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Human Mechanical Impact Response Characteristics - Response of the Human Neck to Inertial Loading by the Head for Automotive Seated Postures

RATIONALE

The members of the SAE Human Biomechanics and Simulations Standards Steering Committee have reviewed J1460-2 and made a conscientious decision to stabilize this Information Report. Additional research into the inertial loading response of the human neck has been conducted since this Information Report was last revised. The neck design of the Hybrid III family of dummies was developed based on the data included in J1460-2. This Information Report has historical value.

STABILIZED NOTICE

This document has been declared "Stabilized" by the SAE Human Biomechanics and Simulations Standards Steering Committee and will no longer be subjected to periodic reviews for currency. Users are responsible for verifying references and continued suitability of technical requirements. Newer technology may exist.

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Foreword—Human mechanical impact response is defined as the reaction of a body segment in terms of measurable engineering parameters such as forces, accelerations, and deflections due to direct or indirect impact loading. The impact response of a body region can depend upon the test conditions used to generate the data (such as impactor shape, stiffness, mass, and body region boundary conditions) and thus, in most cases, can only be defined in terms of those conditions. Accordingly, the impact response of a test dummy component must be evaluated under test conditions similar to those used to obtain the defined response data.

A number of problems need to be addressed in defining human impact response characteristics, since most impact response studies use cadaver or animal surrogates to obtain data at severe impact levels. The impact response of surrogates can differ from that of living humans due to lack of physiological effects, such as muscle tone in the cadaver subject, and lack of geometric similitude with animal subjects. In cases where sub-injury tests are conducted with volunteers, there are also problems with extrapolation of the response data to represent the response at higher impact severities. Some studies only include response of the body region up to the injurious level while others include response well beyond the initiation of tissue structural failure associated with injury.

In addition, the human form is not of unique size, shape, and proportion. There are significant geometric differences between and among adults and children, and males and females. The available response data in the literature dictates that treatment of this topic be constrained essentially to guidelines for average adult male responses, with scaling used to define equivalent responses for the small adult female and the large adult male.

Finally, there may be response variability introduced by age, physical conditioning, and other factors not discussed here. This variability is discussed for those body regions where such information is available.

1. Scope—This series of reports provides response characteristics of the head, face, neck, shoulder, thorax, lumbar spine, abdomen, pelvis, and lower extremities. In each report, the descriptions of human impact response are based on data judged by the subcommittee to provide the most appropriate information for the development of human surrogates.

1.1 Purpose—This is one of a series of reports which define human mechanical impact response characteristics for specific body regions. These reports update SAE J1460 which is intended for use by anthropomorphic test dummy designers and analytical modelers who need quantitative definitions of human mechanical impact behavior. These reports do not discuss criteria for assessing human impact injury potential, which are the subject of SAE J885. Each document in the series covers material specific to a body region and will be independently updated when new response data become available. The goal of this report is to characterize the response of human neck due to head inertial loading when the occupant is sitting in an automotive posture.

2. References

2.1 Applicable Publications—The following publications form a part of this specification to the extent specified herein. Unless otherwise specified, the latest issue of SAE publications shall apply.

2.1.1 SAE PUBLICATIONS—Available from SAE, 400 Commonwealth Drive, Warrendale, PA 15096-0001.

SAE J211-1 MAR95—Instrumentation for Impact Test—Part 1—Electronic Instrumentation
SAE J211-2 MAR95—Instrumentation for Impact Test—Part 2—Photographic Instrumentation
SAE J885—Human Tolerance to Impact Conditions as Related to Motor Vehicle Design
SAE J1460 MAR85—Human Mechanical Response Characteristics
SAE J1733—Sign Convention for Vehicle Crash Testing

2.1.2 ISO/TR PUBLICATION—Available from ANSI, 11 West 42nd Street, New York, N.Y. 10036-8002

ISO/TR 9790-2—Road vehicles — Anthropomorphic side impact dummy — Part 2: Lateral neck impact response requirements to assess biofidelity of dummy

2.1.3 OTHER PUBLICATIONS

1. Mertz, H. J., "The Kinematics and Kinetics of Whiplash," Ph.D. Dissertation, Wayne State University, 1967.
2. Mertz, H. J. and Patrick, L. M., "Investigation of the Kinematic and Kinetics of Whiplash," SAE 670919, Eleventh Stapp Car Crash Conference, October 1967.
3. Mertz, H. J. and Patrick, L. M., "The Effect of Added Weight on the Dynamics of the Human Head," Final Report on Contract No. DAAG-17-67-C-0202, U.S. Army Natick Laboratories, Natick, Massachusetts, 1971.
4. Mertz, H. J. and Patrick, L. M., "Strength and Response of the Human Neck," SAE 710855, Fifteenth Stapp Car Crash Conference, November 1971.
5. Patrick, L. M. and Chou, C.C., "Response of the Human Neck in Flexion, Extension and Lateral Flexion," Vehicle Research Institute Report VRI 7.3, SAE, 1976.
6. Ewing, C. L., Thomas, D. J., Beeler, G. W., Patrick, L. M. and Gillis, D. B., "Dynamic Response of the Head and Neck of the Living Human to -Gx Impact Acceleration," SAE 680792, Twelfth Stapp Car Crash Conference, October 1968.
7. Ewing, C. L., Thomas, D. J., Patrick, L. M., Beeler, G. W. and Smith, M. J., "Living Human Dynamic Response to -Gx Impact Acceleration, Part II - Accelerations Measured on the Head and Neck," Thirteenth Stapp Car Crash Conference, December 1969.
8. Ewing, C. L. and Thomas, D. J., "Torque Versus Angular Displacement Response of the Human Head to -Gx Impact Acceleration," SAE 730976, Seventeenth Stapp Car Crash Conference, November 1973.
9. Ewing, C. L., Thomas, D. J., Lustick, L., Becker, E., Willems, G. and Muzzy, W. H., "The Effect of the Initial Position of the Head and Neck on the Dynamic Response of the Human Head and Neck to -Gx Impact Acceleration," SAE 751157, Nineteenth Stapp Car Crash Conference, November 197.
10. Ewing, C. L., Thomas, D. J., Lustick, L., Muzzy, W. H., Willems, G. and Majewski, P. L., "The Effect of Duration, Rate of Onset and Peak Sled Acceleration on the Dynamic Response of the Human Head and Neck," SAE 760800, Twentieth Stapp Car Crash Conference, October 1976.

11. Ewing, C. L., Thomas, D. J., Majewski, P. L., Black, R. and Lustick, L., "Measurement of Head, T1 and Pelvic Response to -Gx Impact Acceleration," SAE 770927, Twenty-First Stapp Car Crash Conference, October 1977.
12. Ewing, C. L., Thomas, D. J., Lustick, L., Muzzy, W. H., Willems, G. C. and Majewski, P., "Dynamic Response of the Human Head and Neck to +Gy Impact Acceleration," SAE 770928, Twenty-First Stapp Car Crash Conference, October 1977.
13. Ewing, C. L., Thomas, D. J., Lustick, L., Muzzy, W. H., Willems, G. C. and Majewski, P., "Effect of Initial Position on the Human Head and Neck Response to +Y Impact Acceleration," SAE 780888, Twenty-Second Stapp Car Crash Conference, October 1978.
14. Zaborowski, A. V., "Lateral Impact Studies," Ninth Stapp Car Crash Conference, October 1965.
15. Hu, A. S., Bean, S. P. and Zimmerman, R. M., "Response of Belted Dummy and Cadaver to Rear Impact," SAE 770929, Twenty-First Stapp Car Crash Conference, October 1977.
16. Tarriere, C., "Proposal for Lateral Neck Response Requirements for Severe Impact Conditions," ISO/TC22/SC12/WG5 Document N166, 1986.
17. Ferlic, D., "The Range of Motion of the Normal Cervical Spine," Hopkins Hospital Bulletin 110, 1962.
18. Buck, C.A., Dameron, F. B., Dow, M. J. and Skowlund, H. V., "Study of Normal Range of Motion in the Neck Utilizing a Bubble Goniometer," Archives of Physical Medicine and Rehabilitation, Volume 40, September 1959.
19. Granville, A. D. and Kreezer, G., "The Maximum Amplitude and Velocity of Joint Movements in Normal Male Human Adults," Human Biology, Volume 9, 1937.
20. Foust, D. R., Chaffin, D. B., Snyder, R. G. and Baum, J. K., "Cervical Range of Motion and Dynamic Response and Strength of Cervical Muscles," SAE 730975, Seventeenth Stapp Car Crash Conference, November 1973.
21. Schneider, L. W., Foust, D. R., Bowman, B. M., Snyder, . G., Chaffin, D. B., Abdelnour, T. A. and Baum, J. K., "Biomechanical Properties of the Human Neck in Lateral Flexion," SAE 751156, Nineteenth Stapp Car Crash Conference, November 1975.
22. Mertz, H. J., Neathery, R. F. and Culver, C. C., "Performance Requirements and Characteristics of Mechanical Necks," Human Impact Response - Measurement and Simulation, Plenum Press, NY, 1973.
23. Road Vehicles - Anthropomorphic Side Impact Dummy - Part 2 - Lateral Neck Impact Response Requirements to Assess Dummy Biofidelity, ISO Technical Report TR9790-2, 1989.
24. Mertz, H. J., Irwin, A. L., Melvin, J. W., Stalnaker, R. L. and Beebe, M. S., "Size, Weight and Biomechanical Impact Response Requirements for Adult Size Small Female and Large Male Dummies," SAE 890756, SP-782, March 1989.

2.2 Related Publications—The following publications are provided for information purposes only and are not a required part of this document.

Anatomy, Descriptive and Surgical, Gray, H., Bounty Books, N.Y., N.Y., 1977.

3. Introduction—In automotive collisions, the neck plays an important role in controlling the kinematics of the head. The primary neck characteristics affecting the resulting head kinematics are its length, range of articulation relative to the automotive seated posture and degree of bending resistance produced by active and passive muscle reactions and by ligament stretch. For a crash test dummy to have human-like head motion, its neck structure must be designed to mimic these important characteristics. In addition, the crash test dummy needs to be instrumented to measure internal neck loads to assess the potential for neck injury. This report will provide a summary of studies conducted to quantify neck response characteristics produced by fore, aft, and lateral head inertial loading. Performance requirements for judging neck biofidelity based on these data will be given. The SAE sign convention defined in SAE J211 and SAE J1733 is used in this report (see Appendix A). Response of the neck to pure tension, compression, torsion, shear, or direct impact will not be discussed.

4. ***Neck Anatomy***—The primary load-carrying structures of the neck are the cervical spine, intervertebral disks, ligaments, and muscles. The cervical spine consists of seven bony vertebrae commonly referred to as C1 through C7 (see Figure 1). The top two vertebrae have unique shapes and functions. The top vertebra, C1, is called the atlas, and is shown in Figure 2. It supports the head at two bony articular surfaces called the occipital condyles which are symmetrically located on the base of the skull on each side of the foramen magnum, the hole through which the spinal cord passes, (see Figure 3). This joint allows the head to nod fore-aft relative to C1 without unduly stretching the spinal cord. The second cervical vertebra, C2, is called the axis and is shown in Figure 4. It is characterized by a bony protuberance, the odontoid process, which extends upwards into the atlas. The odontoid process forms an axis for head rotation as in the "no" gesture. The third through seventh cervical vertebrae, C3 through C7, are similar to the other vertebrae of the spine with their size increasing as one proceeds down the spine, (see Figure 5). These vertebrae have smooth superior and inferior bony surfaces called facets which form articular joints with adjacent vertebrae. The bodies of C2 through C7 are separated by fluid-filled fibrous pads called intervertebral disks. The amount of relative motion that can occur between adjacent vertebrae is limited by the articular joints and intervertebral disks as well as by fibrous tissue called ligaments which bind adjacent vertebrae together. Figure 6 is a lateral cross-sectional view of the base of the skull and of C1 and C2 which shows various ligaments, articular joints, intervertebral disks, and the spinal cord canal. Figure 7 is a posterior cross-sectional view of the base of the skull and of C1 and C2 showing the ligamentous connections.

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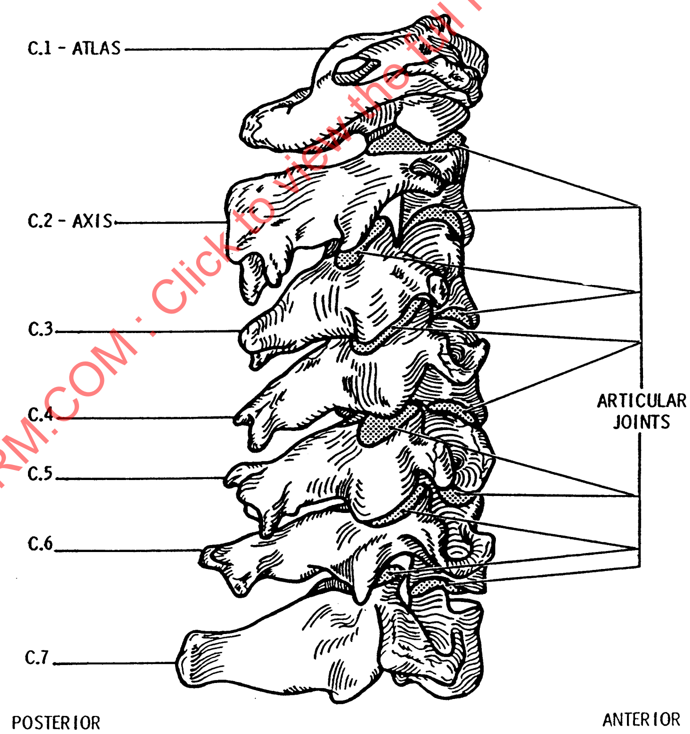

FIGURE 1—LATERAL VIEW OF THE CERVICAL VERTEBRAE SHOWING THE LOCATIONS OF THE ARTICULAR JOINTS

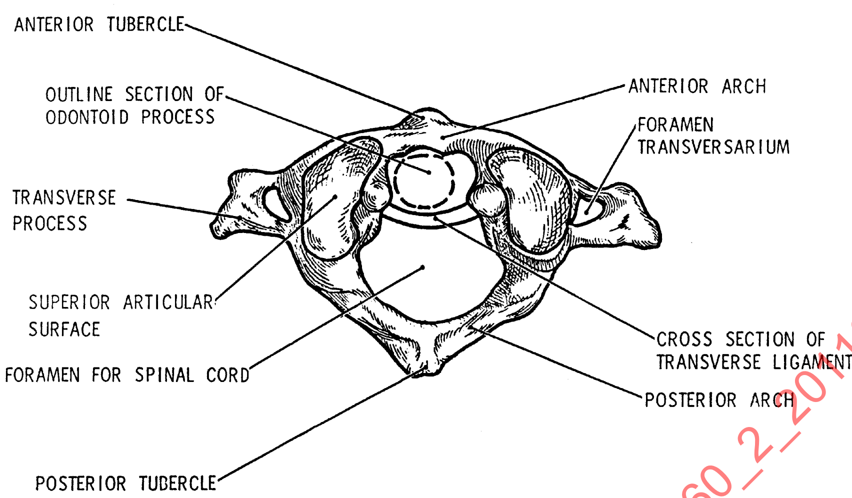

FIGURE 2—SUPERIOR VIEW OF THE FIRST CERVICAL VERTEBRAE, THE ATLAS

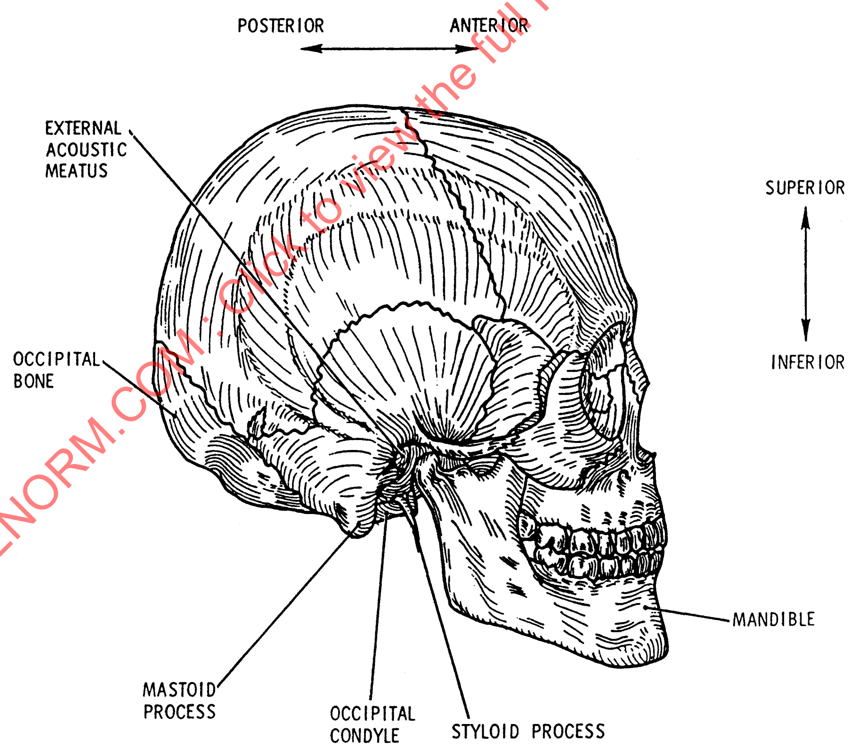

FIGURE 3—POSITION OF THE OCCIPITAL CONDYLES RELATIVE TO LANDMARKS OF THE SKULL

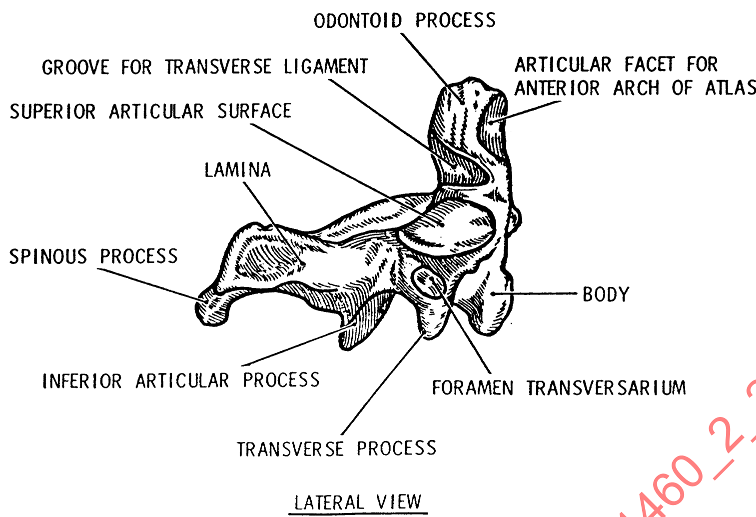

FIGURE 4—THE SECOND CERVICAL VERTEBRA, THE AXIS

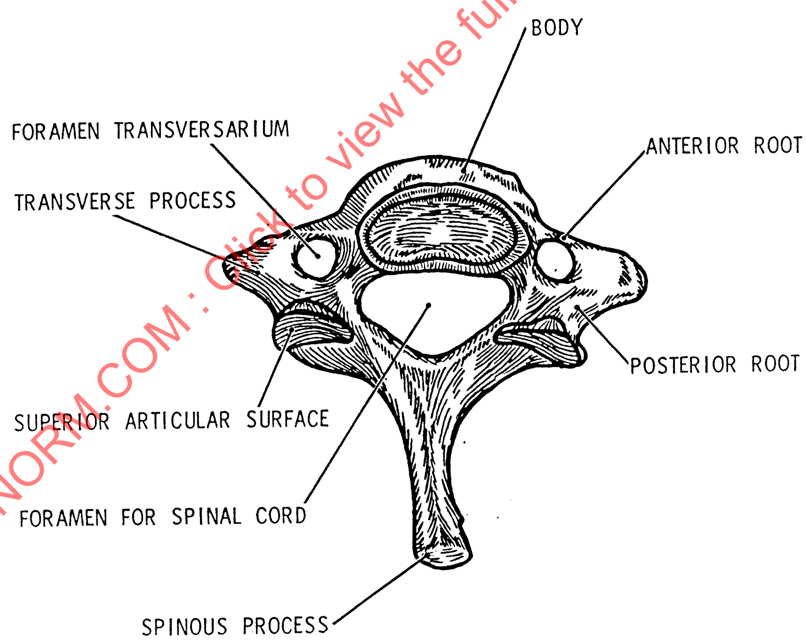

FIGURE 5—SUPERIOR VIEW OF THE SEVENTH CERVICAL VERTEBRA

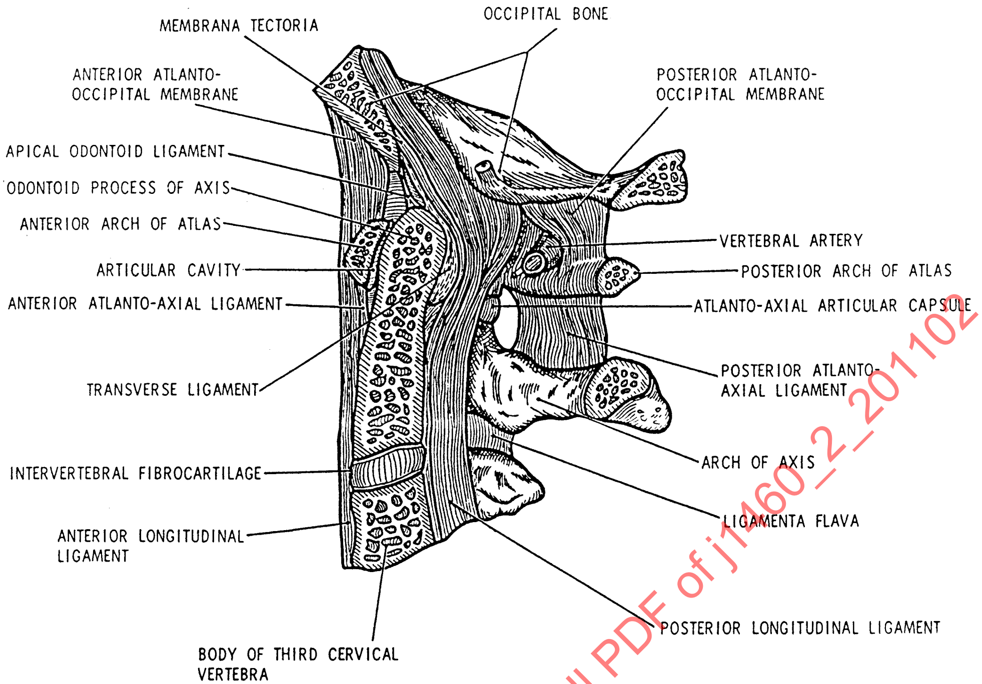

FIGURE 6—MIDSAGITTAL SECTION OF THE OCCIPUT, ATLAS, AND AXIS SHOWING THE LIGAMENTOUS CONNECTIONS

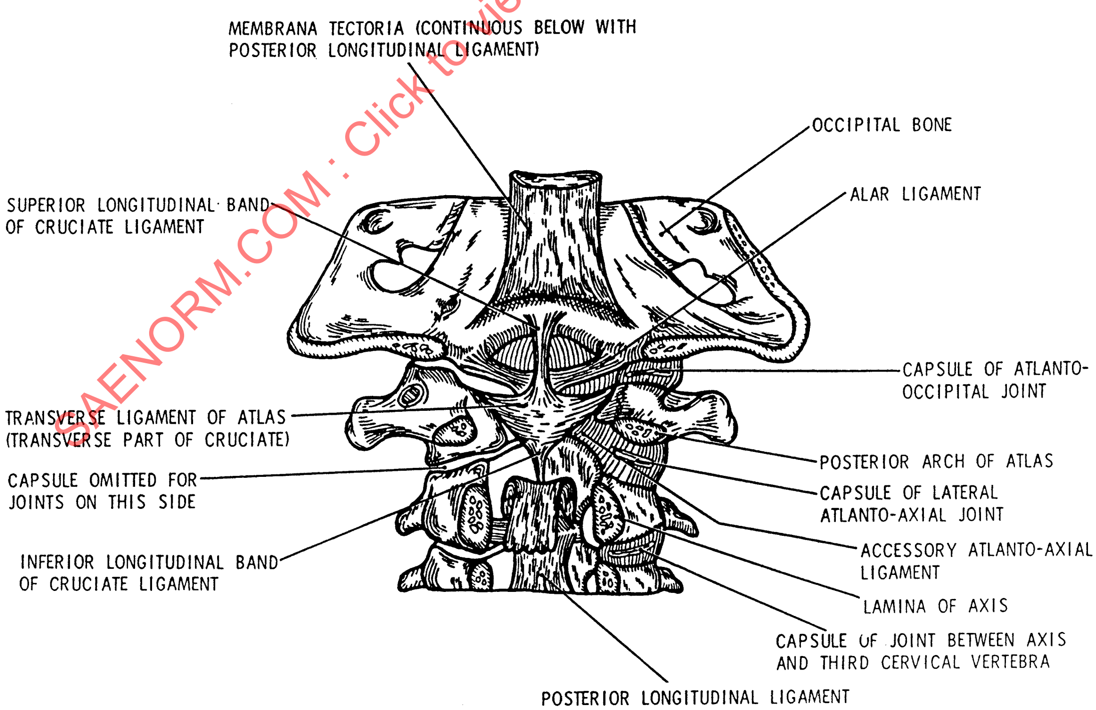

FIGURE 7—POSTERIOR VIEW OF THE OCCIPUT, ATLAS, AND AXIS SHOWING THE LIGAMENTOUS CONNECTIONS

Voluntary movements of the head and the cervical spine are produced by action of the neck muscles. Neck muscles occur in pairs which are symmetric with respect to the sagittal plane, the plane of symmetry of the body, and are attached to the head, individual cervical vertebrae, and/or the torso, (see Figures 8, 9, and 10). It is primarily the forces developed by the neck muscles, active and passive, that produce the bending stiffness of the neck within its voluntary range of motion. The neck ligaments contribute little to the bending resistance within this range. At the extremes of voluntary head-to-torso articulation, significant passive forces are developed in the ligaments which then add to the bending resistance produced by the muscles. Since the majority of the neck muscles are located posterior to the cervical spine (see Figure 11), the neck's bending stiffness is about three times greater in resisting forward bending than rearward bending. The neck's bending stiffness will vary with the degree of active muscle force and the rate of deformation of the muscle tissues which are viscoelastic. Without active or passive muscle force, the neck has very little bending resistance within the voluntary range of motion.

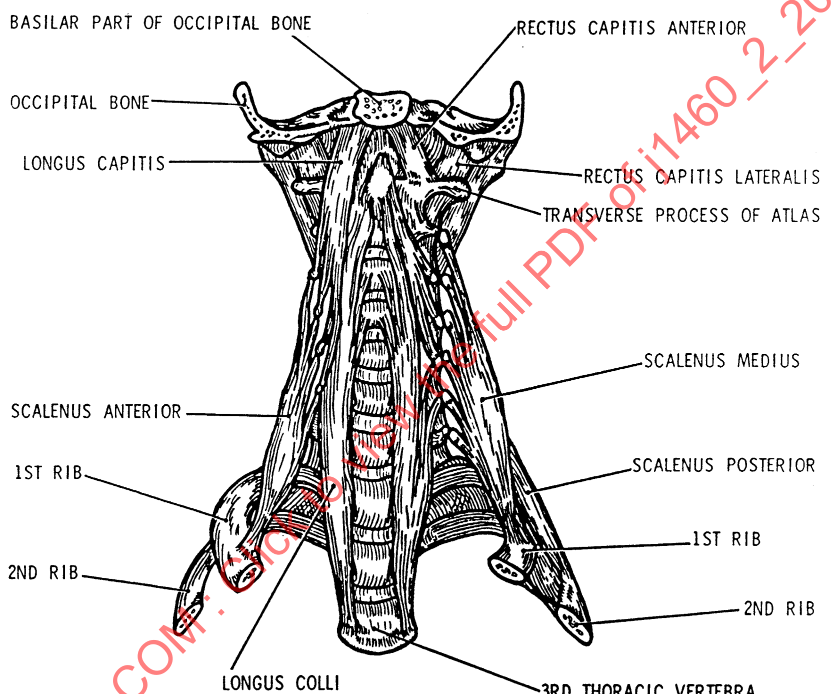

FIGURE 8—ANTERIOR VIEW OF THE PREVERTEBRAL CERVICAL MUSCLES

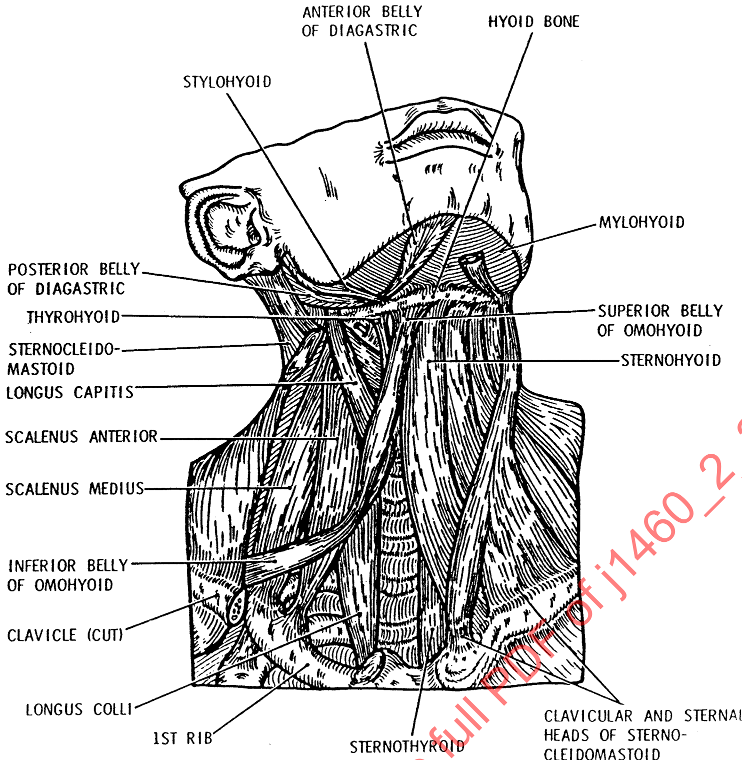

FIGURE 9—ANTEROLATERAL VIEW OF THE MUSCLES OF THE NECK

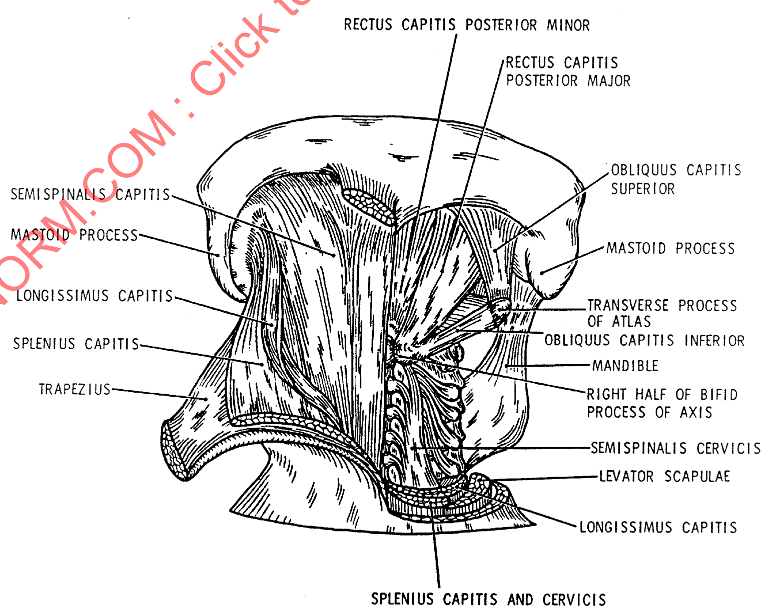

FIGURE 10—POSTERIOR MUSCLES OF THE NECK

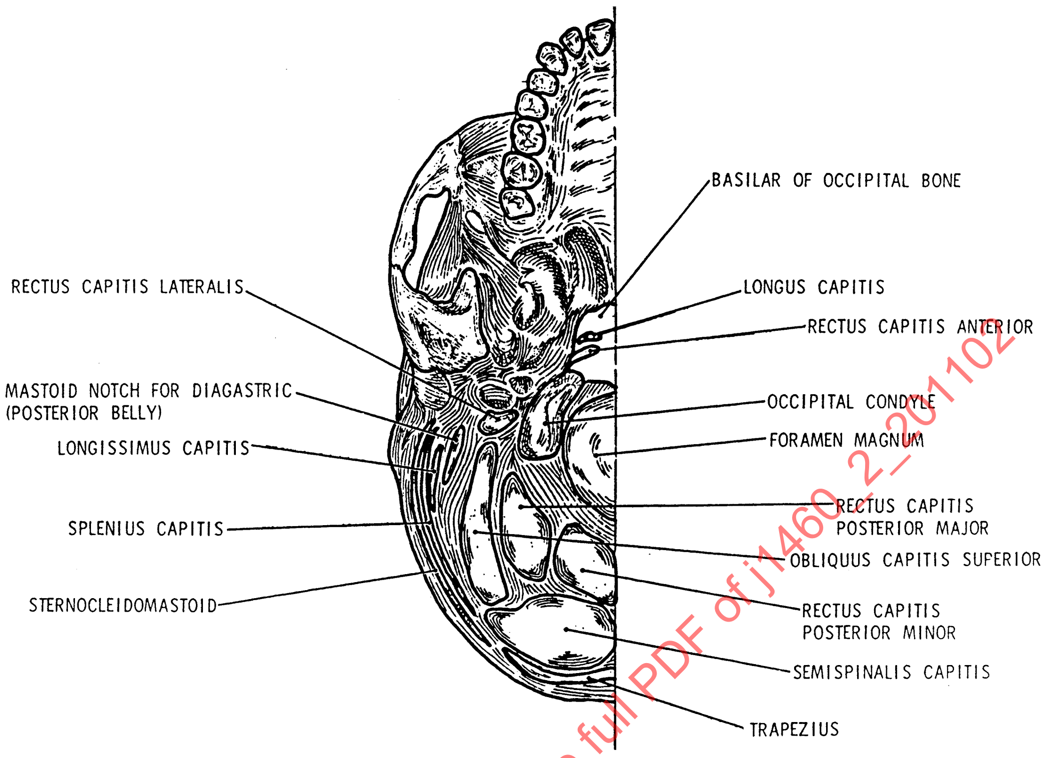

FIGURE 11—INFERIOR ASPECT OF THE SKULL BASE SHOWING NECK MUSCLE ATTACHMENT AREAS

5. *Studies of Inertial Loading of the Neck*

5.1 Mertz and Patrick Results—Mertz and Patrick (see 2.1.3.(1), 2.1.3.(2), 2.1.3.(3), and 2.1.3.(4)) conducted a series of sled tests using a single human volunteer and six embalmed cadavers to characterize the responses of their necks to head inertial loads that produced either flexion or extension of the neck. The volunteer was 49 years old and was fairly representative in height (1.73 m = 5'8") and mass (72.7 kg = 160 pounds) of the 50th percentile adult male. Tests were conducted with the volunteer seated in a rigid chair that simulated a typical automotive seated posture. It had a 15-degree seat back angle and a 5-degree seat pan angle. He was restrained by a lap belt and two shoulder belts that crisscrossed his chest. The belts were 50 mm wide. This restraint system minimized the amount of torso motion relative to the seat during the test. Accelerometers were affixed to his head using a tightly-fitting, light-weight skull cap and bite plate. Outputs from these transducers along with estimates of the inertial characteristics of his head were used to calculate the resultant loading of the neck's reaction on the head. These resultant internal neck reactions acting on the head at the head-neck interface consisted of orthogonal fore-aft shear and axial forces and a bending moment about a lateral axis which is tangent to the inferior surfaces of the occipital condyles. A photographic target and biaxial accelerometers were mounted to a bracket that was tightly strapped to his back at the level of the first thoracic vertebra. A target was also attached to the skull cap. These targets were used to determine the change in head position and orientation relative to the torso.

5.1.1 FLEXION TESTS—Tests were conducted at increasing levels of average sled deceleration, ranging from 3.0 G to 9.6 G. The volunteer was instructed to assume either a tensed or relaxed muscle state prior to sled deceleration. The neck's reaction on the head was characterized in terms of an internal bending moment acting on the head calculated about the occipital condylar axis as a function of the change in the angular position of the head relative to the torso. This characterization is independent of time and provides the relationship between two important response parameters, neck muscle loading and head angular position. Time histories of various parameters were not used to characterize the neck's response because they are very sensitive to the test set-up conditions. However, for the 9.6 G test, the peak neck tension was 789 N and occurred prior to the peak resultant resistive moment (–88.2 Nm) and peak shear force (–789 N). These latter two resultants include the effect of the chin's interaction with the chest.

Moment-angle responses for various sled deceleration levels that produced flexion are shown in Figure 12 for the condition of tensed muscles. The ordinate is the resultant, internal neck moment that is acting on the head and that is calculated about a lateral axis (y-axis) that passes through the occipital condyles. The abscissa is angle of the head relative to the torso. Note that the volunteer anticipated the sled deceleration by rotating his head rearward. Mertz and Patrick did not provide the negative moment-angle data that occurred early in the event since they were mainly concerned with the resistance of the neck to forward flexion. Figure 13 shows a comparison of the neck's moment-angle responses in forward flexion for two 4 G sled tests, one conducted with the neck muscles tensed to minimize head rotation and the other with the neck muscles relaxed which allowed much more rotation. Note that the maximum neck moment in the "relaxed" test was only slightly lower than the maximum moment in the "tensed" test indicating that the volunteer was controlling the degree of rotation of his head through active muscle forces even in the "relaxed" test. Active muscle response was possible since the tests were conducted on a decelerator sled allowing time for the volunteer to anticipate the start and severity of the deceleration pulse.

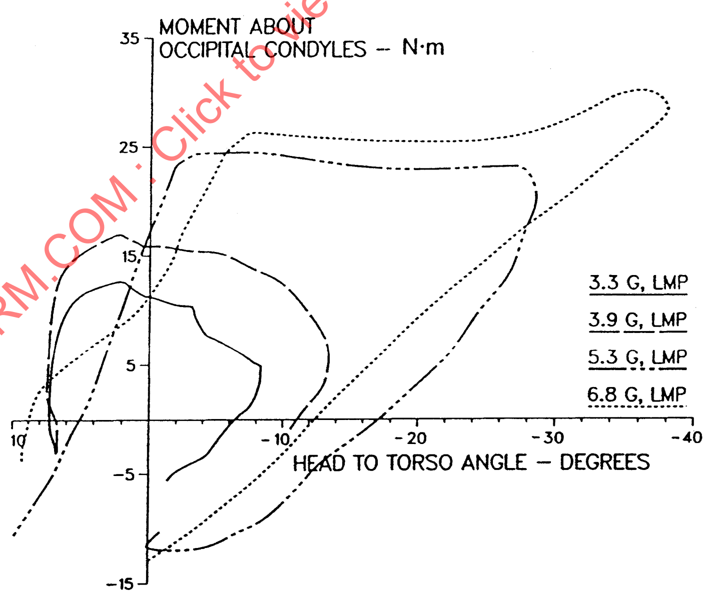

FIGURE 12—MOMENT-ANGLE NECK FLEXION RESPONSES OF VOLUNTEER LMP FOR VARIOUS SLED DECELERATION LEVELS. NECK MUSCLES TENSED (SEE 2.1.3.(4))

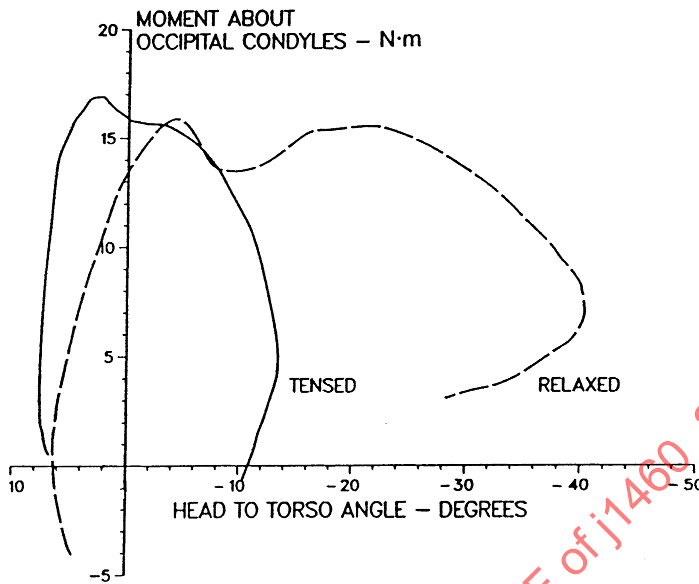

FIGURE 13—COMPARISON OF MOMENT-ANGLE NECK FLEXION RESPONSES FOR TWO 4 G SLED TESTS OF A HUMAN VOLUNTEER. ONE CONDUCTED WITH NECK MUSCLES “TENSED” TO MINIMIZE HEAD MOTION ANDTHE OTHER WITH NECK MUSCLES “RELAXED” ALLOWING HEAD TO TORSO MOTION (SEE 2.1.3.(4))

Tests were also conducted with lead masses attached symmetrically to the skull cap worn by the volunteer. The total mass added was 1.36 kg. The masses were positioned either above, below, or at the center-of-gravity of the head to determine the neck’s response to the added mass and its center-of-gravity location. As expected, for a given average sled deceleration level, the maximum moment applied to the head by the neck occurred when the masses were attached above the center-of-gravity of the head. The smallest maximum moment occurred when the lead masses were attached below the center-of-gravity of the head. For each location of the masses, the maximum shear forces applied to the head by the neck were the same. Shown in Figure 14 are curves of the resultant bending moment developed by the neck acting on the head as a function of the change of head to torso angle for various sled deceleration levels for the conditions of the masses positioned above the center-of-gravity of the head and neck muscles tensed. Again, the volunteer anticipated the sled deceleration by rotating his head rearward and the negative moment-angle data that occurred early in the event is not shown.

Mertz and Patrick also conducted similar sled tests using instrumented embalmed cadavers. They noted that the bending resistance of the cadaver’s neck was dependent on the degree of stiffness remaining in the embalmed neck muscle tissue. Figure 15 shows the moment-angle responses for a cadaver with a very stiff neck and a cadaver with a very flaccid neck. Note in the flaccid cadaver, appreciable bending stiffness is only developed at the extreme of the articular range where the neck ligaments are loaded.

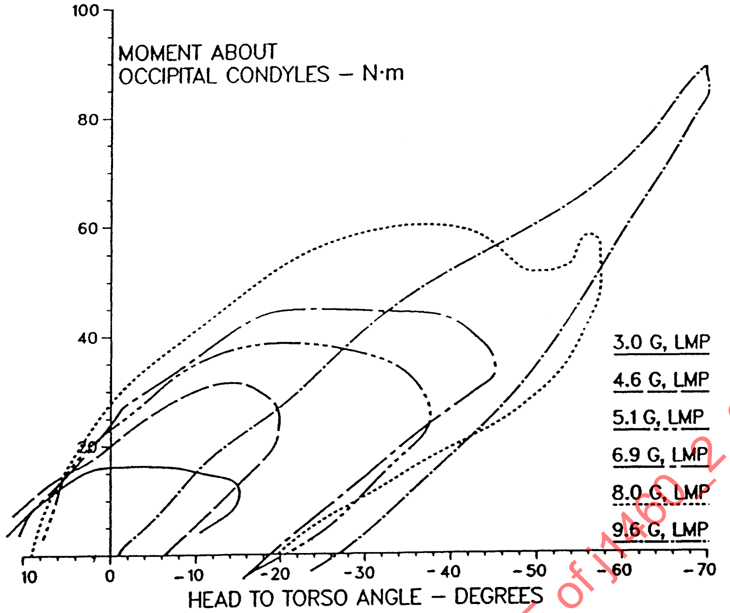

FIGURE 14—MOMENT-ANGLE NECK FLEXION RESPONSES OF VOLUNTEER LMP FOR VARIOUS SLED DECELERATION LEVELS. NECK MUSCLES TENSED (SEE 2.1.3.(4))

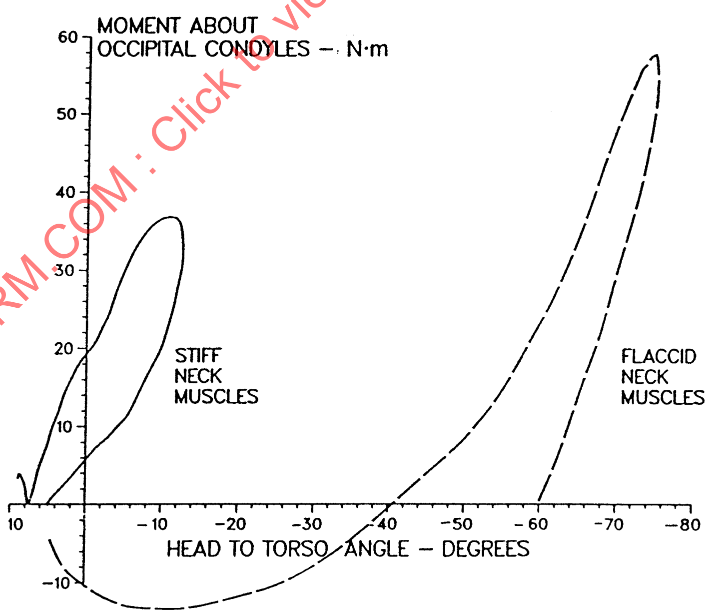

FIGURE 15—COMPARISON OF MOMENT-ANGLE NECK FLEXION RESPONSES OF TWO EMBALMED CADAVERS TESTED AT 4 G SLED DECELERATION. ONE CADAVER WITH A HIGH DEGREE OF MUSCLE RIGOR (“STIFF NECK MUSCLES”) AND THE OTHER WITH MINIMAL RIGOR (“FLACCID NECK MUSCLES”)

5.1.2 EXTENSION TESTS—The neck extension tests of Mertz and Patrick (see 2.1.3.(1) and 2.1.3.(2)) were conducted by rotating the rigid seat that was described previously 180 degrees on the sled so that neck extension would be produced when the seat was decelerated from a prescribed initial velocity. Again, the neck's response was characterized in terms of the internal bending moment acting on the head calculated about the occipital condylar axis as a function of change in angular position of the head relative to the torso. Figure 16 shows the response of the tensed volunteer for the most severe test condition, 3.2 G sled deceleration, and the response of an embalmed cadaver with a flaccid neck for a 5.4 G sled deceleration.

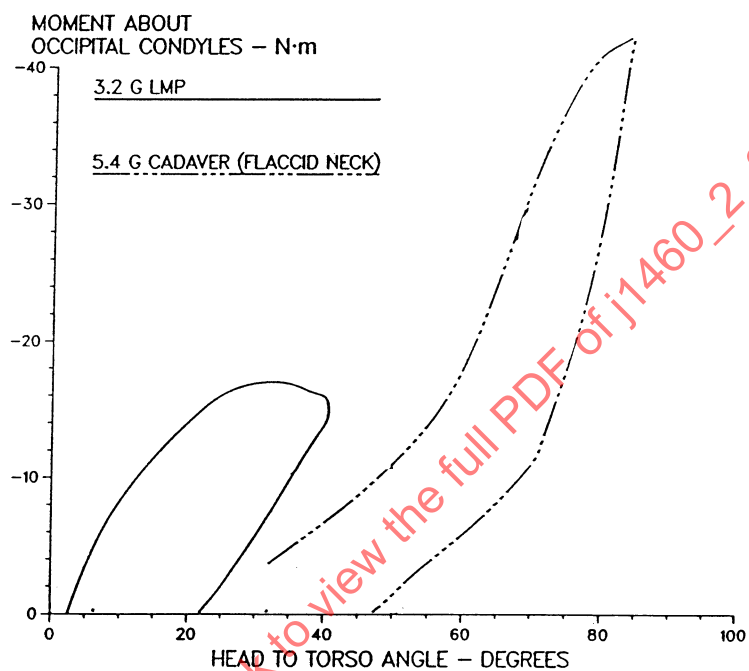

FIGURE 16—MOMENT-ANGLE NECK FLEXION RESPONSES OF A VOLUNTEER WITH TENSED MUSCLES AND A CADAVER WITH FLACCID MUSCLES

5.2 Patrick and Chou Results

5.2.1 FLEXION TESTS—Patrick and Chou (see 2.1.3.(5)) conducted a follow-up program to the Mertz and Patrick study using the same seat and restraint fixture. Fifty-four tests were conducted in the flexion mode involving four volunteers with sled deceleration levels ranging from 0.8 G to 11.7 G. Figure 17 shows the neck moment-angle responses of three volunteers to various sled deceleration levels. These curves are similar in shape for equal decelerations with those given by Mertz and Patrick, (see Figure 12), and extend the response data to higher sled deceleration levels. The neck moment-angle response curve for the most severe sled deceleration level (11.7 G) was not provided by Patrick and Chou.

5.2.2 LATERAL BENDING TESTS—Eighteen tests were conducted that produced neck response data for lateral head inertial loading. For these tests, a side board was added to the seat and the seat was mounted sideways on the sled. Two volunteers were involved in these tests. Figure 18 shows the neck moment-angle responses of the volunteers for sled deceleration levels ranging from 2.8 G to 6.7 G. For lateral loading, the internal neck moment acting on the head was calculated about a fore/aft axis lying in the midsagittal plane at the level of the occipital condyles. Note that the volunteer LMP anticipated the sled deceleration by tilting his head away from the direction of the impending head motion. The negative moment-angle response that occurred early in the event was not reported by Patrick and Chou.

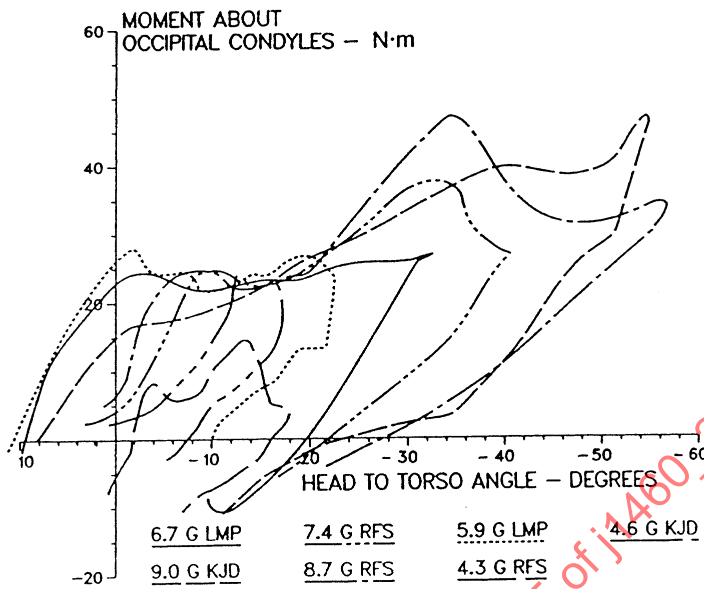

FIGURE 17—COMPARISON OF MOMENT-ANGLE NECK FLEXION RESPONSES OF THREE VOLUNTEERS FOR VARIOUS SLED DECELERATION LEVELS. NECK MUSCLES TENSED (SEE 2.1.3.(5))

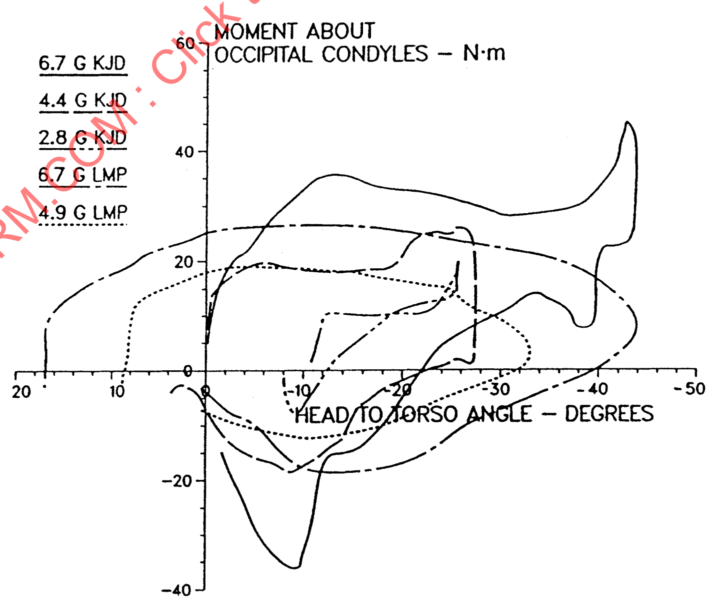

FIGURE 18—COMPARISON OF MOMENT-ANGLE NECK LATERAL BENDING RESPONSES OF TWO VOLUNTEERS FOR VARIOUS SLED DECELERATION LEVELS. NECK MUSCLES TENSED (SEE 2.1.3.(5))

5.2.3 EXTENSION, OBLIQUE EXTENSION, AND OBLIQUE FLEXION TESTS—Patrick and Chou conducted a single extension test at 4.8 G. The moment-angle response of the volunteer LMP is shown on Figure 19. They also conducted two oblique extension tests (45 degrees from pure extension) at 3.3 G and 4.8 G and six oblique flexion tests (45 degrees from pure flexion) ranging from 1.9 G to 5.2 G. They did not publish any neck moment-angle curves for these oblique tests. They did note that the volunteer LMP had commented that the 4.8 G oblique extension test was the most severe exposure for the given level of sled deceleration

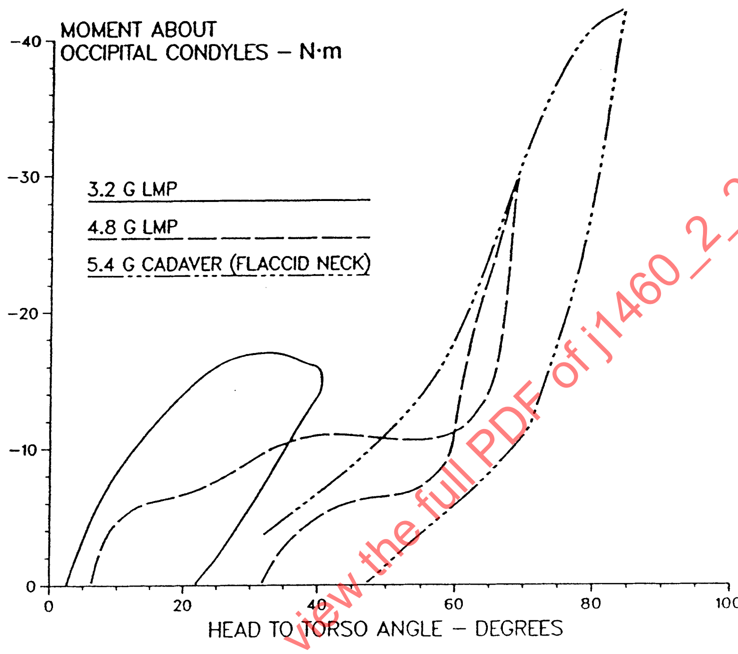

FIGURE 19—COMPARISON OF MOMENT-ANGLE NECK EXTENSION RESPONSES OF A VOLUNTEER WITH TENSED NECK MUSCLES AND A CADAVER WITH FLACCID MUSCLES (SEE 2.1.3.(1), 2.1.3.(2) AND 2.1.3.(5))

5.3 Ewing et al., Results—Ewing et al., (see 2.1.3.(6), 2.1.3.(7), 2.1.3.(8), 2.1.3.(9), 2.1.3.(10), 2.1.3.(11), 2.1.3.(12), and 2.1.3.(13)) have conducted numerous human volunteer sled tests that produced forward or lateral bending of the neck. In their tests, the volunteers, all young, physically-fit military men, were seated in a rigid chair in an erect posture (thoracic spine vertical and thighs horizontal). For the flexion tests, the volunteers were restrained by a standard military aircraft harness, which consisted of a lap belt with inverted-V crotch straps and left and right shoulder belts. For the lateral neck bending tests, the seat was mounted sideways on the sled and a lightly padded vertical plate was added to restrict sideways motion of the torso and legs.

The head of each volunteer was instrumented with a rate gyro in addition to linear accelerometers attached to a tightly-fitting, lightweight skull cap and a bite plate, in order to measure linear and angular head motion. This resulted in a bite plate instrumentation with greater mass compared to that used by Mertz and Patrick. An instrumentation package, consisting of linear accelerometers and a rate gyro was also strapped over the posterior region of the first thoracic vertebra. All tests were conducted on a Hyge sled that was accelerated to a prescribed velocity. Both the rate of sled acceleration and peak sled acceleration were used to specify test severity. Figure 20 shows the resultant bite-plates accelerations of six volunteers subjected to 10 G, 250 G/s sled tests that produced neck flexion. The large variation in peak resultant accelerations is probably due, in part, to the differences in head size and shape of the volunteers which produced different locations of the bite plate accelerometers relative to the centers of gravity of their heads.

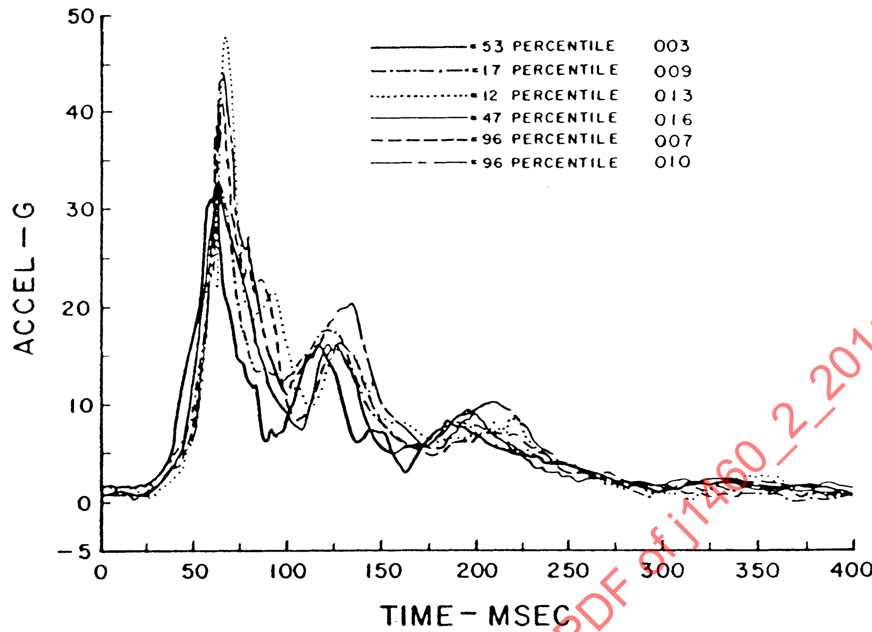

FIGURE 20—COMPARISONS OF MOUTH RESULTANT ACCELERATIONS FOR 10 G, 250 G/S HYGE SLED TESTS THAT PRODUCED NECK FLEXION OF SIX VOLUNTEERS WITH TENSED NECK MUSCLES AND ERECT SEATED POSTURE (SEE 2.1.3.(7))

Ewing et al., (see 2.1.3.(8)) used the technique developed by Mertz (see 2.1.3.(1)) to calculate the internal neck moments about the lateral occipital condylar axis for a number of the flexion tests. The moment-angle responses of two of these subjects are shown in Figures 21 and 22 for various sled acceleration levels, and show two differences compared to those of Mertz and Patrick (see Figures 12 and 14). For the higher severity tests, Ewing et al., volunteers demonstrated 20 to 25 degrees more rotation of the head relative to the torso prior to the point where neck stiffness increased. A second difference is that the response curves for Ewing et al,. volunteers show a more pronounced initial negative moment than in the Mertz and Patrick volunteer curves.

The Ewing et al., test results also show a different kinematic response, such that forward rotation of the head is delayed, producing a “chin-out” type of movement. A review of numerous high-speed films of sled tests of volunteers and cadavers seated in an automotive posture and restrained by a conventional 3-point belt system indicates that this “chin-out” response is not characteristic of volunteers seated in the automotive posture. The heads of these volunteers were not sufficiently instrumented to allow the calculation of internal neck forces. However, kinematic analysis of head-to-torso motion is being done and will be included later as an addendum to this report.

There are three possible reasons for these differences in kinematic and kinetic responses between the Ewing et al., volunteers and those of other studies - seating posture, instrumentation mass, and level of voluntary muscle activity. The greater flexion range of motion in the Ewing et al., tests may be due, in part, to the fact that the subjects were seated upright compared to the 15-degree reclined posture used in the Mertz and Patrick tests. The increase in flexion angle is approximately equal to the difference in thoracic spine angles for the two test series.

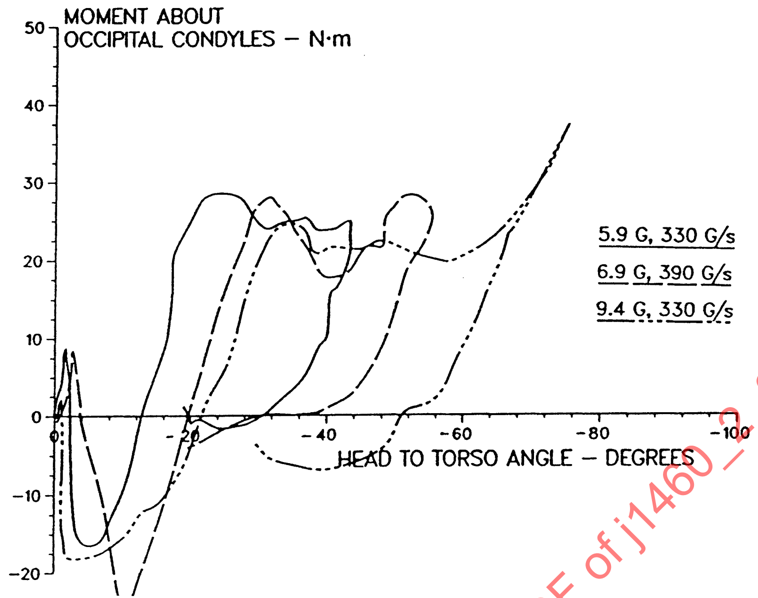

FIGURE 21—MOMENT-ANGLE NECK FLEXION RESPONSES OF VOLUNTEER SUBJECT 4 FOR VARIOUS HYGE SLED ACCELERATION AND ONSET LEVELS. TENSED NECK MUSCLES AND ERECT SEATED POSTURE (SEE 2.1.3.(8))

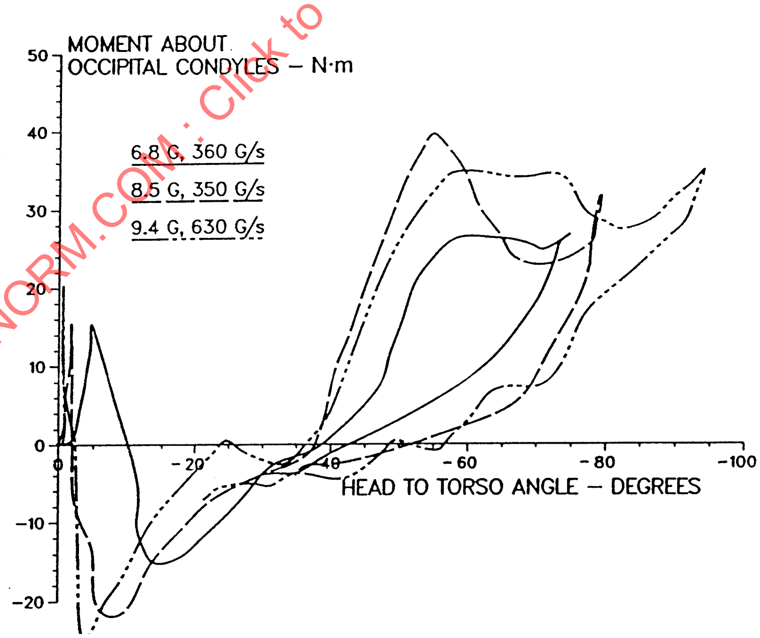

FIGURE 22—MOMENT-ANGLE NECK FLEXION RESPONSES OF VOLUNTEER SUBJECT 7 FOR VARIOUS HYGE SLED ACCELERATION AND ONSET LEVELS. TENSED NECK MUSCLES AND ERECT SEATED POSTURE (SEE 2.1.3.(8))

The heavier instrumentation mass attached to the bite plates in Ewing et al., tests would tend to increase the rotational inertia of the head and move the center-of-gravity of the combined head and instrumentation mass forward and downward. This would increase the initial negative moment acting on the head about the occipital condyles and delay forward head rotation. A similar effect was noted by Mertz and Patrick (see 2.1.3.(3)) when the added mass was attached to the skull cap so that it was located below the occipital condyles of the subjects.

Finally, it appears that Ewing et al., volunteers may have learned, through previous low-impact tests, to produce higher neck muscle force and thereby better control their head movement to prevent the bite plate from impacting the chest. Considerable neck muscular reaction is evident in the films of the tests. Such learned responses may not provide appropriate response data for anthropomorphic test devices that should simulate people subjected to an impact event for the first time.

5.4 Zaborowski Results—Zaborowski (see 2.1.3.(14)) conducted 87 lateral deceleration sled tests on 57 male volunteers, the average age being 23.6 years. A modified aircraft ejection seat with a 12-degree seatback angle was mounted to the sled sideways to the direction of travel. The subjects were restrained by a lap belt and a shoulder harness which consisted of two 76-mm wide straps that were attached at one end to the lap belt, passed over each shoulder, and were sewn together behind the shoulders with this end being attached to the seatback. The seat had a side plate which allowed 30 degrees of torso rotation prior to subject interaction. Triaxial accelerometer packages were strapped over the sternum and the side of the head. Head and torso motions were recorded with high-speed movies. Sled decelerations ranged from 4 G to 12 G in 2 G increments with pulse durations from 220 ms to 90 ms. The testing was terminated after the second 12 G sled test when the subject experienced bradycardia (slowing in heart rate) and fainted immediately post test. Table 1 is a summary of the averages and ranges of the peak chest accelerations, head accelerations, and head-to-torso angles for the various sled deceleration levels. Note the wide dispersion of responses for any given sled deceleration level. Of particular interest is the large range of peak head-to-torso angles which may reflect the subjects' anticipation of the sled deceleration level and variations of neck muscle strength. For example, for the 10 G sled tests, the average peak head-to-torso angle was –33.3 degrees with a range of –11 degrees to –63 degrees. These wide variations make it difficult to select a set of peak responses from this data set to characterize the average lateral neck response for a given sled exposure.

TABLE 1—SUMMARY OF LATERAL SLED TEST RESULTS OF HUMAN VOLUNTEER EXPERIMENTS REPORTED BY ZABOROWSKI (SEE 2.1.3.(14))

Sled Deceleration (G)	Number of Subjects	Peak Chest Acceleration (G) Average	Peak Chest Acceleration (G) Range	Peak Head Accleration (G) Average	Peak Head Accleration (G) Range	Peak Head/ Torso Angle (Degrees) Average	Peak Head/ Torso Angle (Degrees) Range
4	20	5.9	4.7 to 6.8	7.1	5.5 to 10.2	–15.8	3 to –32
6	20	8.9	5.4 to 13.7	10.1	4.4 to 21.2	–29.0	–8 to –57
8	25	12.6	9.4 to 14.8	15.0	9.6 to 24.8	–32.0	–19 to –57
10	20	18.6	13.6 to 23.8	19.9	12.5 to 45.4	–33.3	–11 to –63
12	2	22.1	17.6 to 26.7	21.7	18.6 to 24.7	–38.0	(1)

1. Data available for only 1 subject.

5.5 Hu et al., Results—Hu et al., (see 2.1.3.(15)) conducted six unembalmed cadaver sled tests that were to simulate a 51.4 km/h (19-20 G) rear-end collision of two cars of equal mass. The cadavers were seated on an automotive bench seat and restrained by a 3-point harness. The head restraint was adjusted to its lowest position. For three of the cadavers, the seatback was rigidly supported. For the other three cadavers, the seatback was allowed to deform rearward due to the subjects' inertial loading. The heads of the cadavers were instrumented with accelerometers from which the peak accelerations and HIC values were 25.4 G and 229 for the deforming seatback tests and 46.5 G and 292 for the rigidly supported seatback tests, respectively. All three cadavers in the deformable seatback tests and two of the cadavers in the rigidly supported seatback tests experienced significant neck damage (subluxation, fracture, and/or disk ruptures). The sixth cadaver had no significant damage post test. Because the values of peak resultant head acceleration and HIC were low for these tests, neither appears to be an appropriate indicator for assessing the potential for neck injury. Hu et al., also calculated internal reactions between the head and neck at the level of the occipital condyles. These calculations are in error since they did not include the load applied to the head by the head restraint.

5.6 Tarriere Results—Tarriere (2.1.3.(16)) conducted five cadaver sled tests to obtain data on the neck's lateral bending response. One cadaver was exposed to 6.6 G sled deceleration which is similar in severity to the maximum sled exposures of the human volunteers of Patrick and Chou, and Ewing et al. The other four cadavers were exposed to higher sled deceleration levels (12.2 G, 14.2 G, 14.0 G, and 14.6 G). For these higher level exposures, the maximum lateral rotation of the head with respect to the torso ranged from –36 degrees to –78 degrees, (see Table 2). This large spread in head to torso angle is probably due to differences in the degree of stiffnesses left in the neck muscles of the cadaver specimens.

TABLE 2—RESULTS OF HIGH G-LEVEL CADAVER TESTS OF TARRIERE (SEE 2.1.3.(16))

	Tests MS 249	Tests MS 297	Tests MS 361	Tests MS 359
Peak Sled Deceleration - G	12.2	14.2	14.0	14.6
Initial Sled Velocity - m/s	6.08	6.19	6.25	8.61
Peak Horizontal Acceleration at T1 Level - G	20.0	44.0	31.5	34.4
Peak Horizontal Acceleration of the Head C.G. - G	36.0	17.3	8.2	9.7
Head Lateral Flexion - Degrees	78	36	59	78
Peak Head Torsion - Degrees	42	30	70	102
Maximum Horizontal Displacement of the Head C.G. Relative to the Sled - mm	294	445	260	415
Maximum Vertical Displacement of the Head C.G. Relative to the Sled - mm	79	78	64	110
Maximum Horizontal Velocity of the Head C.G. Relative to the Sled - m/s	4.3	5.3	4.8	5.7

6. ***Voluntary-Range-of Motion Studies—***The determination of the range-of-motion is an important parameter for surrogate neck designs. At the extremes of voluntary range-of-motion, the neck ligaments (and/or muscles) are loaded passively, resulting in a marked increase in the neck's bending resistance. This non-linear characteristic in neck bending response needs to be incorporated in any surrogate neck design.

A number of investigators have conducted studies to determine the range-of-motion of the head with respect to the torso (see 2.1.3.(1), 2.1.3.(2), 2.1.3.(3), 2.1.3.(4), 2.1.3.(5), 2.1.3.(17), 2.1.3.(18), 2.1.3.(19), 2.1.3.(20), and 2.1.3.(21)). The results given in Tables 3, 4, 5, 6, 7, and 8A indicate that (1) there is a large variation in the voluntary range of motion of the head relative to the torso within the adult population, (2) the range-of-motion dramatically decreases with age, and (3) neither sex nor size has a large influence on range-of-motion. Ranges-of-motion for fore/aft bending given by Foust et al., and Schneider et al., (Tables 6 and 7) are 20 degrees less than the ranges obtained by other investigators (see Tables 3 and 5). Mertz et al., (2.1.3.(22)) provided a curve of voluntary trajectory of the center-of-gravity of the head of subject LMP for fore/aft bending of the neck (see Figure 23). Noted on the curve are the angular positions of the head relative to the torso. The total fore/aft excursion was 120 degrees which is 13 degrees less than the excursion measured by Patrick and Chou for the same volunteer (see Table 8A). This points out the large variation that can occur and the need to restrict torso motion when conducting these types of tests.

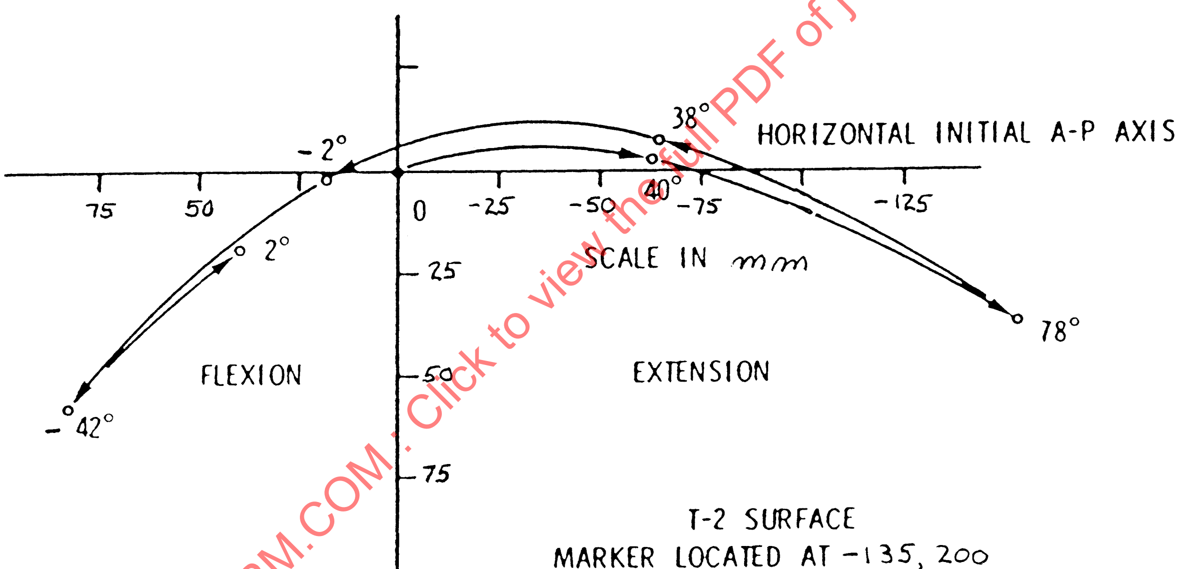

FIGURE 23—VOLUNTARY TRAJECTORY OF THE CENTER-OF-GRAVITY OF THE HEAD OF VOLUNTEER LMP (SEE 2.1.3.(22))

TABLE 3—FORE/AFT VOLUNTARY RANGES-OF-MO TION OF THE HEAD RELATIVE TO THE TORSO AS MEASURED BY FERLIC (SEE 2.1.3.(17))

Age (Years)	Average ± Standard Deviation (Degrees)
15 – 24	139 ± 19
25 – 34	127 ± 22
35 – 44	120 ± 19
45 – 54	120 ± 15
55 – 64	116 ± 22
Average	127 ± 20

TABLE 4—VOLUNTARY RANGES-OF-MOTION OF THE HEAD REALTIVE TO THE TORSO FOR FLEXION AND EXTENSION, AGES 18 TO 23, BUCK ET AL. (SEE 2.1.3.(18))

Subjects Sex	Number of Subjects	Flexion (Degrees)	Extension (Degrees)	Combined (Degrees)
Male	47	–66 ± 8	73 ± 9	139
Female	53	–69	81	150

TABLE 5—VOLUNTARY RANGES-OF-MOTION OF THE HEAD RELATIVE TO THE TORSO FOR FLEXION AND EXTENSION, AGES 20 TO 40, GRANVILLE AND KREEZER (SEE 2.1.3.(19))

Subjects Sex	Number of Subjects	Flexion (Degrees)	Extension (Degrees)	Combined (Degrees)
Male	10	–60 ± 12	61 ± 27	121

TABLE 6—VOUNTARY RANGES OF FORE/AFT HEAD-TO-TORSO MOTION, FOUST ET AL. (SEE 2.1.3.(20))

Age	Sex	Range-of-Motion by Percentile 1–20 Mean (Degrees)	Range-of-Motion by Percentile 1–20 S.D. (Degrees)	Range-of-Motion by Percentile 40–60 Mean (Degrees)	Range-of-Motion by Percentile 40–60 S.D. (Degrees)	Range-of-Motion by Percentile 80–99 Mean (Degrees)	Range-of-Motion by Percentile 80–99 S.D. (Degrees)	Average (Degrees)
18 – 24	Male	133	16	139	9	140	13	137
35 – 44	Male	105	13	109	18	118	25	109
62 – 72	Male	101	10	83	22	103	19	94
18 – 24	Female	132	15	136	18	147	15	138
35 – 44	Female	117	16	123	20	130	15	122
62 – 72	Female	106	15	104	17	98	22	99

TABLE 7—VOLUNTARY RANGES OF HEAD-TO-TORSO MOTION, SCHNEIDER ET AL. (SEE 2.1.3.(21))

Age	Sex	Total Range of Motion Fore/Aft (Degrees)	Total Range of Motion Lateral (Degrees)	Total Range of Motion Rotational (Degrees)
18 – 24	Male	129	86	150
35 – 44	Male	103	73	137
62 – 74	Male	77	48	114
All Males		103	70	134
18 – 24	Female	124	86	150
35 – 44	Female	105	74	144
62 – 74	Female	84	56	124
All Females		104	72	139
All Subjects		104	71	137

TABLE 8A—VOLUNTARY RANGES OF HEAD-TO-TORSO MOTION OF HUMAN VOLUNTEERS OF PATRICK AND CHOU (SEE 2.1.3.(5))— FORE/AFT BENDING MODE (DEGREES)

Volunteer	Flexion	Extension	Total Range
LMP	–51	82	133
KJD	–65	73	138
SAT	–63	69	132

TABLE 8B—VOLUNTARY RANGES OF HEAD-TO-TORSO MOTION OF HUMAN VOLUNTEERS OF PATRICK AND CHOU (SEE 2.1.3.(5))— LATERAL FLEXION MODE (DEGREES)

Volunteer	Flexion	Extension	Total Range
LMP	–42	43	85
SAT	–35	39	74

TABLE 8C—VOLUNTARY RANGES OF HEAD-TO-TORSO MOTION OF HUMAN VOLUNTEERS OF PATRICK AND CHOU (SEE 2.1.3.(5))— 45 DEGREES FORWARD FLEXION MODE (DEGREES)

Volunteer	Flexion	Extension	Total Range
LMP	35	56	91

TABLE 8D—VOLUNTARY RANGES OF HEAD-TO-TORSO MOTION OF HUMAN VOLUNTEERS OF PATRICK AND CHOU (SEE 2.1.3.(5))— 135 DEGREES REARWARD EXTENSION MODE (DEGREES)

Volunteer	Flexion	Extension	Total Range
LMP	35	56	91

7. ***Guidelines for Assessing Neck Bending Response Biofidelity***—The data presented in the preceding sections have been used by various investigators to develop guidelines for assessing the biofidelity of the bending responses of surrogate neck structures. The human response data of Mertz et al., (see 2.1.3.(1), 2.1.3.(2), 2.1.3.(3), and 2.1.3.(4)) and Patrick and Chou (see 2.1.3.(5)) are directly applicable to the automotive collision environment since all of their testing was done for the automotive seated posture. The human response data of Ewing et al., (2.1.3.(6), 2.1.3.(7), 2.1.3.(8), 2.1.3.(9), 2.1.3.(10), 2.1.3.(11), 2.1.3.(12), and 2.1.3.(13)) are not directly applicable to the automotive environment. All their volunteer testing was done with an upright, spine-vertical posture. That produced a "chin-out" kinematic response that is not characteristic of the kinematic response for the automotive posture. A dummy neck structure designed to be used in both postures would have to be adjustable to obtain the different initial positions and ranges of motion. Since assessing the biofidelity of dummy neck structures for the automotive seated posture is the primary focus of this document, only neck bending response guidelines for such posture are given. Ewing et al., volunteer data with appropriate changes to the data to account for the postural differences, are compared to these guidlines. The results of human volunteer tests of Zaborowski (see 2.1.3.(14)) are not used to define neck response guidelines because only highly scattered, peak responses were given. The cadaver data of Hu et al. (see 2.1.3.(15)) and Tarriere (see 2.1.3.(16)) are not used to define neck response guidelines since no moment-angle responses are given.

7.1 **Flexion Response Requirements**—Mertz and Patrick (see 2.1.3.(4)) and Mertz, et al., (see 2.1.3.(22)) formulated requirements for assessing the biofidelity of the flexion responses of surrogate neck structures representative of a mid-size adult male in the automotive seated posture. Based on an analysis of their dynamic human volunteer and cadaver tests and their static neck strength tests, they developed a response corridor for the relationship between the internal neck bending moment acting on the head taken with respect to a lateral axis passing through the occipital condylar joints and the change in angular position of the head relative to the torso (see Figure 24). The plateau portion represents the maximum moment that the neck muscles of a mid-size male can generate in resisting a static forward head motion before appreciable head rotation occurs. This value was taken from the static pull test results (see 2.1.3.(4)). The change in slope at the point (45 degrees, 61 N·m) represents the increased stiffness that occurs when the normal articular voluntary range-of-motion of the neck is reached. At this point, the neck ligaments and/or passive stretch of the neck muscles increase the bending resistance of the neck. For larger angles, these effects are more pronounced and are represented by the greater slope of the corridor for angles exceeding 60 degrees. The lower portion of the corridor was drawn to contain the unloading data of the volunteer and reflects the elastic response of the ligaments and the energy dissipation of the muscles. Since the volunteer whose response data were used to develop the corridor was representative of the 50th percentile adult male (1.73 m = 5'8", 72.7 kg = 160 pounds, 49 years old, sedentary occupation), Mertz and Patrick concluded that the resistive portion ($+M_y$) of the moment-angle response of a surrogate neck structure would have to lie within the corridor to be considered representative of the response of the 50th percentile adult male neck.

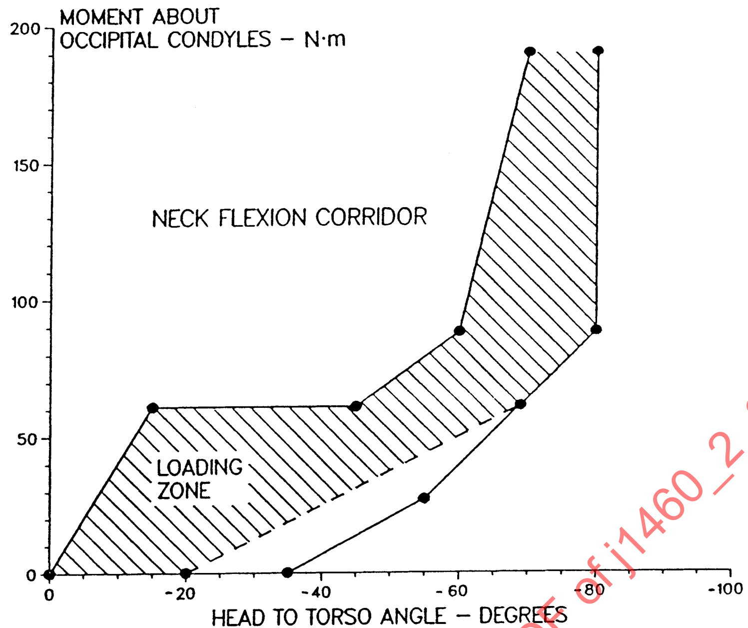

FIGURE 24—MOMENT-ANGLE NECK FLEXION CORRIDOR AND LOADING ZONE FOR THE MID-SIZE ADULT MALE SEATED IN AUTOMOTIVE POSTURE (SEE 2.1.3.(4) AND 2.1.3.(22)).

The moment-angle responses of relaxed and tensed volunteers are compared to the corridor in Figure 25. Figure 26 compares the moment-angle response of two Ewing et al., volunteers (see 2.1.3.(8)) to the Mertz and Patrick corridor. The Ewing et al., curves were shifted to account for the differences in seated posture (15 degrees) and the assumed differences in range-of-motion of the younger subjects tested by Ewing (5 degrees). Note that the initial portion of the moment-angle curves are negative and are not part of the guideline. Further, it should be noted that the corridor can be used to judge the biofidelity of neck structures subjected to all levels of forward bending exposures and for all levels of muscle tone for the automotive seated posture. In fact, a simple pendulum test of the head/neck system can be used to evaluate the moment-angle response relative to this corridor.

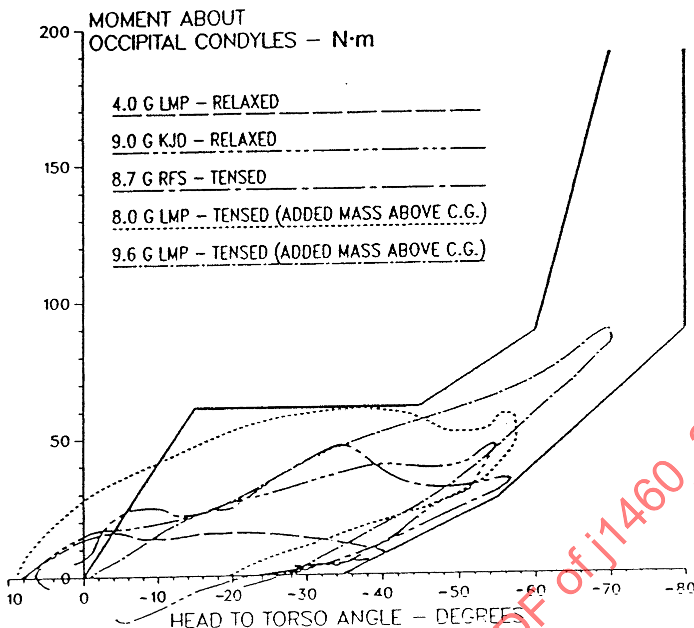

FIGURE 25—MOMENT-ANGLE NECK FLEXION RESPONSES OF VOLUNTEERS WITH TENSED AND RELAXED NECK MUSCLES COMPARED TO THE NECK FLEXION CORRIDOR FOR THE MID-SIZE ADULT MALE SEATED IN AUTOMOTIVE POSTURE

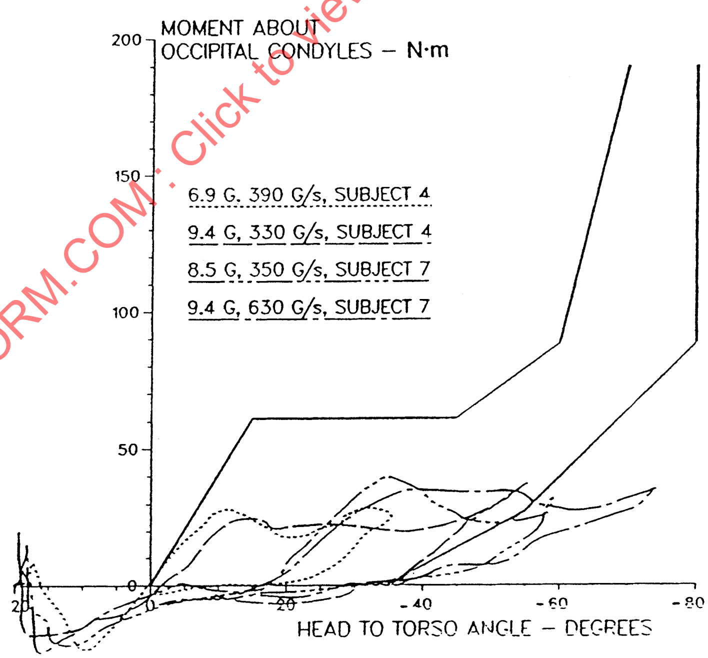

FIGURE 26—MOMENT-ANGLE NECK FLEXION RESPONSES OF EWING ET AL. VOLUNTEERS (REF. 8) COMPARED TO THE NECK FLEXION CORRIDOR FOR THE MID-SIZE ADULT MALE SEATED IN AUTOMOTIVE POSTURE. EWING DATA SHIFTED BY 20 DEGREES TO ACCOUNT FOR POSTURAL DIFFERENCES (15 DEGREES) AND DIFFERENCES IN RANGE-OF-MOTION OF YOUNGER SUBJECTS (5 DEGREES)

Mertz et al., (see 2.1.3.(22)) also defined a trajectory requirement (see Figure 27) in terms of the change in position of the center-of-gravity of the head that was observed in the 9.6 G sled test of Figure 14, their most severe human volunteer flexion test with the added mass above the center-of-gravity of the head. In addition to the moment-angle and the trajectory requirements, the head and neck of the surrogate should meet the geometric constraints of the mid-size adult male and the neck structure should have sufficient degrees of freedom to approximate the flexion curvature of the neck.

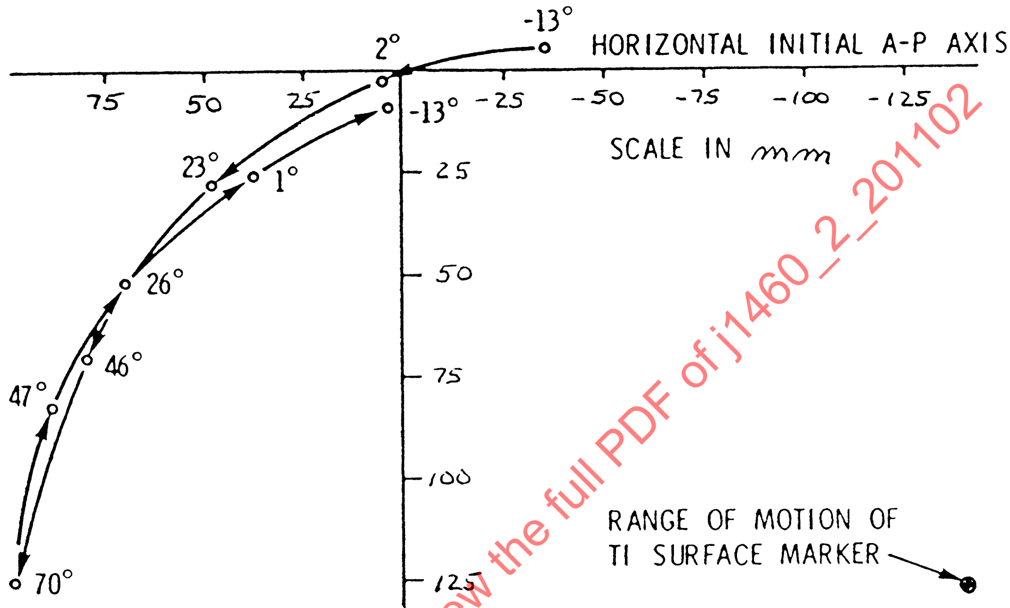

FIGURE 27—HEAD C.G. TRAJECTORY FOR 9.6 G NECK FLEXION SLED TEST
OF MERTZ AND PATRICK (SEE 2.1.3.(4))
ADDED MASS ABOVE C.G. OF HEAD

7.2 Extension Response Requirements—Mertz and Patrick (see 2.1.3.(4)) defined requirements for assessing the biofidelity of rearward bending responses of surrogate neck structures designed to represent the mid-size adult male. Using their dynamic human volunteer and cadaver data and their strength pull test results, they constructed an extension corridor defined by the solid lines of Figure 28. This corridor constrains the relationship between the internal neck bending moment acting on the head taken with respect to a lateral axis passing through the occipital condyles and the change in angular position of the head relative to the torso for the automotive seated posture. Later, Mertz et al., (see 2.1.3.(22)) defined a “loading zone” for the extension corridor, also shown in Figure 28. The plateau was chosen based on static neck strength data for resisting extension (see 2.1.3.(2)). The angle range for increasing the stiffness was based on the voluntary articulation range of the volunteer when seated in an automotive posture. The maximum stiffness slopes represent the increase in stiffness that occurs at the end of the voluntary articulation range when the ligaments begin to add to the bending resistance. The lower portion of the corridor reflects the elastic behavior of the ligaments and muscles, and energy dissipation of the muscles that occurs during rebound. Since the volunteer whose responses were used to construct the corridor was representative of a mid-size adult male, Mertz and Patrick concluded that the resistive portion ($-M_y$) of the moment-angle response of a surrogate neck structure would have to lie within the corridor to be considered representative of the response of the 50th percentile adult male.