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Responses of the Human Cervical Spine to Torsion

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ABSTRACT

The passive torsional responses of the human cervical spine were investigated using unembalmed cervical spines in a dynamic test environment. Kinematic constraints were designed to simulate *in vivo* conditions. A physiologic axis of twist was determined based on a minimum energy hypothesis. Six-axis load cells completely described the resultant forces. Results include viscoelastic responses, moment-angle curves, and piecewise linear stiffness. The Hybrid III ATD neckform was also tested, and its responses compared with the human. The Hybrid III neckform was stiffer than the human, was more rate sensitive than the human, and unlike the human, was relatively insensitive to the axis of twist. A rotational element to improve the biofidelity of the Hybrid III neckform in rotation was developed, and the results presented. In addition, this data was compared with volunteer sled tests to determine the contributions of the cervical musculature to the stabilization of the neck in rotation during lateral acceleration.

THIS PAPER DESCRIBES THE responses of the unembalmed cadaver cervical spine to axial rotations of the head about a vertical axis through the dens. The work arises from concern for the contributions of rotation in determining the head contact site of an automobile occupant in both frontal and side impacts, the role of axial rotation in the development of head and neck injury, and the potential for torsional neck injuries in the high g environments of fighter aircraft ejections. It is hoped that this information may be used to increase the database upon which the biofidelity of both computer-based and mechanical simulations may be assessed.

A considerable portion of the literature has been devoted to the characterization of the responses of the cervical spine to loading in the saggital plane (frontal impacts) [1] and in the coronal plane (side impacts) [2,3,4,5]. However, only a few references are available which describe the axial rotations, angular accelerations and torque responses of the head-neck system. Bowman *et al.* [6] report a rotational stiffness of 0.339 N-m/degree. Wismans and Spenny [4] report a piecewise linear response produced from volunteer sled tests in the lateral direction. Mean stiffnesses were reported as 0 N-m/degree for 0 to 10 degrees of axial rotation from the neutral position, 0.5 N-m/degree for 10 to 30 degrees, and 0.25 N-m/degree for rotations greater than 30 degrees. Subsequent reports from these authors [3] list the stiffness as varying from 0.4 N-m/degree in lateral impacts to 0.75 N-m/degree in oblique impacts.

Axial rotation has been shown to play a role in car crash kinematics in a number of studies. Wismans *et al.* [3] report the magnitude of axial rotation during volunteer lateral accelerations as equal to the head flexion angle, and as having maximum excursions of approximately 23 to 46 degrees from the neutral position.

Viano and Culver [7] report asymmetric thoracic motions with axial rotation of a test dummy interacting with a shoulder belt in three-point belted frontal decelerations. The relatively common occurrence of head-steering wheel impact in frontal collisions with three-point belted passengers [8,9], together with the above, suggests that axial rotation plays a role in determining the site of impact in head steering wheel impacts. Further, Nusholtz *et al.* [10] have reported that angular accelerations and the initial position of the head in the saggital plane at the time of impact, play an important role in the potential for brain injury. This suggests the importance of accurate characterization of axial rotations and accelerations in the car crash environment.

The kinesiology of the cervical spine has been studied in various ways [11,12,13,14]. Static weight pulley systems have been used to describe the responses of the spine [15,16,17]. However, the complexity of the spine and the failure of rigid body mechanics to adequately describe the responses of the spine [18] require the use of viscoelastic, and dynamic analyses.

Dynamic test systems typically provide the necessary control for viscoelastic testing and controlled injury induction; however, they constrain the specimen to motions defined by the actuator. The need to recreate physiologic motions and end conditions while using such devices is fundamental to the generation of useful data.

The occipitoatlantal joint is irrotational [19]. In contrast, the atlantoaxial joint shows striking mobility at very low torques, accounting for more than fifty percent of cervical axial rotation. The odontoid process of the second cervical vertebra (the dens) articulates with the atlas (C_1) such that the dens acts as the center of upper cervical rotation. Coupling of rotation with axial displacement and lateral bending in the atlantoaxial joint has been reported, but not quantified [20].

In the lower cervical spine, kinematics are more complicated. Coupling of rotation and lateral bending has been described such that the spinous processes rotate axially into the convexity of the coupled lateral curve. This is thought to be mediated by the obliquely oriented facet joints [16]. Coupling of rotation and axial displacement is also thought to occur but has not been quantified.

The existence and selection of an appropriate axis of twist has received considerable attention [17]. Strictly, the coupling of motions makes spinal kinematics nonplanar. Notwithstanding, an axis of twist must be identified for use in a dynamic test mode. Static domain testing identified the center of lower cervical rotation as the anterior most portion of the vertebral body along the midsaggital line [12,13]. The method used two x-ray images (A-P and Lateral) for each applied load. These images were digitized, and the center of rotation computed using the method of instant centers of velocity. Unfortunately, this method is computationally cumbersome, cannot be performed quickly, and is only suitable for static domain tests [12,13].

Facet geometry has been implicated as a factor in defining the center of rotation [21]. Specifically, the intersection of perpendicular bisectors of the anteriomedially oriented facet joint surfaces is thought to establish a preferred center of rotation. However, reduction of geometric data from Liu *et al.* [22] demonstrates anteriolateral facet joint orientation, asymmetry of facet orientation, and an absence of intersection of the perpendicular bisectors within the vertebral body. As such, facet joint orientation cannot be used to determine the center of twist.

As the center of lower cervical rotation appears to be on the midsaggital line, we defined the center of rotation as the point along the midsaggital line which showed the minimum torsional stiffness [17]. This method was adopted because it complies with the thermodynamic concept that statically indeterminant structures distribute load to seek a minimum strain energy. This occurs when the stiffness is a minimum. This concept provides a repeatable means for establishing a center of rotation for torsional responses of the lower cervical spine.

Based on these considerations, the purpose of this paper is to define the kinematic and kinetic responses of the passive elements of the cervical spine in torsion, to compare the responses with the Hybrid III Anthropometric Test Dummy (ATD) neck simulator, to develop modifications of the Hybrid III neckform to better replicate the response of the human cervical spine in torsion, and to compare the torsional responses of the cadaver cervical spine with existing volunteer data.

METHODS

SPECIMEN TYPES AND PROCUREMENT – Unembalmed human cervical spines were obtained shortly after death, sprayed with calcium buffered, iso-tonic saline, sealed in plastic bags, frozen and stored at –20 degrees Celsius. Cervical spine specimens included the base of the skull, approximately two centimeters around the foramen magnum, and the first thoracic vertebra at the caudal end. All ligamentous structures were kept intact, with the exception of the ligamentum nuchae. Medical records of donors were examined to ensure that the specimens did not show evidence of serious degeneration, spinal disease, or other health related problems that would affect their structural responses.

SPECIMEN PREPARATION – Prior to testing, each specimen was thawed at 20 degrees Celsius for 12 hours to place it in the fully equilibrated state. The pretest preparation was performed in an environment chamber which was designed to prevent specimen dehydration and deterioration. A variable flow humidifier pumped water vapor into the chamber to create a 100 percent humidity environment. The end vertebra were cleaned, dried and defatted for casting. Specimens were cast into aluminum cups with reinforced polyester resin so that the cup ends were parallel and the resting lordosis of the spine preserved. The cup centers were aligned along the center of the neural canal. The distance from the rostral cup center to the midpoint of the dens (d), and the distance from the caudal cup center to the anterior of the vertebral body (L) were recorded. During casting, the aluminum cups were cooled in a flowing water bath to dissipate the latent heat of polymerization.

TEST INSTRUMENTATION – Tests were conducted with a Minneapolis Testing System (MTS) servocontrolled hydraulic testing machine composed of a load frame with rotary actuator, a 25 gpm, 3000 psi hydraulic pump, two nitrogen filled accumulators and an angular feedback control system. Load was measured at the caudal end of the specimen using a six axis array of strain gauge load cells composed of two GSE three-axes ATD neck load cells and a GSE torsion cell, arranged to quantify force and moment in three orthogonal axes. A system to permit free changes in axial length of the specimen was implemented using a linear bearing to couple the rostral cup to the rotational actuator. Changes in axial length were quantified by a linear variable differential transformer (LVDT), which was connected to the system using a ball bearing adapter. Axial rotation was quantified using a rotational variable differential transformer (RVDT) mounted directly to the rotary actuator. An MTS digital function generator and controller were used to apply waveforms to the actuator at rates exceeding 500 degrees per second without overshoot. A dial gauge was used to align the specimen along the axis of twist and identify the location of the axis of minimum stiffness (see Figure 1). In addition, *in situ* flouroscopic images were recorded on videotape.

A digital measurement and analysis system was developed utilizing a data logging computer to record the eight channels of transducer output. The multichannel microcomputer data acquisition system incorporated an RC Electronics ISC-67 Computerscope for the digitization and storage of data. This system, which consists of a 16-channel A/D board, external instrument interface box, and Scope Driver software, has a 1 MHz aggregate sampling rate capability with 12 bit resolution and writes data directly to a hard disk.

EXPERIMENTAL METHODS – Pretest A-P and lateral radiograms were performed prior to casting. The specimen was mounted in the load frame and aligned to place the center of the dens along the axis of twist. The lower cervical spine was mounted to align the anterior portion of the disc of the lowest motion segment along the axis of twist as an estimate of the lower cervical center of twist. A cyclic test was performed using a 1 Hz haversine for 50 cycles to exercise the specimen and place it in a mechanically stabilized (reproducible) state [23]. Angle of twist was estimated to produce 10 to 20 percent of the expected load to failure.

A minimum stiffness protocol was performed to identify the axis of twist in the lower cervical spine. The lower spine was mounted such that the axis of twist lay on the midsaggital line anterior to the vertebral body. A ramp and hold rotation was applied over 0.5 seconds, and the dynamic torsional stiffness (K=torque/twist angle ($K = T/\theta$)) recorded. This was repeated two times and the results averaged. The specimen was then moved to realign the axis of twist 0.508 cm (0.200 inches) posterior from its previous position and the stiffness tests were repeated. The procedure was performed from the anterior of the vertebral body through to the center of the neural arch. A third-order polynomial was least squares-fitted to the data to accurately determine the center of minimum stiffness along the midsaggital line. This point was identified as the lower cervical center

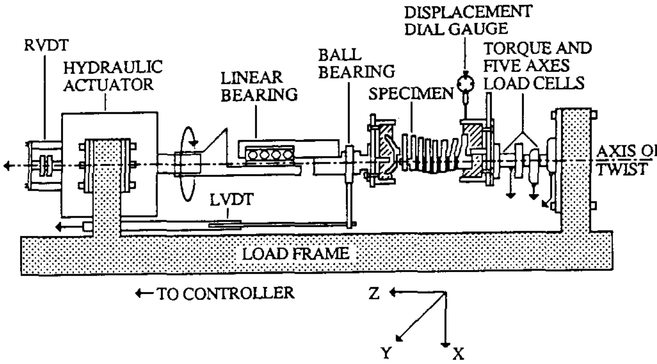

Figure 1: Test Apparatus

of rotation and placed along the axis of twist for the remaining tests.

A viscoelastic test battery was performed using relaxation and constant strain rate tests. The specimen was then loaded to failure by applying a ramp-and-hold at approximately 500 degrees/second. Magnetic resonance and CT images were performed to identify bony and ligamentous injuries to the specimen.

Since all failures were confined to the atlantoaxial joint, the joint was dissected and the failures described. The specimens were recast at the level of the axis, the viscoelastic test battery repeated, and a second failure test performed. Failures were again documented by magnetic resonance, CT, and dissection. As the failure tests provided the greatest quantity of load deflection information, they were used to determine the incremental stiffness.

In order to assess the performance of the Hybrid III dummy neck, the test battery, including viscoelastic tests, and minimum stiffness protocol, was performed on the Hybrid III. Finally, to assess the viability of modification to the Hybrid III, a rotational element was added to the rostral portion of the neck to emulate the behavior of the atlantoaxial joint, and the "Modified" Hybrid III was tested. Both the modified neck and the Hybrid III neck were then assessed in flexion-extension using a previously described load frame [1] to assess the influence of the rotational element on the dummy neck's biofidelity in the saggital plane.

TEST RESULTS

A total of 6 full cervical, and 6 lower cervical test batteries were performed.

KINEMATICS – Increases in axial length (z direction) were observed with rotation from the neutral position. Initial tests which rigidly constrained the axial length resulted in large axial forces, facet joint binding and non-physiologic failures. Subsequent testing using the linear bearing system described above allowed for free axial growth and more physiologic kinematics.

The upper cervical center of rotation was the dens. The center of rotation in the lower cervical spine was identified for each specimen using the minimum energy hypothesis, as defined by the X axis in Figure 2. A sample plot of dynamic torsional stiffness ($K/K_{\max}$) versus normalized midsaggital position (X/L) is shown in Figure 3. The mean center of rotation was found to lie at $X/L = 0.83 \pm 0.16$. Referenced against the vertebral body, the mean center of rotation was found to lie about a point approximately 1/5 the length of the vertebral body from the anterior of the vertebral body (i.e. $b/B = 0.20$, see Figure 2). Not surprisingly, large flexion-extension moments and anterior posterior shearing forces were observed during rotations about centers other than the center of minimum stiffness.

The center of rotation of the Hybrid III neckform was assessed. Figure 4 compares the minimum stiffness data for the Hybrid III neckform with the cadaver. The center of minimum stiffness in the Hybrid III neckform compared poorly with the cadaver, and was found to lie in the posterior portion of the vertebral body ($X/L = -0.15$). The torsional stiffness of the Hybrid III was also noted to be relatively insensitive to the selection of the axis of twist (Figure 4).

VISCOELASTIC TESTS – Relaxation tests were performed using ramp-and-hold command signals with 0.25 second rise times. The deflection was then held constant for the next 150 seconds. A typical response is shown in Figure 5. The Hybrid III shows similar relaxation behavior, Figure 6. The relaxation behavior suggests the presence of viscoelastic elements in both the Hybrid III neckform and the human cervical spine. Using linear viscoelastic theory, the dependence of amplitude on frequency could be predicted using hereditary integrals. To study this behavior, constant velocity tests were conducted on the mechanically stabilized spines using triangular wave deformations at frequencies of 0.02, 0.2 and 2.0 Hz. The maximum ram displacement was the same in all tests. The Hybrid III neckform, like many viscoelastic structures, demonstrates frequency

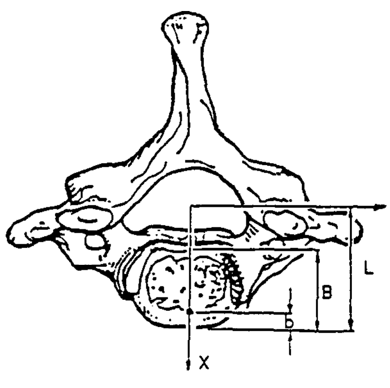

Figure 2: Reference Axes

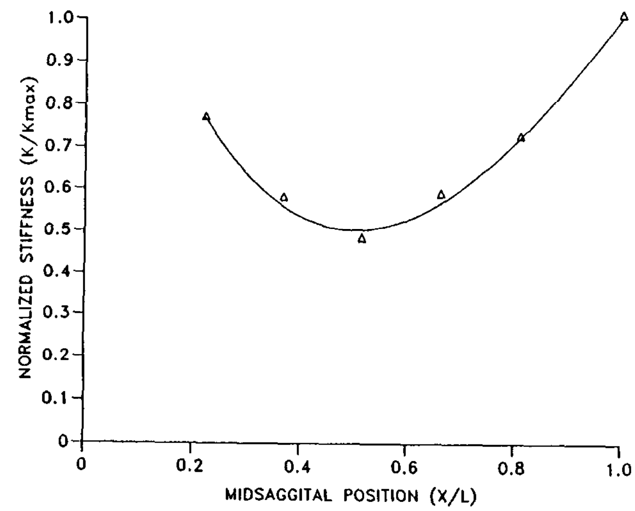

Figure 3: Typical Torsional Stiffness of a Human Cervical Spine

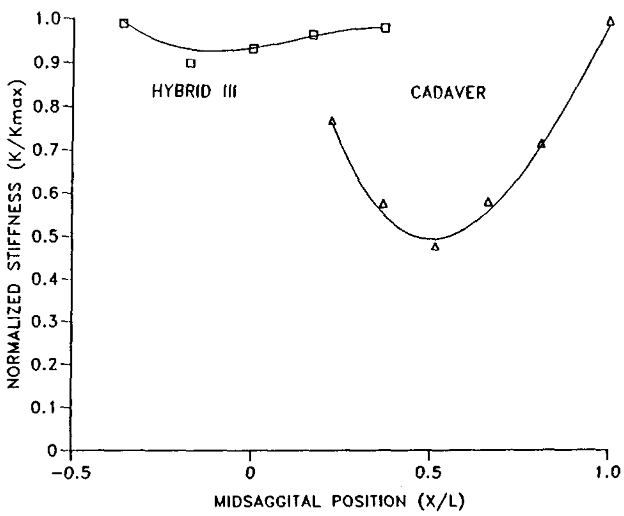

Figure 4: Torsional Stiffness of the Hybrid III neckform and a Typical Human Cervical Spine

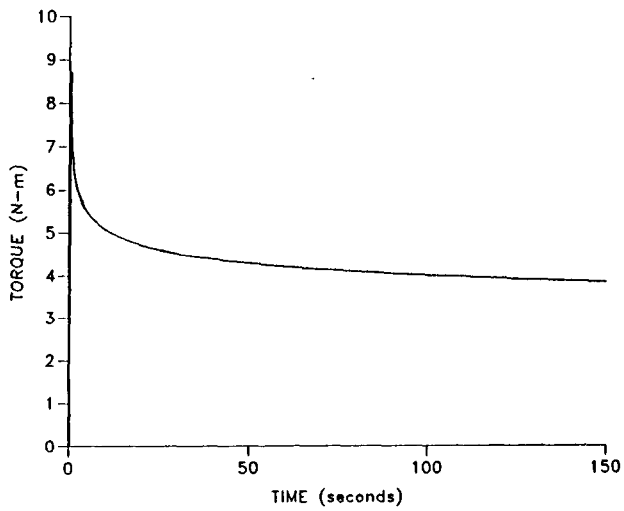

Figure 5: Typical Torque Relaxation for a Human Cervical Spine

dependence (Figure 7). In contrast, the torsional response of the cervical spine is virtually frequency independent (Figure 8). This nonlinear behavior has been observed in other modes of loading [1], and other biologic tissues [24]. This combination of relaxation and frequency independence obviates the use of linear viscoelastic theory, and requires the use of the more complex Maxwell-Weichert quasi-linear model.

FAILURE TESTS – High velocity failure tests were performed using ramp-to-failure constant velocity displacements. The purpose of these tests was to provide a database representing the lower bound (no muscle action) of the stiffness of the human neck in rotation. Figure 9 shows the load to failure of 5 cervical spines (solid lines). This data was least squares fitted to a piecewise linear model (the center dotted line) with an initial zero stiffness region, and a high stiffness region with a mean stiffness of 0.472 N-m/degree beginning at 66.8 degrees of rotation. This is bounded above and below by dotted lines representing one standard deviation from the mean (±0.147 N-m/degree and ±6.18 degrees). Using this piecewise model, the performance of the Hybrid III neckform was assessed (Figure 10). The Hybrid III neckform was considerably stiffer than the human, K=3.15 N-m/degree at 2.0 Hz with no initial low stiffness region. This is an expected result, as the Hybrid III neckform was not designed for axial rotation.

To assess the viability of combining axial rotation with existing flexion-extension and lateral bending neck simulators, the Hybrid III was modified. A purely rotational element was added to the rostral portion of the Hybrid III neck, adjacent to and caudal to the atlas. The element was rigid in all other directions of motion. The purpose of this element was to emulate the behavior of the atlantoaxis. This design was chosen both because it could be implemented without significantly altering the form of the Hybrid III neckform, and because

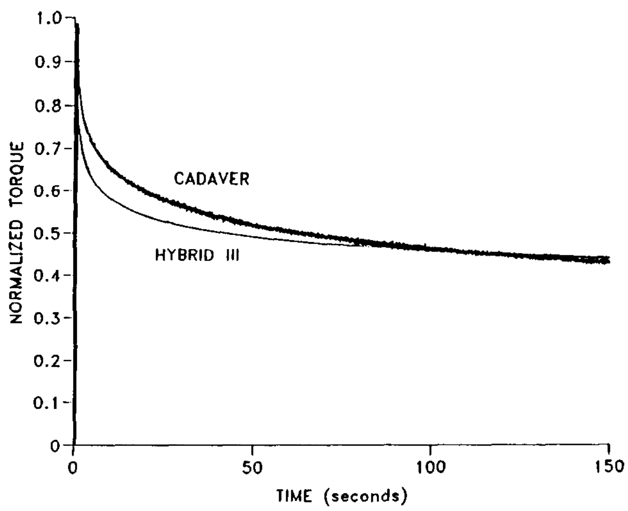

Figure 6: Torque Relaxation of the Hybrid III Neckform and a Typical Human Cervical Spine

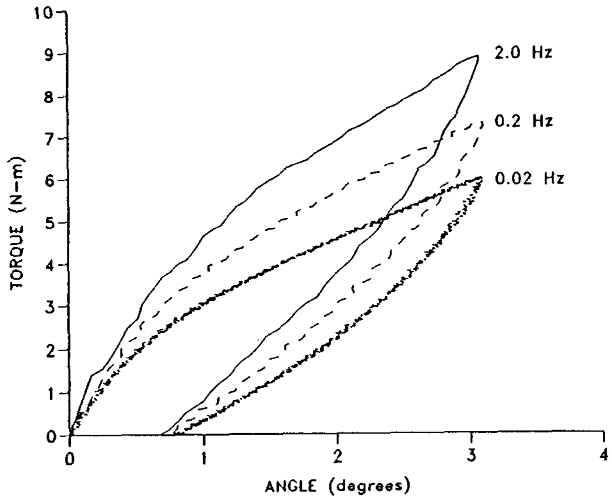

Figure 7: Constant Velocity Profile of the Hybrid III neckform

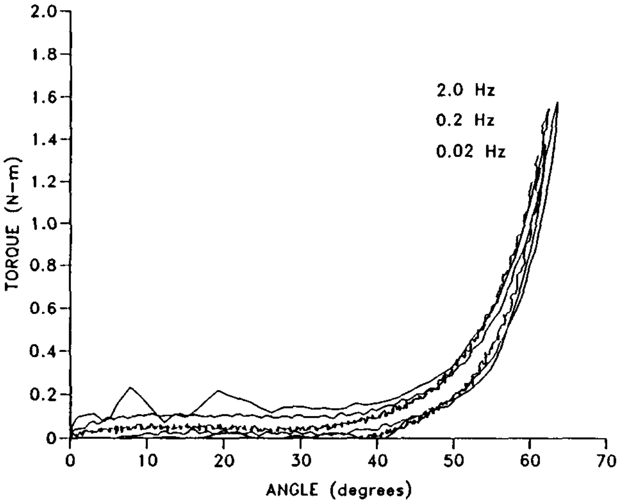

Figure 8: Typical Constant Velocity Profile of a Human Cervical Spine

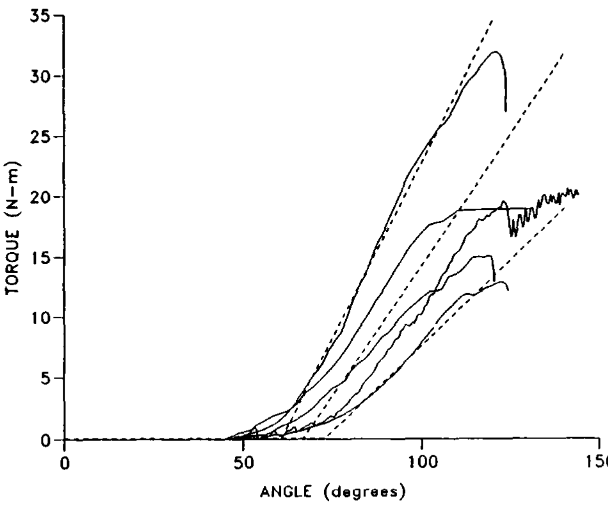

Figure 9: Failure Tests and Piecewise Linear Model of Human Cervical Spine Response

it better reflected the kinematics of the cervical spine. The latter, on the recognition that the atlantoaxis accounts for most of the low stiffness region of the cervical spine because of the relative laxity of both the bony and ligamentous constraints of that joint.

Figure 10 shows the effect of the addition of the atlantoaxis simulator to the performance of the Hybrid III. Denoted as the Modified Hybrid III, the figure shows the behavior of the modified neck against the Hybrid III, and against the cadaveric response window demarcated in Figure 9. To ensure that the modification did not compromise the performance of the simulator in the saggital plane, combined flexion and compression loading was applied. The results, shown in Figure 11, demonstrate no significant difference in the performance of the Hybrid III neckform and the Modified Hybrid III in the saggital plane.

To assess the influence of muscles on the performance of the neck in rotation induced by +Gy decelerations, the load deflection responses of the cadaver were compared to volunteer sled tests. The latter, shown together with the mean cadaveric response in Figure 12, was derived from data listed in Wismans and Spenny [12]. (This data is presented with permission of the 1983 Society of Automotive Engineers, Inc.)

DISCUSSION

KINEMATICS – The use of a dynamic test system affords many advantages in the study of the time dependent responses of the spine. Non-physiologic failures in specimens in which free axial growth was not permitted is a good example of this. Duplication of *in vivo* kinematics in the dynamic test environment is therefore a primary goal upon which all other work is based.

The center of rotation is one such kinematic parameter. Considering the spine as a statically indeterminant structure, we applied the minimum energy theorem [25] to define the physiologic center of rotation. This theory implies that motions resulting from an unconstrained structure are such that the strain energy of the structure is a minimum. As the elastic energy for a given angle of twist is a linear function of the stiffness, the minimum energy, and hence unconstrained axis of twist, occur at the point of minimum stiffness. The dynamic stiffness $K = T/\theta$, was used to establish this minimum and the preferred axis of twist. The parabolic shape of the stiffness versus axis of twist curves (Figure 3) allowed for easy and accurate identification of the preferred axis of twist in each of the specimens tested. However, inter-specimen variation, and the presence of large forces at non-physiologic axes of twist, require the identification of the minimum energy axis for each specimen. Finally, the validity of the minimum stiffness method is supported by its agreement with the center of rotation determined in earlier static domain radiographic studies [12,13].

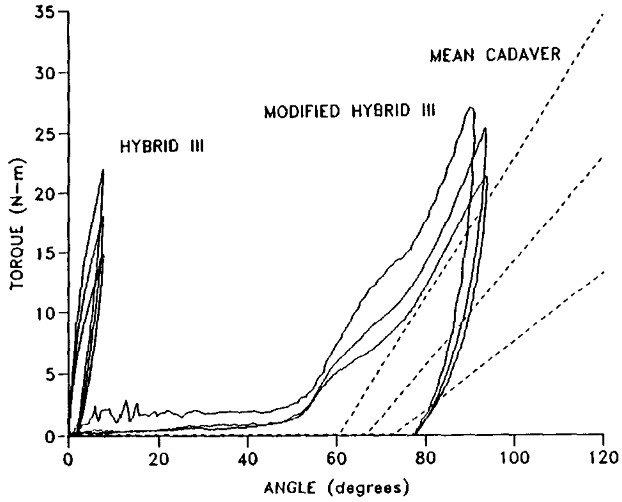

Figure 10: The Effect of the Atlantoaxial Simulator on the Hybrid III neckform Torque Response

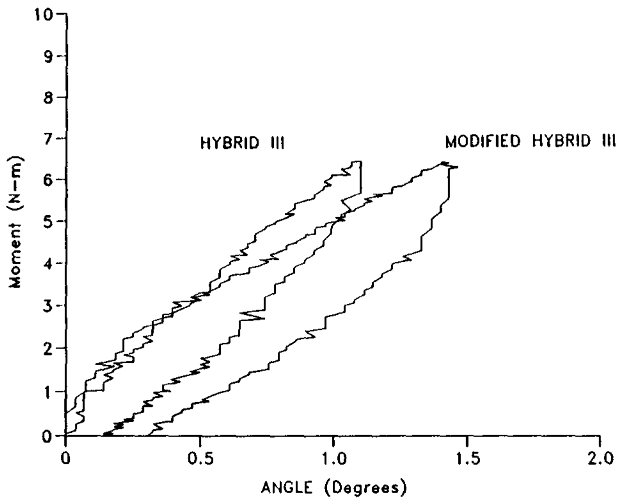

Figure 11: Bending Reponse of the Hybrid III and the Modified Hybrid III neckforms

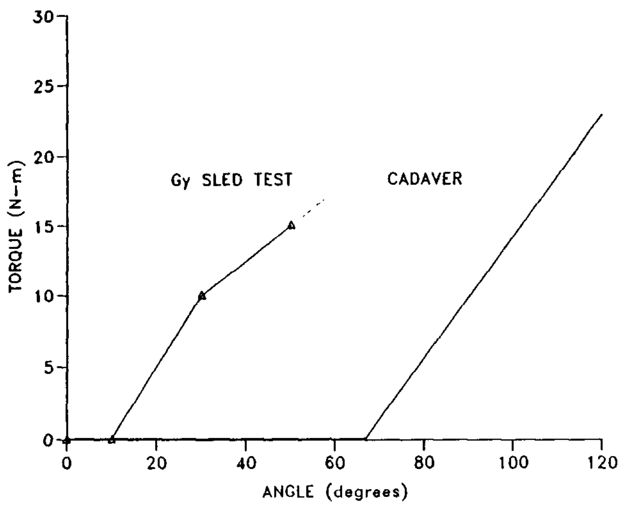

Figure 12: Passive and Active Responses of the Human Cervical Spine

KINETICS – Review of the literature has identified significant axial rotation of the head in response to lateral impacts. It has also implicated axial rotation as a contributor in determining the site of impact of the head during head-steering-wheel impact in purely frontal collisions of three-point seat belted occupants. Further, as angular acceleration is a known mediator of central nervous system injury, the possible contribution of axial rotation to brain injury is also in question.

Thus, because of its importance, this paper has investigated the performance of the cadaver neck in axial rotation. Results suggest that a piecewise linear model with an initial load free region, followed by a high stiffness region is an adequate model of the passive response of the neck to rotation. The stiffness in this region had a mean value of 0.472 N-m/degree, beginning at an initial value of 66.8 degrees of axial rotation from the neutral position.

Comparison of the Hybrid III neckform with the cadaver cervical spine in torsion has shown that the Hybrid III neckform shows no initial low stiffness region, and is nearly an order of magnitude stiffer than the cadaver at the frequencies of interest. The results also suggest that significant improvements in the biofidelity of the ATD neck in torsion can be achieved, without compromise of the performance of the neck in the saggital (frontal) plane by the addition of a removable, modular "atlantoaxial" element. We are not advocating the use of this particular design in car crash simulation with dummies because of the significant reduction in repeatablity that would occur. However, the results presented here suggest that design and implementation of such a device is possible, and should be considered for improved biofidelity in situations where accurate torsional response of the ATD is desired.

Finally, for the purpose of simulation, both cadaver tests, and volunteer deceleration testing are routinely performed. Both models, however, suffer in that they do not exactly reflect the human response to the car crash environment. The former fails in that it represents the response of the passive spinal elements only. The latter fails in that the effects of the musculature are overemphasized because of the volunteers awareness of the upcoming event. The "true" biological response probably lies between the two. As such, these two models may be used to define a window for simulator performance. Figure 12 represents such a window.

CONCLUSIONS

1. A dynamic test apparatus which recreates *in vivo* kinematics has been developed to study the time dependent passive responses of the cervical spine in torsion.

2. Interspecimen variation and the development of non-physiologic forces at incorrect centers of rotation, create the need for accurate identification of the lower cervical center of twist for each specimen tested. The minimum stiffness method is suitable for this purpose.

3. The mean center of lower cervical rotation is located in the anterior portion of the vertebral body at a point 1/5 the A-P length of the body from the anterior of the body.

4. The responses of the cadaver neck in rotation may be modeled by a piecewise linear model of zero initial stiffness, and 0.472 N-m/degree stiffness region beginning at 66.8 degrees from the neutral plane.

5. The Hybrid III neck was found to be stiffer than the human in axial rotation, and did not exhibit an initial low stiffness region.

6. The Hybrid III was modified to accommodate an atlantoaxial joint. The response of this modified neck showed improved biofidelity in rotation without significantly altering performance in the saggital plane. The results suggest that addition of an atlantoaxis simulator is viable, and should be considered in situations where axial rotation is considered important.

7. Comparison of cadaver tests with volunteer acceleration tests provides information on the performance of the musculature, and may be used as a window to define the performance of simulators.

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