)	Neck Impulse (Ns)	Neck Impulse duration* (ms)	Peak Neck Load* (N)	Head-Neck Time Lag* (ms)	Initial Head Impulse* (Ns)	Peak Head Load* (N)
0	72.0 ± 1.3	18.7 ± 1.2	6984.5 ± 94.7	5.8 ± 0.4	15.5 ± 0.5	17631.6 ± 531.1
104	72.0 ± 0.3	15.7 ± 0.2	8451.7 ± 197.0	2.2 ± 0.2	16.2 ± 0.2	19478.2 ± 983.0

The intervertebral angle patterns were very similar for subsequent impacts in either follower load condition. However, the pattern of motion was considerably different for each follower load condition. There was more rotation after impact in the absence of follower load. However, the postures before impact were different for the two follower loads as the surrogate head/neck was inverted and hanging prior to impact. Gravitational forces/moments that influenced posture were being resisted by the presence of the follower load. In neither follower load condition was there any significant rotation (greater than 6 degrees) during the impact.
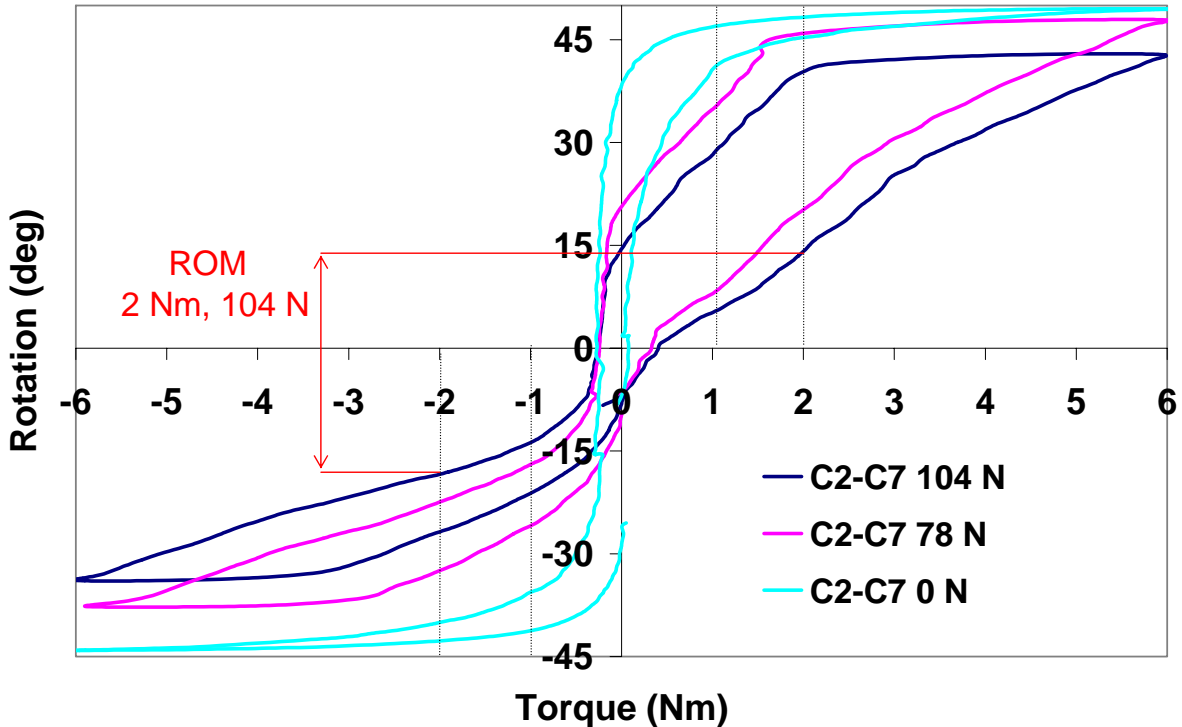
Flexibility Testing

Figure 3: C2 to C7 rotation for 3 follower loads during flexibility testing. A graphical definition of ROM is shown for the 104 N follower load at 2 Nm of applied motor torque.

The ROM in flexibility testing was found to be considerably lower with each incremental follower load applied. This is inconsistent with the theory behind the development of a follower load and with studies using subaxial (C2-T1) cadaveric spines with 0, 50, 100, and 150N of follower load applied (Cripton P.A., personal communication). If the compression is being applied directly through the center of rotation there should be no moments or shear forces that would affect ROM.

Subaxial ROM data (C2-C7) for 1 and 2 Nm moments with a follower load were not in good agreement with published follower load *in vitro* studies utilizing a 100 N follower load nor *in vivo* averages.(Miura, Panjabi et al. 2002) ROM were acceptably on the high side without a follower load but much too small with a 78 or 104 N follower load. Figure 4 shows the subaxial results compared to those by Miura.
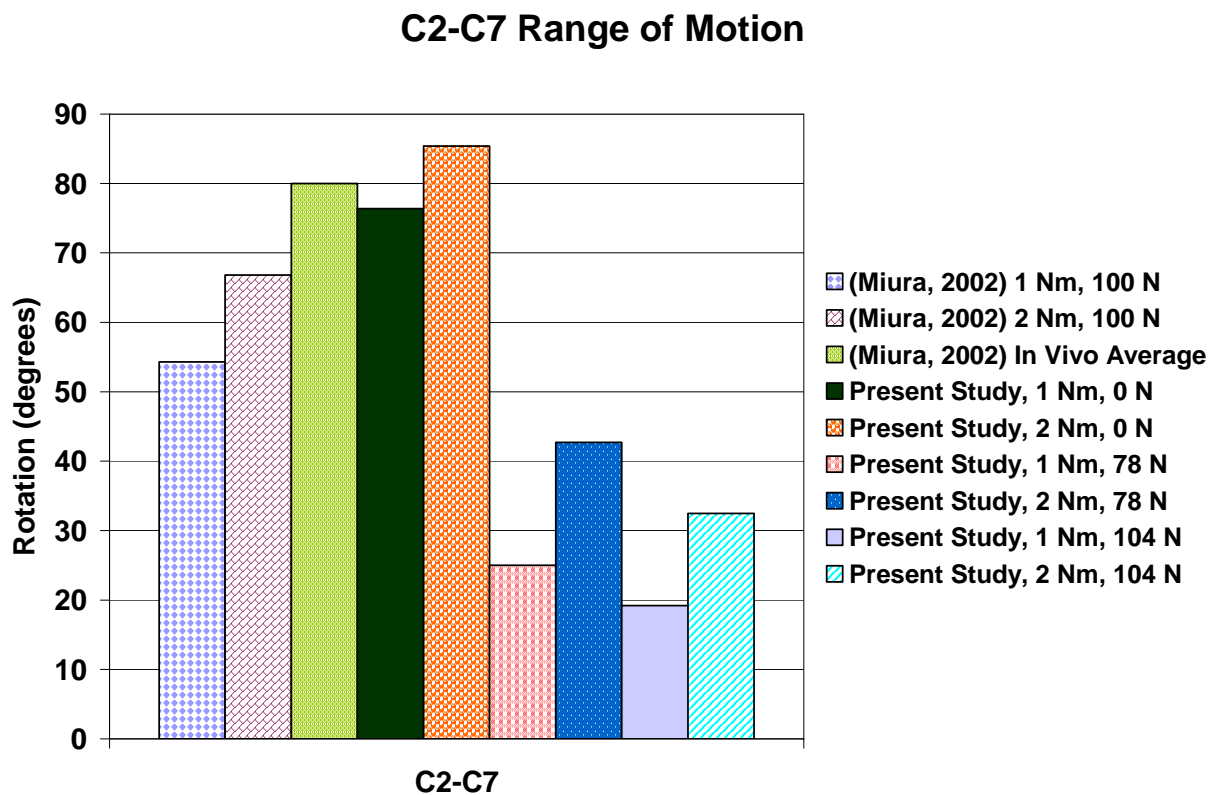
C2-C7 Range of Motion

Figure 4. C2-C7 Range of Motion data for 3 follower load magnitudes (0,78, and 104 N) at 1 and 2 Nm moments compared to *in vitro* data for cadaveric cervical spines with a 100 N follower load at 1 and 2 Nm moments as well as *in vivo* average data.(Miura, Panjabi et al. 2002)

Segmental ROM data for 1 and 2 Nm moments were in fairly good agreement with published *in vitro* studies utilizing a 100 N follower load and *in vivo* averages (Miura, Panjabi et al. 2002) when no follower load was applied. For both 78 and 104 N follower loads the segmental ROM were considerably lower than both the *in vitro* results and *in vivo* averages. Figure 5 shows the intersegmental results compared to those by Miura.
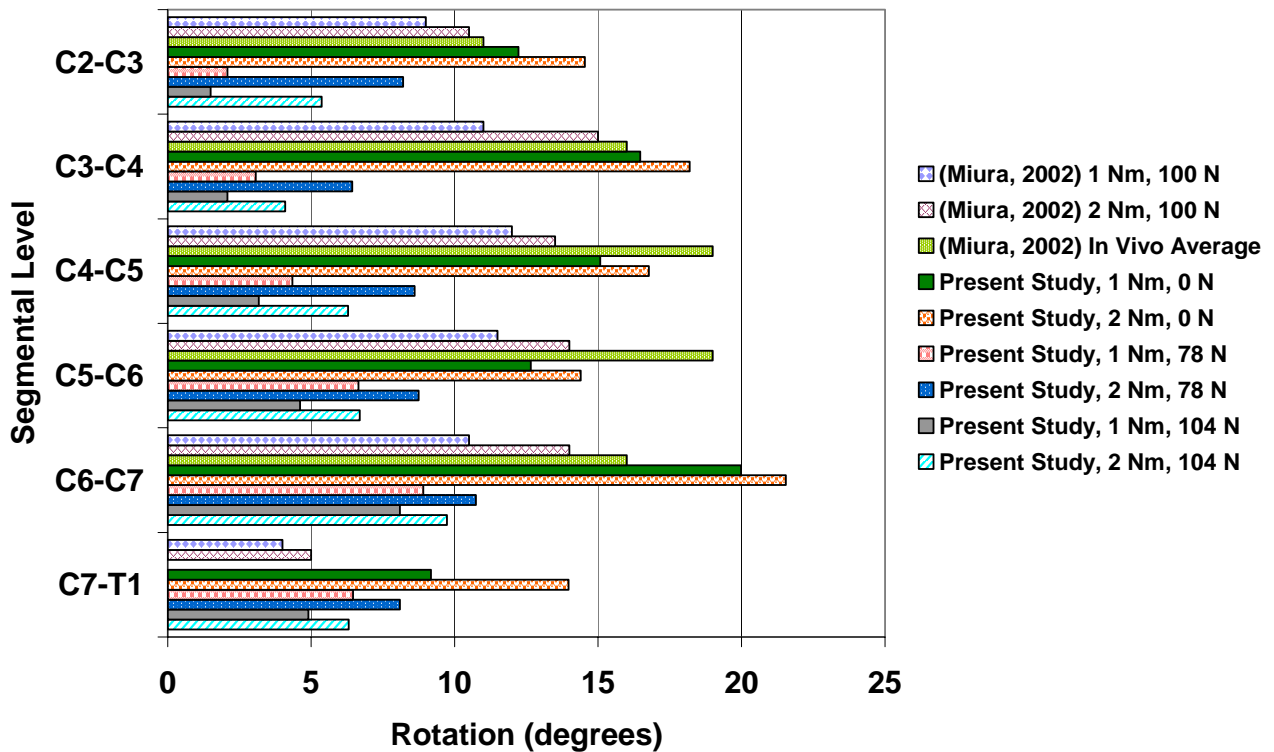
Segmental Range of Motion

Figure 5. Segmental range of motion data for 3 follower load magnitudes (0,78, and 104 N) at 1 and 2 Nm moments compared to *in vitro* data for cadaveric cervical spines with a 100 N follower load at 1 and 2 Nm moments as well as *in vivo* average data.(Miura, Panjabi et al. 2002)

Discussion

Drop Test Kinetics   The overall shape of the head and neck loading curve exhibiting decoupled response upon impacting a rigid surface are in excellent agreement with cadaveric head and neck impacts not utilizing a follower load.(Nightingale, McElhaney et al. 1996) It was observed in the present study that a follower load of 104 N did not have a significant effect upon the neck impulse or neck impulse duration, however it did significantly increase peak neck loading and lower the time lag from when head contact first occurred to when neck loading at T1 began to develop. The observed increase in peak neck load could suggest that compressive injuries are more probable in the presence of muscle contraction. One can imagine that if the neck were perfectly rigid, then there would be no decoupling between the head and neck, thus this decrease in time lag is expected and with increasing magnitudes of follower load the time lag would continue to decrease until no such decoupling is observed.

Flexibility Testing   The ROM data without a follower load was within an acceptable tolerance of published values but the addition of a 78 or 104 N follower load drastically lowered the ROM. The segmental flexion-extension moment rotation response was non linear and exhibited a large neutral zone.   This is typical of some *in vitro* spine specimen responses.(Goertzen, Lane et al. 2004) The choice of maximum moment (6 Nm) in this study's loading protocol was to ensure the surrogate spine moved through its entire range of motion. However, for proper comparison to the referenced literature, separate tests should have been performed up to 1 and 2 Nm respectively. The neutral zone is determined by the residual difference in the flexion and extension rotations on the unloading cycle when a continuous dynamic loading protocol is used.(Goertzen, Lane et al. 2004). In this sense, the neutral zone is a function of the loading history, and for the data gathered in this study, it would only be valid to compare studies also utilizing a 6 Nm maximum moment. In addition, only 1 test was performed at each follower load magnitude as this was the first attempt at characterizing the surrogate neck's response. This precludes any sort of repeatability or statistical analysis.

CONCLUSIONS

The surrogate neck shows a highly repeatable response both kinetically and kinematically to head-first impacts. While the absolute values for neck and head forces are much higher than human, the pattern observed at these higher loads is very similar to values obtained from cadaveric testing. The addition of a follower load had no effect upon neck impulse magnitude but did effect impulse duration and thus peak neck load. The lag time between initial head contact and neck load developing was significantly reduced with the addition of a 104 N follower load. While the addition of a follower load changed the kinematic patterns upon impact, it also changed the pre-impact posture. Thus the role of the follower load upon kinematics is still unclear.

Flexibility testing showed that in the absence of a follower load, the ROM is near to values thought to exist *in vivo.* The addition of a follower load significantly reduced the ROM. Additional flexibility tests should be performed utilizing cut off magnitudes of 1 and 2 Nm for better comparison to literature, to allow calculation of the neutral zone, and to allow for statistical inferences to be made.

ACKNOWLEDGEMENTS

This paper has not been peer- reviewed and should not be referenced in open literature.

REFERENCES

Camacho, D. L. A., R. W. Nightingale, et al. (1997). <u>Experimental flexibility measurements for the developement of a computational head-neck model validated for near-vertex head impact</u>, Lake Buena Vista, Florida, 41st STAPP Car Crash Conference.

Cripton, P.A. (Personal communication). <u>The Effects of Compressive Physiologic Preload on the In Vitro Flexibility of the Lower Cervical Spine.</u>

Dong, Y., M. Xia Hong, et al. (2003). "Quantitative anatomy of the lateral mass of the atlas." <u>Spine</u> **28**(9): 860-3.

Dvorak, J., M. M. Panjabi, et al. (1991). "In vivo flexion/extension of the normal cervical spine." <u>J.Orthop.Res.</u> **9**(6): 828-834.

Francis, C. C. (1955). "Dimensions of the cervical vertebrae." <u>Anat Rec</u> **122**(4): 603-9.

Goertzen, D. J., C. Lane, et al. (2004). "Neutral zone and range of motion in the spine are greater with stepwise loading than with a continuous loading protocol. An in vitro porcine investigation." <u>J Biomech</u> **37**(2): 257-61.

Hattori, S., H. Oda, et al. (1981). "Cervical intradiscal pressure in movements and traction of the cervical spine." <u>Zeitschrift fur Orthopadie</u> **119**: 568-569.

Kandziora, F., R. Pflugmacher, et al. (2001). "Comparison between sheep and human cervical spines: an anatomic, radiographic, bone mineral density, and biomechanical study." <u>Spine</u> **26**(9): 1028-37.

Katz, P. R., H. M. Reynolds, et al. (1975). "Mid-sagittal dimensions of cervical vertebral bodies." <u>Am J Phys Anthropol</u> **43**(3): 319-26.

McElhaney, J., R. G. Snyder, et al. (1979). "Biomechanical Analysis of Swimming Pool Neck Injuries." <u>Society of Automotive Engineers SP-79</u>: 47-53.

Milne, N. (1991). "The role of zygapophysial joint orientation and uncinate processes in controlling motion in the cervical spine." <u>J.Anat.</u> **178:189-201**: 189-201.

Miura, T., M. M. Panjabi, et al. (2002). "A method to simulate in vivo cervical spine kinematics using in vitro compressive preload." <u>Spine</u> **27**(1): 43-8.

Moroney, S. P., A. B. Schultz, et al. (1988). "Analysis and measurement of neck loads." <u>Journal of Orthopaedic Research</u> **6**: 713-720.

Nightingale, R. W., J. H. McElhaney, et al. (1996). "Dynamic response of the head and cervical spine to axial impact loading." <u>Journal of Biomechanics</u> **29**(3): 307-318.

Nightingale, R. W., B. S. Myers, et al. (1991). "The influence of end condition on human cervical spine injury mechanisms." <u>Society of Automotive Engineers Transactions, Paper 912915</u>: 391-399.

Panjabi, M. M., J. Cholewicki, et al. (1998). "Critical load of the human cervical spine: an *in vitro* experimental study." <u>Clinical Biomechanics</u> **13**(1): 11-17.

Patwardhan, A. G., R. M. Havey, et al. (2000). "Load-carrying capacity of the human cervical spine in compression is increased under a follower load." <u>Spine</u> **25**(12): 1548-54.

Patwardhan, A. G., K. P. Meade, et al. (1998). <u>A "follower load" increases the load carrying capacity of the lumbar spine in axial compression - Part 2: Muscle activation and stabilization</u>, New Orleans, 44th annual meeting of the Orthopaedic Research Society.

Pintar, F. A., N. Yoganandan, et al. (1989). <u>Kinematic and anatomical analysis of the human cervical spinal column under axial loading</u>, Warrendale,PA, 33rd STAPP car crash conference.

Torg, J. S., J. T. Guille, et al. (2002). "Injuries to the cervical spine in American football players." <u>J Bone Joint Surg Am</u> **84-A**(1): 112-22.

AUTHOR LIST

1. Tim S. Nelson
    2054 – 6250 Applied Science Lane, Vancouver, BC, V6T-1Z4
    phone: (604) 822-0670
    E-mail: tsnelson@mech.ubc.ca

2. Peter A. Cripton
    2054 – 6250 Applied Science Lane, Vancouver, BC, V6T-1Z4
    phone: (604) 822-6629
    E-mail: cripton@mech.ubc.ca