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Tensile Testing of the Ligamentous Cervical Spine: Biomechanical Considerations for a Proposed Testing Methodology

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ABSTRACT

Tensile neck injuries are amongst the most serious cervical injuries. Recent advancements in automotive safety devices, while enhancing overall safety, have increased the importance of tensile neck injury research. Unfortunately however, neither reliable tensile tolerance data nor tensile structural data is currently available. Moreover, standard methods for evaluating the beam-strut behavior of the ligamentous spine are absent from the literature. Therefore, the purpose of this study is to develop a tensile test methodology to quantify the tolerance and kinetic behavior of the ligamentous cervical spine at both the whole spine and motion segment levels. The test methodology developed produces whole spine end condition response corridors, motion segment stiffness, and tolerance data. By adopting a standardized test methodology, tensile injury data generated from amongst different ages, genders, species, and institutions will be more directly comparable and progress more rapid.

INTRODUCTION

Cervical spine injuries due to tensile neck loading have serious and often fatal consequences. They include basilar skull fractures, craniocervical dislocations, Hangman's fractures, odontoid fractures, and a variety of lower cervical injuries (Aberdare, 1886; Braakman and Vinken, 1967; Bucholz and Burkhead, 1979; Bucholz et al., 1979; Fielding et al., 1989; Portnoy et al., 1971; Alker et al., 1975, 1978; Huelke et al., 1980; Mcelhaney et al., 1995). Tensile injuries occur as a result of mandibular and craniofacial impact, as well as non-head-contact accelerations and decelerations (Voigt and Skold, 1974; Harvey and Jones, 1980; Huelke et al., 1988; Mcelhaney et al., 1995). Huelke et al., (1993) hypothesized that cervical spine injuries to restrained occupants in frontal collisions in which head contact did not occur was due to the inertial loading of the neck by the unrestrained head. While airbags have decreased overall injury incidence, the importance of tensile injury mechanisms has increased. In addition, occupant interaction with air-bags can produce tensile neck loading that may result in serious and fatal neck injuries even for low velocity collisions (Sci, 1998; Maxeiner and Hahn, 1997; Blacksin, 1993; Perez, 1996; Kleinberger and Summers, 1997). With the increasing penetration of airbags in the automotive fleet the incidence of these types of injuries is likely to rise. Unfortunately however, there is a paucity of data describing the mechanisms and tolerance of injury of the human neck under tensile loading. Owing to the lack of human tolerance data and models, investigators have been limited in their ability to evaluate modifications in air bag design or other additional injury prevention strategies. Indeed, several current guidelines are limited in part because they are based on scaling animal data (Prasad and Daniel, 1984; Mertz and Weber, 1982)

Given the pressing need for the understanding of the tensile mode of injury, extensive research efforts are currently being conducted at several institutions. Ongoing research includes tensile loading of the adult and pediatric ligamentous spine, whole body cadaver studies of tensile neck loading, size and age scaling studies using the goat and rhesus monkey as surrogates, and computational modeling efforts to understand the contribution of neck bending properties on tensile injury mechanisms (Yoganadan and Pintar, 1999; Ching, 1999).

Compared to other modes of loading in the cervical spine there are relatively few studies on tensile loading. Proposed tensile tolerances have ranged from 1.1 to 6.2 kN. In a set of studies conducted by Patrick on himself, he withstood a tensile load of 1.13 kN, without injury (Mertz and Patrick, 1967 & 1971). This non-injurious static strength was offered as a lower bound for the corresponding dynamic strength. Clemens and Burow (1972) simulated the exposure of motor vehicle crash injuries with cadaveric specimens. In frontal impacts, the vertical cranial accelerations were between 40 and 50 G, with a tensile load at C1 estimated between 1.6 and 2.0 kN using inverse dynamics. Cheng et al., (1982) reported flexion distraction injuries as the results of air bag chest impacts to cadavers. Inverse dynamic methods were used to estimate a neck load tolerance of 6.2 kN. Shea et al., (1992) applied extension-tension loading to nine female ligamentous cervical spine specimens and found the average failure load to be 0.50 ± 0.15 kN. Yoganandan et al.,(1996) reported on the experiments of Sances et al., (1981), including the results of additional tests. Experiments were conducted on four isolated ligamentous spine preparations from skull to T3. Failure loads ranged from 0.80 kN to 1.90 kN. A second set of tests reported by Yoganandan et al., (1996) was conducted on three whole human cadaver specimens. The failure loads were 2.4, 3.8, and 3.9 kN. These studies while representing the first efforts to understand these injuries are limited by small sample sizes and insufficient instrumentation to fully quantify the combined loading at the site and time of injury. While new research efforts are underway to study tensile injuries, a standardized test method to be able to maximize resources and compare results among gender, age, stature, species, and other institutions does not currently exist. Accordingly, it is the goal of this study to design an experimental methodology for the study of tensile neck injury.

METHODS

A standard coordinate system was defined to report both local body fixed and global spatially fixedmotion (FIG. 1). The coordinate system origin was defined as the midpoint of the anterior surface of the vertebral body so that the position of the origin could be determined throughout the test using optical markers without the need for coordinate system transformation. Using the pin array locations for each vertebra, along with the positional data describing the vertebral origin marker, relative incremental translational and rotational motions of the cervical vertebrae were calculated using Euler angle decomposition (Woltring, 1991). The order of Euler decomposition of rotations was chosen to minimize error and reduce the risk for gimbal lock and was flexion-extension, axial torsion, then lateral bending. These Cardan-Bryant angles are reported as the rotations of the body fixed coordinate system of the superior vertebra relative to the body fixed coordinate system of the inferior vertebral body.

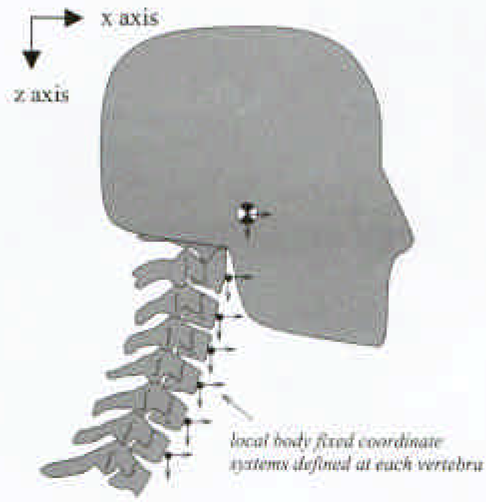

FIG. 1. Coordinate system used in defining and reporting cervical spine kinetics.

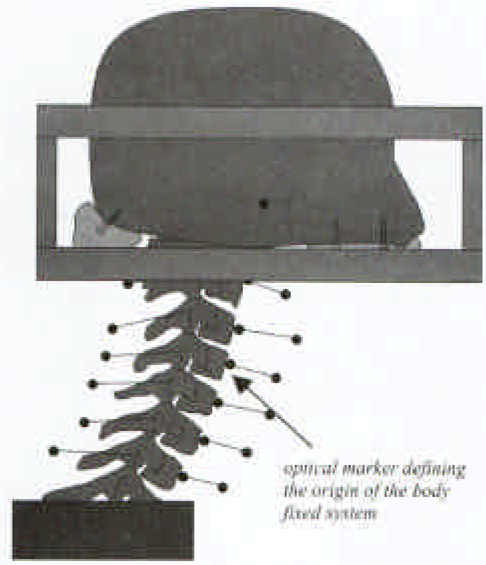

FIG. 2. Illustration of the coupling of the head and the lower spine to the experimental frame mounts. Optical markers are placed in each vertebra to track the motion and locate the body fixed coordinate systems.

Four unembalmed human cadaver specimens from the head through T2 were obtained. Medical records and pre-test radiographs of the specimens were examined to ensure that there were not any unrecognized spinal pathologies which might degrade the structural integrity. The musculature was removed to better visualize the ligamentous cervical spine motions and injury mechanisms. The mandible was removed to allow visualization of the upper cervical spine and to allow for application of loads directly to the maxilla. T1 and T2 were cleaned of muscular tissue and cast into an aluminum cup with reinforced polyester resin allowing unconstrained motion at the C7-T1 level. Casting of T1-T2 was performed with the T1 vertebra oriented with a downward pitch of 25 degrees from the horizontal in the casting cup to preserve normal cervical lordosis (Matsushita et al., 1994). The skull was coupled to the head mount platform using bone screws and fiber reinforced acrylic (FIG. 2). Care was taken to allow full motion at the Occiput-C1 level. In addition, the mount points of the skull were chosen remote from the base of the skull to mitigate the significance of any stress concentrations in the regions where basilar skull injuries commonly originate (Mcelhaney et al., 1995). Finally, opaque target pins (4.0 mm diameter) were inserted in the anterior vertebral body and the posterior portion of the lateral masses of C2-C7 (FIG. 2). Pins were located so that the body fixed coordinate systems could be obtained directly from the images without errors associated with transformation.

The center of gravity (CG) was located by scaling the data reported in Walker et al., (1973) of the average position of the head CG with reference to the external auditory meatus and the inferior orbital margin. First, a head scaling factor was determined for each specimen. The distance from the glabella to the opisthocranion was measured using calipers. A head scaling factor was determined by dividing this measured distance by the average distance of 183 mm reported by Byars et al., (1970). Walker et al., (1973) report the average location of the head CG to be 24.2 mm at an angle of 292.3 degrees clockwise with respect to a line connecting the external auditory meatus and the inferior orbital margin. The 24.2 mm was scaled by the head scaling factor and an optical marker was placed on the specimen to approximate the head CG.

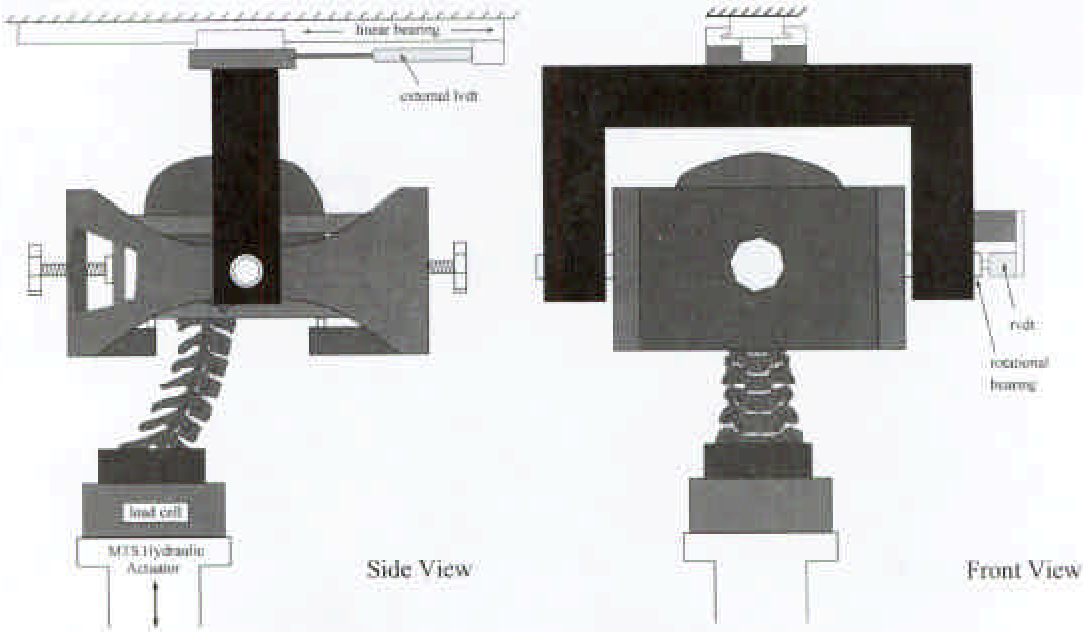

FIG. 3. Test fixture used for tensile test experiments. The fixture produces a pure vertical load at the center of the rotational bearing by minimizing the shear and rotation through the use of the linear and rotational bearing respectively. Both the linear and rotational degrees of freedom can be fixed to apply varying end conditions.

The head and neck were then placed in the experimental test frame (FIG. 3). The test fixture applied a pure vertical load (based on the global coordinate system) at the center of the rotational bearing aligned with the CG of the head. Pure tensile loading was obtained by use of the linear and rotational bearings coupling the head to the hydraulic actuator. An RVDT located at a rotational bearing quantified head rotation. Two LVDTs were used to monitor the hydraulic actuator position and the linear bearing position. Either of the two sagittal degrees of freedom (AP translation and rotation) could be locked to examine the effect of cranial end conditions. A six axis GSE load cell coupled the lower portion of the spine to an MTS hydraulic actuator. Data were collected using digital data acquisition (National Instruments; Austin, TX). A CCD camera was used to capture the motion within the sagittal plane. The head mount platform was adjusted so that the line of action of the loading vector passed through the head CG. The head position was adjusted within the load frame so that the Frankfort plane was horizontal defining the reference position.

Following specimen preparation, one of two different test protocols, Whole Spine and Partial Spine, were applied. Both methods used the same techniques for mechanical stabilization. That is, after the specimen was placed in the frame, a 2 mm/s constant velocity test was performed to a 300 N load interrupt under fixed-fixed end conditions. Peak displacement was recorded. Mechanical stabilization was then performed consisting of 50 cycles of a 0.5 Hz Sine wave with amplitude and mean of 25% of the peak displacement of the previous test.

The Whole Spine protocol consisted of nondestructive end condition testing followed by a series of sequential failure tests. Nondestructive end condition testing was performed using a 2 mm/s constant velocity test to a force level of 200 N. The four cranial end conditions, fixed, rotational, translational, and free, were tested. For each test, a stiffness and low-load elongation region were defined by regressing the tensile force elongation response between 150 and 200 N. The slope was defined as the stiffness and the abscissa intercept was defined as the low-load elongation. Failure tests consisted of a 2 mm/s constant velocity test to complete distraction failure. The first failure test was conducted on the whole specimen, Occiput-T1. After a complete distraction failure the remaining superior portion of the spine was re-cast and another failure test was conducted. This was repeated until C2 was separated from Occiput.

The Partial Spine protocol consisted of sectioning the spine into motion segments (Occiput-C2, C4-C5, C6-C7) prior to testing. Following mechanical stabilization, each motion segment underwent a stiffness test with fixed-fixed end conditions using a 2.0 mm/s constant velocity test to a force level of 300 N. A 2 mm/s constant velocity failure test was then conducted using a free end condition.

RESULTS

Coupling to the specimen for tensile testing proved to be particularly demanding. Initial casts using methods shown to be reliable in numerous prior studies resulted in casting failures in 50% of specimens loaded to failure. Refinement in casting for the demands of tensile failure testing reduced potting failures to 10% of the tests. For the upper cervical segments, casting two vertebrae (C2, C3) using a combination of bone screws and k-wires pre-molded into fiber reinforced acrylic was found most successful. For lower cervical vertebra, supra-pedicular loops were used which traveled from the casting material up through the vertebral foramen over the pedicle and down the transverse foramen. The supra-pedicular loops were used in conjunction with translaminar vertebral body k-wires and fiber reinforced acrylic (FIG. 4).

Head and neck tensile response was greatly influenced by the cranial end condition exhibiting both decreases in stiffness of 60% and increases in the low-load elongation of 1000% as the end conditions changed from fixed to free (FIG. 5). Stiffness for the fixed, rotational, translational and free end conditions were 69, 56, 42, and 27 N/mm respectively, and low-load elongations were 1.1, 1.1, 6.5, and 10.5 mm, respectively.

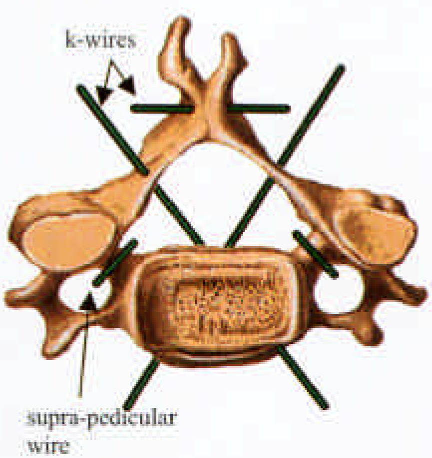

FIG. 4. Illustration of the fixation technique of the lower spine vertebra including the supra-pedicular wire fixation.

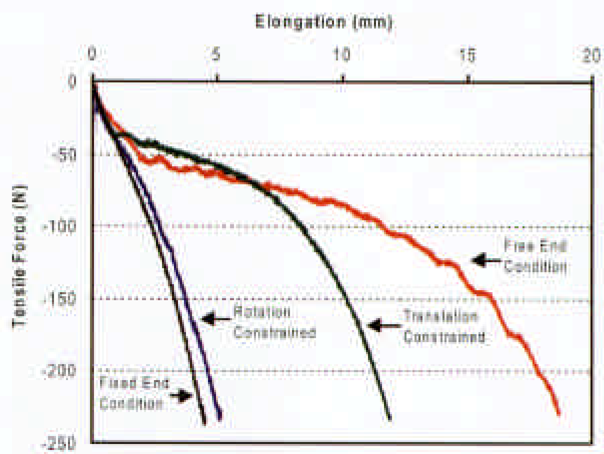

FIG. 5. Force-deflection history of a specimen in Whole Spine nondestructive end condition testing. Results show cranial end conditions have a dramatic affect on stiffness and low-load elongation.

The Partial Spine method more accurately determined motion segment tensile stiffness properties when compared the Whole Spine method. Whole Spine tests with pure tensile loads applied to the head resulted in combined tension-extension loading at the motion segment level. On average lower cervical segments were found to extend 3 degrees at failure while the O-C2 complex extended

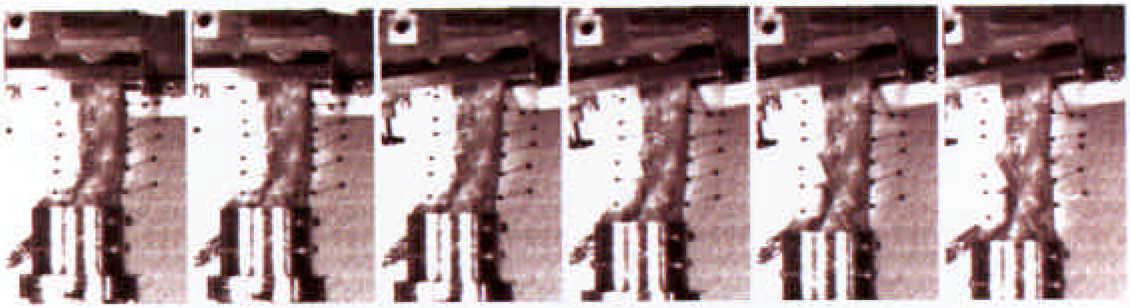

FIG. 6. Sequential sagittal views of a whole spine failure test. Optical markers were digitized to determine local coordinate systems and motion segment response.

an average of 20 degrees at failure. These combined loads complicated the determination of motion segment tensile force-displacement properties, by requiring a structural matrix decomposition of the combined loads. Determining the force-displacement response required the motions and the loads to be resolved at the pure moment flexion-extension center of rotation. Further, using Whole Spine methods, optical methods were used to quantify local motions resulting in measurement accuracy of displacement of ±0.5 mm and rotation of ±0.2 degrees (FIG. 6). The combination of determining the pure moment center of rotation and the limitation in displacement measurement accuracy resulted in a variation of stiffness measurement from 78 N/mm to 813 N/mm for a single specimen. In contrast, using Partial Spine methods, a fixed-fixed stiffness test could be performed on a single motion segment resulting in the direct measurement of the tensile stiffness response (FIG. 7). This direct measurement, using the LVDT to measure displacement (±0.015 mm) and not requiring any decomposition of motion or loads, resulted in a smaller variation of stiffness measurement from 360 N/mm to 380 N/mm for a single specimen.

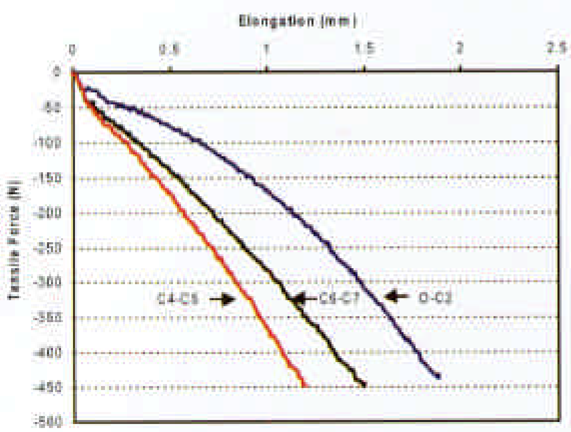

FIG. 7. Motion segment stiffness response of a fixed-fixed test. The Partial Spine protocol allows for the direct measurement of motion segment stiffness properties.

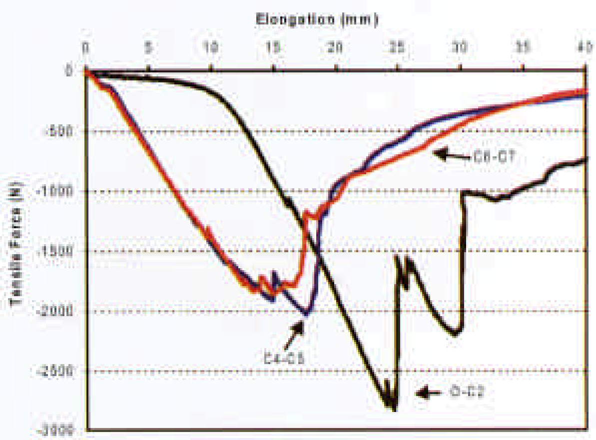

FIG. 8. Failure histories of a set of motion segments from the Partial Spine Protocol. Results from this specimen suggest the lower spine tolerance is weaker than the upper spine under tensile loading.

Partial Spine failure test methods resulted in clinically observed upper cervical injuries and allowed for the direct comparison of upper versus lower cervical spine tolerance. In contrast, Whole

Spine failure tests resulted in failures in the lower spine, which are uncommon clinically and demonstrated that repeat testing compromised the upper cervical tolerance. For example, an Occiput-C3 test segment failed with a peak load of 2.4 kN sustaining a C2–C3 intervertebral diastasis. The specimen was then recasted from Occiput-C2 and the resulting failure load was 2.0 kN with a Occiput-C1 dislocation. This drop in failure load showed that the Occiput-C2 section accumulated damage in the previous test resulting in a lower failure load in the following test. For each of the sequential failure tests of the Whole Spine method, the failure site was at or near the inferior casting site. Upper cervical spine failures occurred only when the upper cervical spine comprised the entire test segment. In contrast, Partial Spine methodology resulted in failure test data of uncompromised upper and lower cervical motion segments and allowed for direct comparisons in relative strengths of these structures (FIG. 8). These tests show that the lower cervical spine is weaker than the upper cervical spine, and explains the predominance of lower cervical injuries in the whole spine tests. Injuries in the lower spine were ligamentous distraction injuries resulting in joint diastasis, while injuries produced in the upper cervical spine included a type III dens fracture and an occipitoatlantal dislocation.

DISCUSSION

A lack of tensile tolerance data and suitable biofidelic models has limited the ability of investigators to analyze and thereby mitigate tensile injuries. Although a large research initiative is underway to understand tensile injury mechanisms in pediatric and adult populations of different genders and statures, a uniform test methodology has not been developed. Such a methodology, were it universally adopted, would maximize resources by allowing direct comparisons between the complimentary studies using different specimen populations delineating the effects of gender, stature, age, and species. In this study, two test protocols, the Whole and the Partial Spine methods, were investigated to better understand their merits and shortcomings.

One limitation inherent in this study is the use of only four specimens. As the purpose of the investigation is protocol development and not definition of tolerance and structural stiffness based on statistical methods, this small number of specimens proved ample. Another limitation of this study is that different rates of loading were not investigated. Rate effects manifest themselves through both viscoelastic and inertial means and both may prove important to understanding tensile neck injury (Myers et al., 1991; Yoganandan and Pintar, 1999). While therefore of considerable importance, rate effects will require a large study and will therefore need to be developed following completion of the quasi-static work.

Any proposed test battery must meet four specific aims in order to fill the voids in the current literature which have limited the understanding of tensile injury mechanisms and their mitigation. The battery should define the tolerance of the cervical spine under tensile loading. The resulting experimental model must reproduce the clinically observed tensile neck injuries. The battery should provide sufficient data for the development and refinement of computational models and anthropometric test devices used to study tensile injury. Finally, the battery should provide a database for the validation of these models.

Based on these specific aims, and the results of the Partial Spine and Whole Spine tests, we propose a revised tensile test methodology (Table 1). The proposed test methods are a combination of the Whole and Partial Spine methods consisting of a series of whole spine non-destructive tests to provide a model validation data followed by motion segment stiffness and failure tests to provide tolerance and model construction data.

This revised methodology allows the determination of the tolerance throughout the spine, while still considering the effects of combined loading, and without the confounding effects of specimen degradation owing to repeat failure testing. For example, the Whole Spine failure tests, which represent a pure load applied to the head, result in a combined tension-extension load throughout the lower cervical spine. Analysis of the lower cervical kinematic data showed that this combined load tension-extension load caused angular rotations of approximately three degrees. By performing motion segment tests so that the extension angle produced is three degrees, the combined loading conditions observed in the whole spine tests are reproduced in the proposed test method. Thus, the tolerance numbers produced from this methodology for the upper and lower cervical spine represent a whole spine loading configuration.

Table 1: Tensile Test Methodology

Non-Destructive Whole Spine Testing (Occiput-T1)
- Fixed-Fixed Test
  - 2 mm/s to a 300 N interrupt
  - Record peak displacement *Lp*
- Mechanical Stabilization (30 cycles)
  - Fixed end condition
  - 0.5 Hz sine wave with mean and amplitude of 25% *Lp*
- End Condition Tests (4 Tests)
  - 2 mm/s to a peak load of 200 N
  - Vary the cranial constraint from Fixed, Translational, Rotational to Free
- Line of Action Tests (4 Tests)
  - 2 mm/s to a peak load of 200 N
  - Free end condition
  - Tensile loading vector 3cm posterior of condyles, condyles, CG, 3 cm anterior of condyles

Section the Spine into Motion Segments (Occiput-C2, C4-C5, C6-C7)

Non-Destructive Motion Segment Testing
- Fixed-Fixed Test
  - 2 mm/s to a 300 N interrupt
  - Record peak displacement *Lp*
- Mechanical Stabilization (30 cycles)
  - Fixed end condition
  - 0.5 Hz sine wave with mean and amplitude of 25% *Lp*
- Stiffness Test
  - Fixed end condition
  - 2 mm/s to a peak load of 300 N

Motion Segment Failure Test
- Occiput-C2
  - Loading at the CG
  - 2 mm/s to complete distraction failure
- C4-C5, C6-C7
  - Load vector placed to produce small amount of extension (3 degrees)
  - 2 mm/s to complete distraction failure

Additionally, the proposed method also has the benefits of obtaining uncorrupted tolerance data. By testing the upper and lower cervical spine sections separately, both upper and lower cervical regions of tensile injuries will be determined. An analysis by Nightingale et al., (1998) reported that as many as 90% of all cervical injuries to children involving airbags occurred in the upper cervical spine. While in adults the trend is less pronounced, there are more upper cervical injuries than in the general injury population. For these reasons, an injury model of the upper cervical spine must be a part of any tensile injury model. Yet, the Whole Spine tests reported herein and previous studies of the cervical spine in tension and combined tension-extension produce predominantly lower cervical injuries (Sances et al., 1981; Shea et al., 1992; Yoganandan et al., 1996). While sequential tests of the Whole Spine method eventually resulted in an upper cervical injury, the tolerance value associated with the failure was compromised due to the successive nature of the failure tests. In contrast, the Partial Spine method produced clinically observed injuries including dens fractures and occipital-atlantal dislocations on uncompomised specimens. In addition, the partial spine method also allows for investigations of distraction injuries in the lower cervical spine. As such, the proposed methodology tests the upper and lower cervical spine in isolation.

The proposed method also provides data for model development. The Partial Spine methods produced significantly improved tensile motion segment stiffness data. By using Whole Spine end condition tests which are nondestructive in addition to the Partial Spine methods, suitable data for validation of ligamentous spine models are generated. That is, these tests allow for the creation of whole spine structural response corridors including flexion and extension motions. The results of this preliminary study show that alteration of the end condition significantly alters the observed behavior. While this behavior has been described in compression beam-column studies of the cervical spine (Myers et al., 1991a), it has yet to be reported in studies of combined loading of the cervical spine in tension-bending. Because of a 2 fold change in stiffness associated with altered end condition and a 10 fold change in low-load elongation as demonstrated in these experiments, the proposed test battery will generate a wide range of response corridors for model validation. Moreover, by moving the line of action of the tensile force along an anterioposterior line traveling parallel with the Frankfurt plane through the CG, the degree of flexion and extension combined with the tensile distraction can be controlled. The addition of the line of action tests would further increase the robustness of the validation set by including increased flexion and extension motions.

CONCLUSIONS

Based on the results of Whole Spine and Partial Spine testing, we propose a standardized test methodology for examining the tensile behavior of the cervical spine. The proposed methodology quantifies tolerance and stiffness of the upper and lower cervical spine separately. In addition, data for the development and validation of computational and physical models of the cervical spine are collected. If adopted by other investigators, the results of studies using this methodology will allow for the direct comparison of cervical biomechanical data in which age, gender, species, and institutions are all variables.

ACKNOWLEDGMENTS

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