1

BIOMECHANICS RESEARCH: Experimental and Computational, Proceedings of the Twenty Sixth International Workshop.

Experiments On the Bending Behavior of Cervical Spine Motion Segments

Roger W. Nightingale, Beth A. Winkelstein,
Kurt E. Knaub, and Barry S. Myers

INTRODUCTION

The NCSA has been conducting a Special Crash Investigation (SCI) since 1991, on all airbag injuries in low-to-moderate severity crashes. As of June 1, 1998, there have been over 120 serious or fatal airbag injuries reported with the majority of cases (64%) involving infants or children. The specific mechanisms by which neck injuries occur in airbag deployments are still unclear. It is generally assumed that neck injury occurs through direct loading of the head and mandible (Maxeiner, 1997, Blacksin, 1993). The resulting acceleration of the head applies traction and extension on the cervical spine and causes injury. Cadaver studies have supported this mechanism, as have patterns of facial and mandibular injury. However, little is known about the relationship between tensile displacement, extension rotation, and injury mechanism. Interestingly, basilar skull fractures and catastrophic upper cervical spine injuries have also been demonstrated in cadaver experiments involving blunt impact to the torso (Clemens, 1972, Cheng, 1982). This raises the possibility that out-of-position occupants may sustain cervical spine injuries due to rapid posterior acceleration of the torso relative to the head, with tensile loading of the skull base by the neck (Nightingale, 1998).

Trends in the SCI database indicate that as many as 90% of pediatric neck injuries in airbag deployments are above C2. The predominance of upper cervical spine injuries in airbag deployments in the pediatric population is consistent with the epidemiology. The data for the adults is not yet conclusive; however, there is a trend of more upper cervical spine injuries in airbag deployments than in the general injury population. It is apparent from the injuries to date that the biomechanics of the upper cervical spine should be a primary area of focus in efforts to reduce airbag-related fatalities.

There is an obvious need for a better understanding of the mechanisms of airbag related cervical spine injuries and of the tolerance of the cervical spine in tension and in bending. MADYMO modeling of airbag deployments with out-of-position occupants has shown that the loads imparted are considerable, and that the neck loads greatly exceed the recommended tolerance values in moment and in tension (Kleinberger, 1997, Nightingale, 1997). Unfortunately, the current tolerance values are somewhat inferential, and there is not much quantitative experimental data on the strength of the cervical spine in tension and bending. Without this knowledge, it will be difficult to develop more advanced mechanical surrogates and, consequently, it will also be difficult to develop rational standards for the aggressivity of airbag deployments.

There are considerable data on the stiffness and flexibility of the cervical spine in flexion and extension (Liu, 1982, Panjabi, 1986, Panjabi, 1988, Moroney, 1988, Mcelhaney, 1988, Shea, 1991, Oda, 1992, Camacho, 1997) However, most of these studies focused on the low load responses with the goal of quantifying the amount of laxity (or neutral zone) in the motion segments. Therefore, much of the data is for small applied moments, which are not relevant to failure. In addition, there are no studies that report failure properties for motion segments in pure bending. Since failure of the cervical spine during inertial or airbag loading may have a large bending component, the moment tolerance of the cervical spine is an important value. The only existing estimates for the tolerance of the cervical spine in bending are defined for specific cases of voluntary flexion and belted deceleration (Mertz, 1973). Given the demonstrated coupling between tension and bending (Shea, 1992) and the highly inferential nature of the current bending tolerance, the existing data on bending failures of the neck are not adequate for airbag applications.

METHODS

Pure moment testing was performed for 25 motion segments from seven cervical spine donors. Lower cervical motion segments (C3--C4, C4--C5, C5--C6, C6--C7, C7--T1) were cast into aluminum cups with reinforced polyester resin. Upper cervical specimens (Occiput--C2) were inverted and mounted in the test frame using halo fixation of the head and casting of the C2 vertebra.

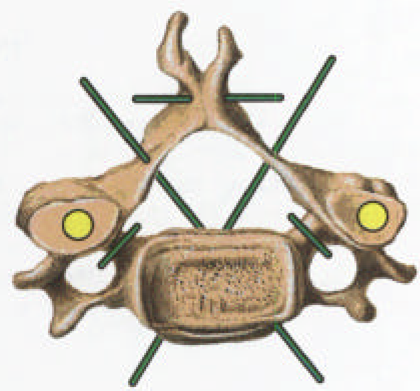

Figure 1: Illustration of the translaminar wire fixation technique with pedicle loops.

Perhaps the greatest challenge in bending and tension tests of cervical spine motion segments is achieving adequate fixation. This is not as critical in compression testing because loads can be easily distributed over large contact areas in the casting material. In tests which involve large components of tension; however, the fixation must carry most, if not all, of the load. It is impossible to achieve good fixation without creating stress concentrations in the bone. As a result, any osseous injuries that are produced originate in, or propagate through, the fixation points. In addition, the reduced strength of the vertebrae increases the likelihood of a bony failure rather than a soft tissue failure. Our fixation technique has been evolving for several years and we currently have a high success rate. The fixation includes crossed, translaminar K-wires which pass through the vertebral body as close to the endplates as possible. A wire is also passed through the spinous process. The most important fixation comes from wires that are looped around each pedicle (Figure 1). These are pre-bent and inserted in the vertebral artery foramen so that one end of the wire is in the casting material. In order to minimize straightening of the wire during specimen testing, the largest possible gage is used.

After being mounted in the test frame (Figure 2), specimens are preconditioned for 30 cycles of flexion and extension. A six-axis load cell is used to measure the loads at the base of the specimen and ensure that the imposed bending moment remain pure. Failures are imaged at 125 frames/s. Failure is defined as a decrease in moment with increasing rotation. ANOVA is used to determine differences between the upper and lower cervical spine failure loads. Specimen dissection is performed to document injuries.

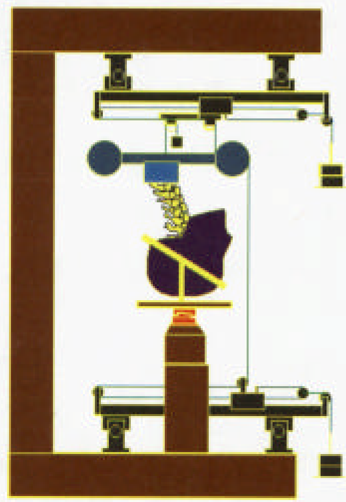

Figure 2: A schematic of the bending flexibility test frame. Although the figure shows a whole cervical spine, the same setup was used for motion segment testing. Bending moments are kept pure using 2-D translation stages supporting the force couple.

RESULTS

Preliminary results suggest that the upper cervical spine (O-C2) is more than twice as trong as the lower cervical spine. For the lower cervical spine, the average failure moment in flexion is 17.03 ± 4.55 N-m (n=10) and the failure moment in extension is 18.09 ± 5.10 N-m (n=9). For the upper cervical spine, the average failure moment in flexion is 46.09 ± 31.86 N-m (n=2) and the failure moment in extension is 44.70 ± 13.91 N-m (n=5). Despite the small number of specimens, a t-test shows that the difference in strength in extension between the upper and lower cervical spines is statistically significant $(p<0.01)$. The injuries to the upper cervical spine were two Type III dens fractures, and an O-A dislocation after fracture of spontaneously fused occipital condyles (Table 1).

Table 1: Tolerances and Failures for Upper Cervical Spine Motion Segments

ID	Sex	Age	Segment	Mode	Moment	Failure
B01	F	66	O-C2	Extension	-45.19	Type III Dens
B02	M	72	O-C2	Extension	-65.76	Fixation
B03	F	51	O-C2	Flexion	68.62	Halo Fixation
B04	F	46	O-C2	Extension	-34.50	Fixation
B05	F	42	O-C2	Extension	-30.03	Fixation
B06	F	49	O-C2	Extension	-48.00	O-A Dislocation
B07	F	52	O-C2	Flexion	23.56	Type III Dens

DISCUSSION

All failures were included in the statistical analysis regardless of whether the failure was judged to be due to stress concentration at the casting. Therefore, the moment values are below the lower bound of tolerance in the cadaver. Interestingly, the Type III dens fractures were produced in both flexion and extension. These fractures have been previously attributed to shear, compression, and extension. Realizing that moments must be supported, in part, by tensile loads in the anterior most portions of the cervical spine, we hypothesize that the C1-C2 injury mechanism in airbag deployments is the result of tensile stresses in the alar and apical ligaments due to combined tension and bending. The resulting tension and bending of the dens causes it to be avulsed from C2.

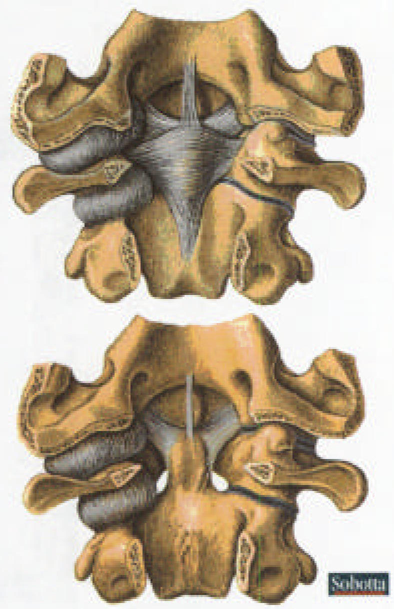

Figure 3: Posterior view of the cervical spine. This 2-layer posterior dissection shows the ligaments between C2 and the skull (Medical Illustration Library, Williams & Wilkins, 1996)

Study of the anatomy of the upper cervical spine suggests that the alar ligaments are the most robust soft tissue structures in the O-C1-C2 complex. They insert on dorsum of the dens (C2) and originate on the inner side of the occipital condyles bilaterally (Figure 3). Inserting on the apex of the dens is the apical ligament, which originates on the anterior margin of the foramen magnum. The tectorial membrane, the posterior longitudinal ligament and the superior cruciform ligament of the atlas also originate on the anterior margin of the foramen magnum. These six ligaments form a fibrous chord between C2 and the skull that is effectively responsible for holding the head onto the cervical spine.

It is possible that the occipito-atlantal dislocation and the atlanto-axial dislocation occur by identical loading mechanisms that result in two different structural failures along the same load path (Figure 4). In the O-A dislocation, tensile stresses on the alar ligaments cause them to fail. This results in rapid failure of the remaining, less robust, ligaments and continued motion of the head with subsequent cord and/or brain stem injury. The atlanto-axial injury may occur when the same tensile stresses cause an avulsion of the dens from the body of C2 (a Type III dens fracture). The failed dens and the intact superior cruciform ligament load C1 and cause in to separate from C2. Which of the two failures occurs may, in part, be related to the age of the victim. In the elderly osteoporotic donor, the dens fracture may be more likely to occur. In the younger donor, the bone may tend to be stronger than the ligaments.

One of our most interesting findings is that the upper ligamentous cervical spine has a greater bending strength than the lower ligamentous cervical spine. However, this is not consistent with the epidemiology. Although it is expected that the weakest point would be the site of failure, the results of the SCI show a large number of upper cervical spine injuries in both adults and children. This discrepancy is most likely due to the effects of the active musculature. The muscles of the cervical spine share tensile loads with the ligamentous cervical spine by providing a parallel load path. Such load sharing increases the overall strength and stability of the neck, and provides greater protection to the caudal motion segments because of the larger size and number of muscles in the lower cervical spine.

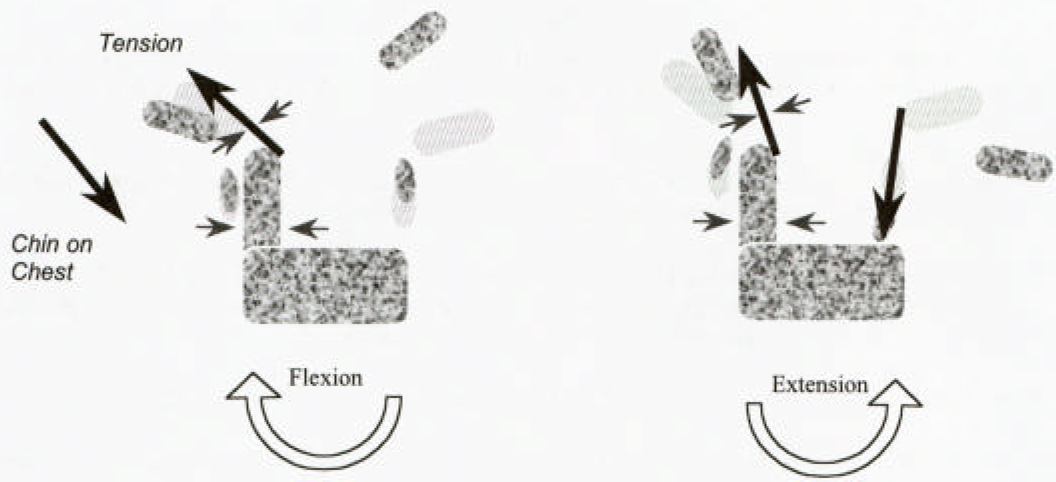

Figure 4: Schematic of the force acting on the dens and the O-C2 ligamentous complex during flexion and extension loading. Flexion moments are reacted by compression of the chin on the sternum and by tension in the dens. Extension moments are reacted by compression in the posterior elements and tension in the dens. Failure (double arrows) occurs in either the dens, or the occipito-axial ligamentous complex.

CONCLUSIONS

- In bending, the adult upper cervical spine is stronger than the adult lower cervical spine. However, the epidemiology suggests that muscles may preferentially protect the lower cervical spine.
- The mechanisms of upper cervical spine injuries in children and adults are likely related to the relative strengths of the dens and the alar ligaments.
- Future research on neck tolerance in tension and bending should account for the effects of the musculature in addition to the properties of the pediatric and adult ligamentous cervical spines

Acknowledgments: NHTSA Contract DTNH22-94-Y-07133. CDC Grant R49/CCR402396-12.

Correspondence: Duke University, Department of Biomedical Engineering, Box 90281, Durham, NC 27708-0281.

REFERENCES

Blacksin, M. F. (1993). Patterns of fracture after air bag deployment. J Trauma, 35 (6):840--3.

Camacho, D. L. A., Nightingale, R. W., Robinette, J. J., Vanguri, S. K., Coates, D. J., and Myers, B. S. (1997b). Experimental flexibility measurements for the development of a computational head-neck model validated for near-vertex head impact. In 41st Stapp Car Crash Conference Proceedings , number 973345, pages 473--486. Society of Automotive Engineers.

Cheng, R., Yang, K., Levine, R., King, A., and Morgan, R. (1982). Injuries to the cervical spine caused by a distributed frontal load to the chest. In Proc. 26th Stapp Car Crash Conference, number 821155.

Clemens, H. J. and Burow, K. (1972). Experimental investigation on injury mechanism of cervical spine at frontal and rear-front vehicle impacts. Proc. of the 16th Stapp Car Crash Conference, (720960).

Kleinberger, M. and Summers, L. (1997). Mechanisms of injuries for adults and children resulting from airbag interaction. AAAM Proceedings of the 41st Annual Conference .

Liu, Y. K., Krieger, K. W., Njus, G., Ueno, K., Conners, M. P., Wakano, K., and Thies, D. (1982). Cervical spine stiffness and geometry of the young human male. Technical Report AFAMRL-TR-80-138, Air Force Aerospace Medical Research Laboratory.

Maxeiner, H. and Hahn, M. (1997). Airbag-induced lethal cervical trauma. J Trauma, 42 (6):1148--51.

McElhaney, J.~H., Doherty, B.~J., Paver, J.~G., Myers, B.~S., and Gray, L. (1988). Combined bending and axial loading responses of the human cervical spine. Proc. 32nd Stapp Car Crash Conference}, (881709):21--28.

Mertz, H., Neathery, R., and Culver, C. (1973). Preformance requirements and characteristics of mechanical necks. In King, W. and Mertz, H., editors, Human Impact Response: Measurement and Simulation, pages 263--288. Plenum Press, New York.

Moroney, S. P., Schultz, A. B., Milles, J. A. A., and Andersson, G. B. J. (1988). Load-displacement properties of lower cervical-spine motion segments. Journal of Biomechanics , 21 (9):769--779.

Nightingale, R. W., Kleinberger, M., and Myers, B. S. (1997a). Effects of upper neck joint stiffness on measured moments in the Hybrid III dummy during airbag loading. Technical report, National Highway Traffic Safety Administration. submitted.

Nightingale, R. W., Winkelstein, B. A., Van Ee, C. A., Myers, B. S. (1998). Injury mechanisms in the pediatric cervical spine during out-of-position airbag deployments. In 42nd Annual Proceedings of the Association for the Advancement of Automotive Medicine, pages 153--164.

Oda, T., Panjabi, M., Crisco, J. J., Bueff, H. U., Grob, D., and Dvorak, J. (1992). Role of the tectorial membrane in the stability of the upper cervical spine. Clinical Biomechanics, 7 :201--207.

Panjabi, M. M., Dvorak, J., Duranceau, J., Yamamoto, I., Gerber, M., Rauschning, W., and Bueff, H. U. (1988). Three-dimensional movements of the upper cervical spine. Spine, 13 (7):726--730.

Panjabi, M. M., Summers, D. J., Pelker, R. R., Videman, T., Friedlaender, G. E., and Southwick, W. O. (1986). Three-dimensional load-displacement curves due to forces on the cervical spine. Journal of Orthopaedic Research, 4 :152--161.

Shea, M., Edwards, W. T., White, A. A., and Hayes, W C. (1991).Variations of stiffness and strength along the human cervical spine. J Biomech 24(2):95-107.

Shea, M., Wittenberg, R. H., Edwards, W. T., A. A. White, I., and Hayes, W. C. (1992). In vitro hyperextension injuries in the human cadaveric cervical spine. Journal of Orthopaedic Research , 10 (6):911--916.