BIOMECHANICAL RESPONSE OF HUMAN CERVICAL SPINE TO DIRECT LOADING OF THE HEAD

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ABSTRACT

Currently, dummies are used to evaluate the potential for neck injuries under both inertial loading and direct contact loading, such as in the case of air bag loading. However, there is currently little known data on the biomechanical response of the human neck under direct head contact loading conditions. The objective of this preliminary research is to develop an experimental approach and analysis method that can characterize the response of human neck under low severity head impact loading conditions and compare the results with those for a dummy.

The tests include three healthy adult male volunteers with average age of 22, as well as a Hybrid III (HY-III) test dummy. Low-level impact loads to the head and face of each subject were applied via a strap at one of three locations in various directions: an upward load applied to the chin, a rearward load applied to the chin, and a rearward load applied to the forehead. The resultant forces and moments at the occipital condyle were calculated using the measured applied loads, inertial properties of the head, and translational and angular accelerations of the head. For the volunteers, activation of the neck musculature was determined using EMG. Although two initial muscle activation conditions, relaxed and tensed, were investigated, only the relaxed results are presented in this study. For each of the test conditions, the three relaxed volunteers showed similar head rotation patterns, with the highest head rotational velocities observed for the chin-upwards loading condition. The head rotation for the relaxed volunteers was also substantial in the chin-upward and forehead rearward cases, up to 25 degrees. By contrast the dummy's head rotation was small for all loading modes, less than 5 degrees. It is expected that as the intensity of the applied load increases and/or the volunteers tense the muscles in the cervical spine, these differences will decrease. The calculated forces at the occipital condyle were comparable for the volunteers and dummy, although the calculated moments in the dummy were somewhat higher than those for the relaxed volunteers.

Although preliminary testing at low impact severity demonstrates that there may be differences in the moment-angle response between relaxed human volunteers and the HY-III dummy, it remains to be demonstrated if these differences will occur for higher-severity impacts associated with air bags or if these differences will occur for tensed human volunteers. However, once the overall moment-angle response of the head and neck to impact loading is characterized for both low and high levels, the next important step is to understand how the forces and moments are partitioned between the ligamentous cervical spine and the surrounding musculature. Thus, if differences exist, this data should contribute to the improvement of the physical dummy head/neck and its instrumentation to achieve better biofidelic interactions with the environment and to have a greater ability to accurately evaluate the potential for injury to the cervical spine and spinal cord.

Key word: Dummies, Neck, Biofidelity, Human Body

THE HYBRID III NECK HAS been used for many years to evaluate the potential for injury in vehicle crash testing. However, more recently, there has been debate [1)-4)] about the performance of the HY-III neck in testing where the head and neck of the dummy are directly loaded by the deploying airbag. It has been observed that sometimes in this scenario, large shear forces and bending moments are measured at the upper neck load cell without significant neck bending deformation. This phenomenon has not been observed in existing biomechanical corridors, which were developed based upon inertial rather than direct contact head loading. Some have suggested this phenomenon is caused by the differences in performance among different types of vehicle occupant restraint systems, such that well-designed systems do not interact with the dummy in a manner that generates this loading pattern, while others [3)] have hypothesized that the design of the HY-III neck, either because of the stiffness of the OC joint or the design of the underside of the chin, allows non-biofidelic interactions with the deploying airbag, resulting in these high shear forces and moments in the absence of significant neck bending deformations.

The human neck is an extremely complex structure, with various ligaments and muscle groups controlling the articulation between the head and the top of the cervical spine as well as controlling the articulations between the cervical spine vertebrae. The head/neck articulation is provided by the occipital condyle joint, but it is controlled by the ligaments and muscles connecting the head to both the top of the cervical spine and structures below. Thus, the overall moment experienced by the head is the sum of the moments produced by the OC and the moment produced by the surrounding musculature. In isolation, the human ligamentous occipital condyle joint exhibits a very low stiffness over the first 10 to 20 degrees of flexion or extension (Figure 1a). Depending on the level of muscular activation, the human muscular neck is capable of generating a range of bending stiffnesses. For instance, the human muscular neck would have a fairly low bending stiffness when the cervical musculature is partially activated to hold the head upright. However in a real world crash, the human muscular neck may have a much higher bending stiffness when occupants brace themselves for an impending collision. For air bag loading to out-of-position occupants, there may not be sufficient time to activate cervical muscles or the occupants may not be aware of the impending crash. Although the stiffness of the muscular spine is important in determining the kinematics and kinetics of the head and neck due to both direct and inertial loads, severe injuries of the ligamentous spine and spinal cord are most likely a function of only the loads transmitted through the ligamentous spine. Thus, it would be ideal for a dummy neck to accurately replicate and measure the loads in the ligamentous spine directly.

The HY-III dummy neck simplifies the structure of the human neck by representing the neck as an elastic column and the articulation between the head and top of the cervical spine as a revolute joint. The HY-III uses one mechanical element, the nodding block, to transmit moments between the head and neck. Consequently, the stiffness of the nodding blocks represents the combined stiffness of the human ligamentous occipital condyle joint and the cervical muscles attached to the head. The moment-angle behavior of the HY-III neck under quasi-static loading is shown in Figure 1b, where the angle is measured between the head and the top of the cervical spine. This level of stiffness is needed in part to keep the dummy's head in a stable, upright position. Current neck injury criteria attempt to account for the single load path in the HY-III neck design by establishing neck tolerance values that are the sum of allowable forces in the ligamentous and muscular spine. For instance, neck tension tolerance values for in-position occupants are typically higher than that for out-of-position occupants, because in-position occupants are assumed to have activated cervical musculature which will carry more load than relaxed musculature[5)].

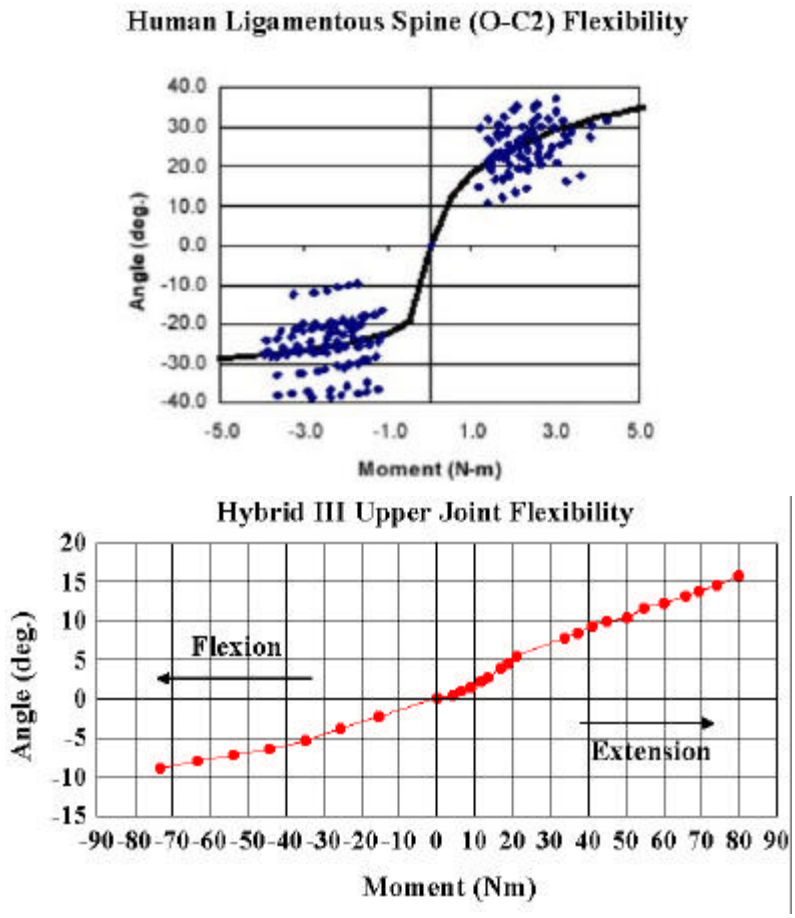


Fig 1a,b: Static Flexibility of the Upper Neck Joint of the Hybrid III and Human Occipital Condyle Joint. The angle is measured between between the occiput and C2 for the 15 female ligamentous OC-C2 spinal motion segments (Used with permission of Nightingale)[6] and between the head and top of the neck for the Hybrid III.

To better understand the partitioning of loads between the ligamentous and muscular neck, three simplified loading conditions were selected for investigation based on analysis of the interaction of the deploying air bag with the head and neck of crash test dummies in rigid barrier and offset frontal crash tests. These loading conditions were: upward load applied to the chin, rearward load applied to the chin, and rearward load applied to the forehead. The objectives of this study are:

1) To develop an experimental approach and analysis method to characterize the response of the human head and neck to direct loading, and
2) To compare the head and neck responses of relaxed human volunteers and HY-III dummy for low-intensity direct head loading.

Then, this work together with additional follow-on efforts, will attempt to determine whether the current approach of adjusting injury criteria limits for particular crash conditions will remain the best overall injury evaluation strategy or if it will be necessary to modify either the dummy neck's physical design and/or the type and location of its instrumentation.

EXPERIMENTAL METHODS

SUBJECTS (VOLUNTEERS) AND INFORMED CONSENT: Test subjects for this study included the three volunteers and the Hybrid III crash test dummy. Three healthy adult male volunteers with average age of 22, without any history of neck injury were selected as the subjects. X-ray photographs confirmed that the volunteers did not show signs of cervical spine degeneration. The experimental protocol was reviewed and approved by the Ethics Committee of Tsukuba University. All volunteers submitted their informed consent in writing according to the Helsinki Declaration[7)] prior to the implementation of the tests. The height, weight and sitting height of each subject are shown in Table 1.

Table 1 Subjects

		Age	Sex	Height (cm)	Sitting Height (cm)	Weight (kg)	Estimated Head Weight (kg)	Estimated Inertia of Head (10*5 g-cm*2)
1	KK	25	M	170	85	78	4.68	2.43
2	KY	22	M	169	86	74	4.52	2.35
3	HT	21	M	173	85	80	4.76	2.48
4	HY-III	-	-	-	91	78	4.48*	2.11*

*: actual value

For the volunteer tests, two initial conditions were investigated, relaxed and tensed muscles. Electromyographs (EMG) were used to determine qualitatively the level of activation of the neck muscles. Skin electrodes were placed over the left sternocleidomastoid muscles (SCM), left paravertebral muscles (PVM), left trapezius muscles (TZM), and right side SCM. Dipole electrodes, which were approximately 5 mm in diameter, spaced at 2 cm intervals for the EMG measurements. For the relaxed muscle condition, the subjects were asked to relax and the EMGs were used to confirm low levels of muscle activation. For the tensed muscle condition, the subjects were asked to statically contract their neck muscles. For both initial conditions, the subjects were not aware of the exact timing of when the tests were initiated.

TEST SETUP

Low-level impact loads to the head and face of each subject were applied via a strap at three locations in various directions (Figure 2), namely:

1) Rearward load applied to forehead (forehead-rearward);
2) Upward load applied to chin (chin-up); and
3) Rearward load applied to chin (chin-rearward).

Loads were applied to the subject by freely dropping a one-kg mass from a height of 20 cm. The mass was connected to a thin steel cable, which in turn was connected to a seat belt webbing strap that was placed around the forehead or chin of the subject. A load cell attached in line with the cable was used to measure the loads applied to the head of the subjects. The peak applied loads were about 150 N and had a duration of about 50 ms.

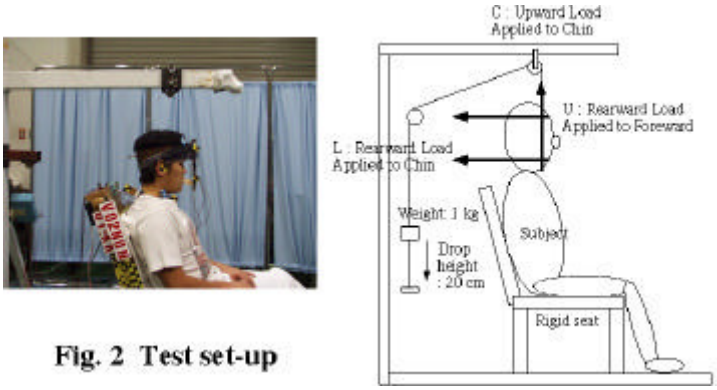


Fig. 2 Test set-up

INSTRUMENTATION: The instrumentation used in the testing is listed in Table 2. Assuming planar motion of the head, two bi-axial linear accelerometers were used to calculate both the rotational and linear acceleration of the center of gravity of the head [8]. The jig shown in Figure 3 (right) was fabricated to fix the accelerometers onto the head. A tooth-impression fixing device molded with dental resin was fixed at the bottom of this jig, and a nylon fastener was attached on the top of jig to fix it on the head of each subject at two areas. The accelerometers were attached as shown in Figure 3 (left), using the head anatomical center of gravity as the reference point located 5 mm in front of ear hole on the Frankfurt line connecting the eye and the ear hole and 2 cm above the vertical line. X-ray photographs were taken with the jig attached to determine the various distances for the calculation of head accelerations and neck moment. Accelerometers in three axial directions of x, y and z (T1 accelerometers) were taped to the skin at the first thoracic vertebra in order to monitor the volunteer's torso motions in impact.

Table 2 Measurement Items

Locations		Items	Axis	Model
Event		Trigger	-	Conatct switch
Volunteer	Head - Upper	Acceleration	x, z	Endevco 7264A (20G Cal.)
	- Lower	Acceleration	x, z	Endevco 7264A (20G Cal.)
	T1	Acceleration	x, z	Endevco 7264A (20G Cal.)
	EMG	Control	Skin	
		SCM(L)	Sternoclaidmastoid muscle (Left)	
		PVM(L)	Paravertebral muscles (Left)	
		TZ(L)	Trapezius muscles (Left)	
		SCM(R)	Sternoclaidmastoid muscle (Right)	
HY-III	Head - Upper	Acceleration	x, z	Endevco 7264A (20G Cal.)
	- Lower	Acceleration	x, z	Endevco 7264A (20G Cal.)
	Head - C.G.	Acceleration	x, y, z	Endevco 7264A (20G Cal.)
	T1	Acceleration	x, y, z	Endevco 7264A (20G Cal.)
	Upper neck	Bending moment	My	Denton 2564
		Axial force	Fz	Denton 2564
		Shear force	Fx	Denton 2564

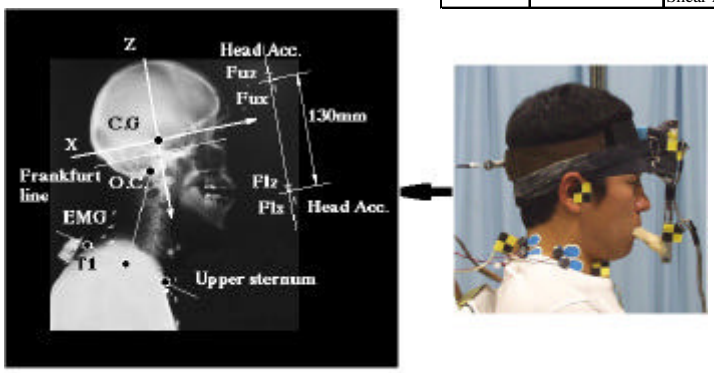


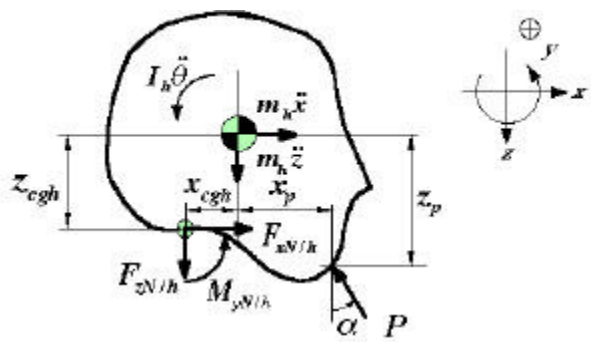
Fig.3 Coordinate system and lateral view of the head/neck/upper torso with mounted accelerometers, EMG electrodes, targets for high speed video

HEAD/NECK KINEMATICS: To analyze the subject's head/neck kinematics during the event, target markers were placed at the following locations: over the head vertical accelerometers, on the side surfaces of the T1 accelerometers, on the front chest (upper sternum), 5 mm below the external auditory meatus, and close to the head center of gravity. The head rotation angle was calculated from the rotation of the line that passes through the center of gravity and is parallel to the line that intersects the Frankfurt and sagittal planes. The torso rotation angle was calculated from the rotation of a line that passes through target markers placed at T1 and the upper sternum. The subject's head/neck/upper torso motions were photographed at the speed of 500 frames per second using a high-speed video camera (MEMORECAMNAC Inc.). Then, the images were analyzed using ImageExpress (NAC Inc.).

CALCULATION OF NECK LOADS: The local shear and axial forces and the bending moment acting on the neck at the occipital condyles were calculated using the analyzed acceleration at the head C.G., estimated head weight, moment of inertia, and external forces applied to the head (Figure 4).

RESULTS

HEAD/NECK KINEMATICS: A comparison of the overall kinematics of one relaxed volunteer and the HY-III are shown in Figure 5 for a vertical load applied to the chin (chin-up case). Approximately 20 ms after impact, both the volunteer's head and dummy head began to rotate. The volunteer's head continued rotating backwards until about 100 ms and exhibited a maximum backward extension angle of about 12 degrees, with respect to an inertial reference frame. The dummy's head, on the other hand, showed the maximum backward rotation angle of about 4 degrees with respect to an inertial reference frame around 60 ms, then it rotated forward back to its initial position.



$$F_{xN/h} = m_h\ddot{x} + P\sin\alpha$$

$$F_{zN/h} = m_h\ddot{z} + P\cos\alpha$$

$$M_{yN/h} = I_h\ddot{\theta} - F_{xN}z_{cgh} + F_{zN}x_{cgh} - Pz_p\sin\alpha + Px_p\cos\alpha$$

Fig.4 Free body diagram of head and the equations of neck forces

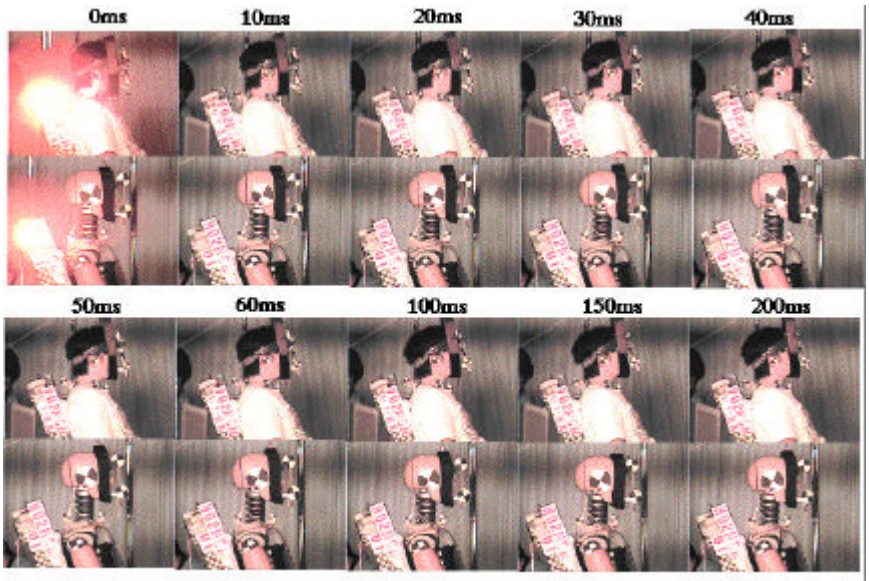


Fig. 5 Comparison of head/neck motion between HY-III and Volunteer (KK) by high-speed video (500 f/s)

The comparison of time histories of the head rotation angles among the three relaxed volunteers and the dummy are shown in Figure 6 for all three loading conditions. The T1 rotations were small (<2 degrees) for both the volunteers and dummy and are not discussed further. In general, the three volunteers had very similar patterns of head rotation with one another for each loading condition. However, the HY-III showed a somewhat different pattern of head rotation from that of the relaxed volunteers. One important difference between the HY-III and the human volunteers is that the HY-III will return back to its initial position when the applied load is removed, which occurs at about 50 ms. Since the human volunteers

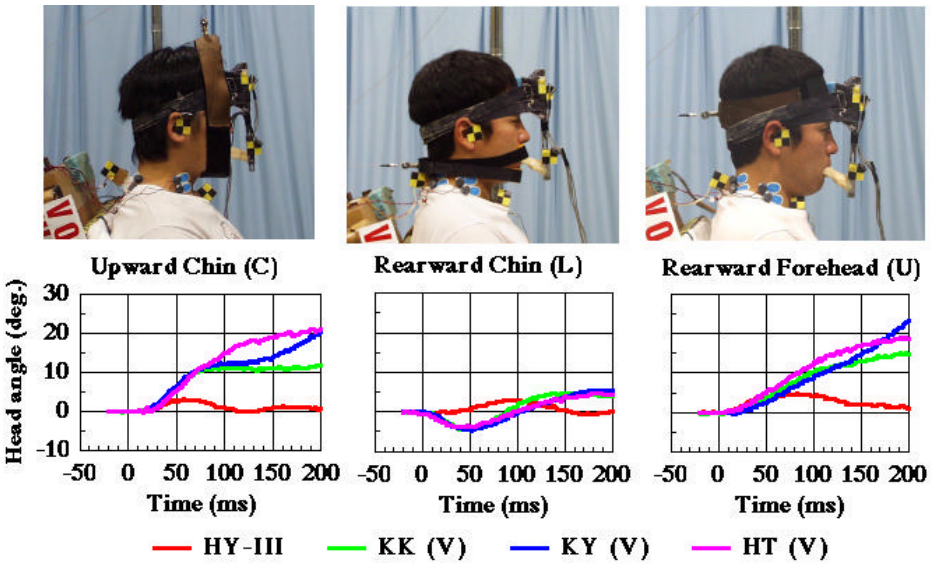


Fig. 6 Patterns of head angle due to three different impact loading conditions

were not given any instructions to return to the initial position, rotation of the head will continue due to inertia after the applied load is removed. For the chin-up case, the volunteers exhibited peak extension head rotations of about 8 degrees during the first 50 ms, whereas the HY-III had a peak head rotation of only 4 degrees. For the chin-rearward case, the volunteers first exhibited a peak flexion rotation of about 5 degrees at 50 ms, followed by an extension head rotation. By contrast, the HY-III exhibited only extension head rotation for the chin-rearward loading condition. For the forehead-rearward case, the HY-III and volunteers exhibited comparable peak head rotations during the first 50 ms. However, the volunteers showed somewhat larger head rotations later in the event.

COMPARISON OF CALCULATED AND MEASURED NECK LOADS: Since the forces and moments at the occipital condyles could not be measured directly for the human volunteers, a validation of the calculation method was performed by comparing the calculated forces and moments derived from the applied loads and bi-axial accelerometers used for both the volunteers and dummy with that measured directly by the HY-III head CG accelerometers and neck load cell.

First, using equations of rigid body dynamics, the accelerations from the two bi-axial accelerometers were used to calculate the acceleration at the head CG of the dummy. The calculated and measured head CG accelerations were comparable. Next, the applied loads and the calculated head CG accelerations were used to compute the forces and moments acting at the occipital condyles, as previously described. For the chin-up case, the calculated forces and moments were within 5% of those obtained with the direct dummy measurements (Figure 7). However, for the chin-backward and the forehead-backward cases, there was a discrepancy between the measured and calculated neck loads and moments (Figures 8-10). The source of this error will be further investigated in future studies. For the calculation of occipital condyle forces and moments in these preliminary tests, it was assumed that the orientation of the applied loads remained constant in the global coordinate system throughout the experiment. It is hypothesized that a more accurate measurement of the direction of the applied loads and the moment arm of these loads throughout the experiment will improve the accuracy of the calculations.

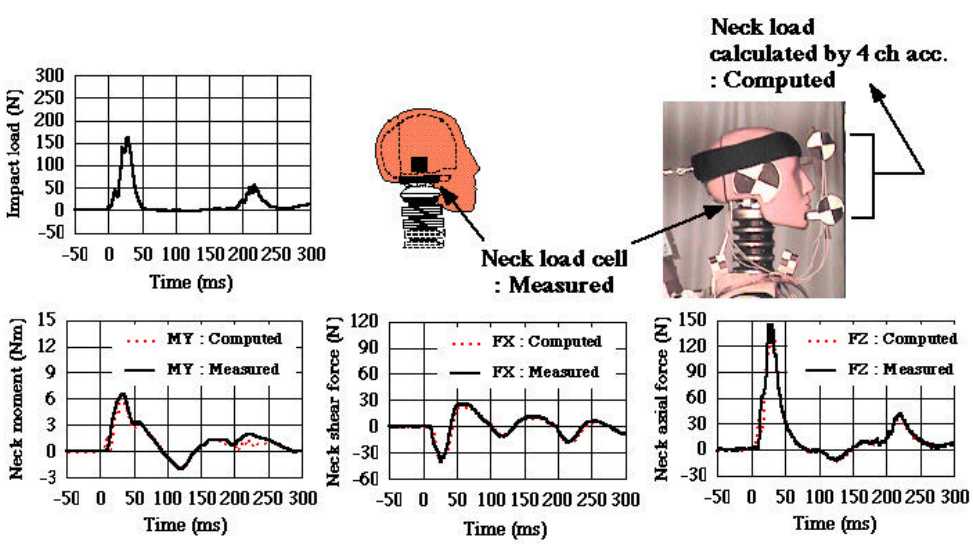


Fig. 7 Validation of calculated methods for neck moment and forces using 4 channel acceleration measurement of the head

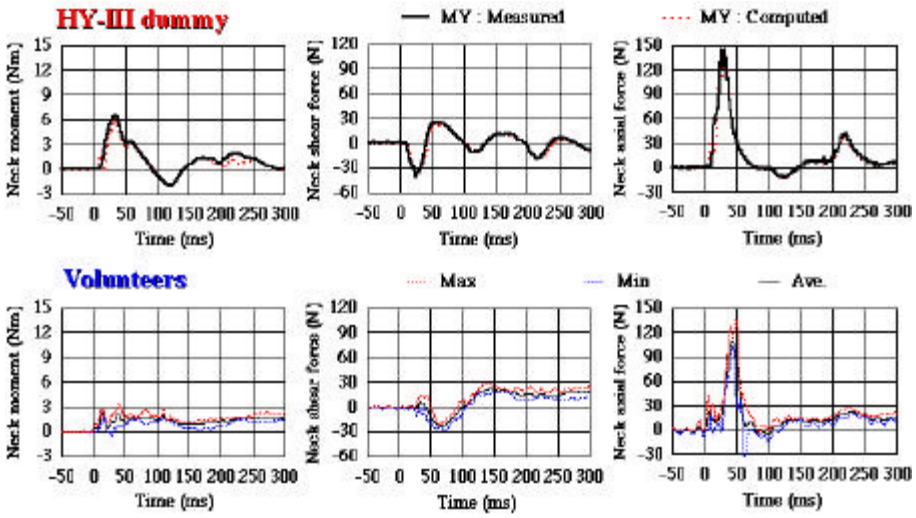


Fig. 8 Comparison of neck moment and forces between HY-III and Volunteers under the condition of upward load applied to chin (Chin-up: C)

COMPARISON OF NECK IMPACT RESPONSES BETWEEN VOLUNTEERS AND DUMMY: The neck loads computed for the volunteers are compared to those of the HY-III dummy. For the dummy, both measured and calculated results are shown to demonstrate the accuracy of the calculations for the various loading directions.

***CHIN-UP CASE*:** The neck moment for the HY-III for the chin-up mode is about twice that of the human volunteers (Figure 8), although the absolute magnitude is small. The HY-III demonstrates a damped oscillation pattern for the bending moment and shear force time-histories, which is not present in the time-histories of the relaxed human subjects. For

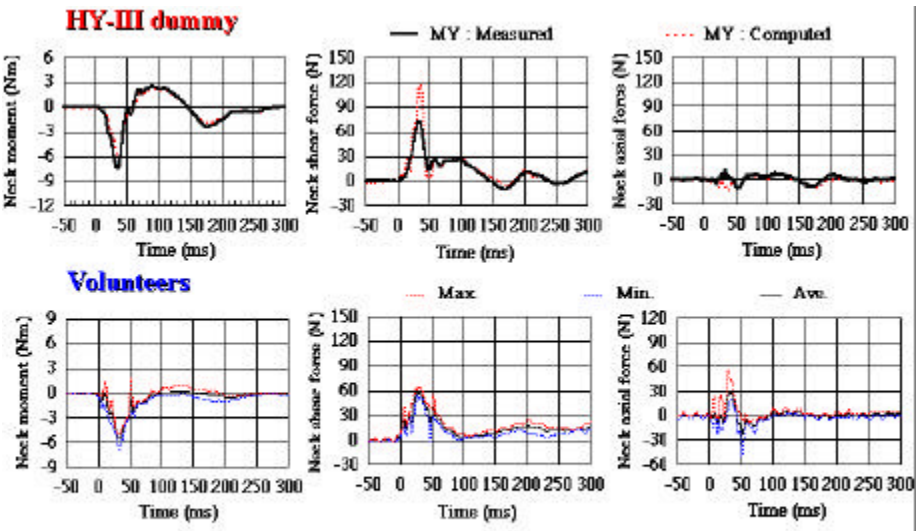


Fig. 9 Comparison of neck moment and forces between HY-III and Volunteers under the condition of rearward load applied to chin (Chin-backward: L)

both the human volunteers and the dummy, the shear force pulls the neck forward as the chin is being pulled upward. The neck axial compression force in the dummy is similar in peak and duration to those of the human subjects.

***CHIN-REARWARD CASE*:** For the human subjects, the soft tissues located between the chin and the upper region of neck may provide an alternate loading path, which is not accounting for in the simplified calculations. Consequently, it will be important in future studies to either account for this load alternate load path or modify the experimental methods to prevent contact of the chin with the soft tissues on the anterior surface of the neck, especially for higher intensity loading conditions.

***FOREHEAD-REARWARD CASE*:** The computed and measured moments for the HY-III show the largest discrepancy for the forehead-rearward loading scenario (Figure 10). Further studies re necessary to determine the causes of these differences

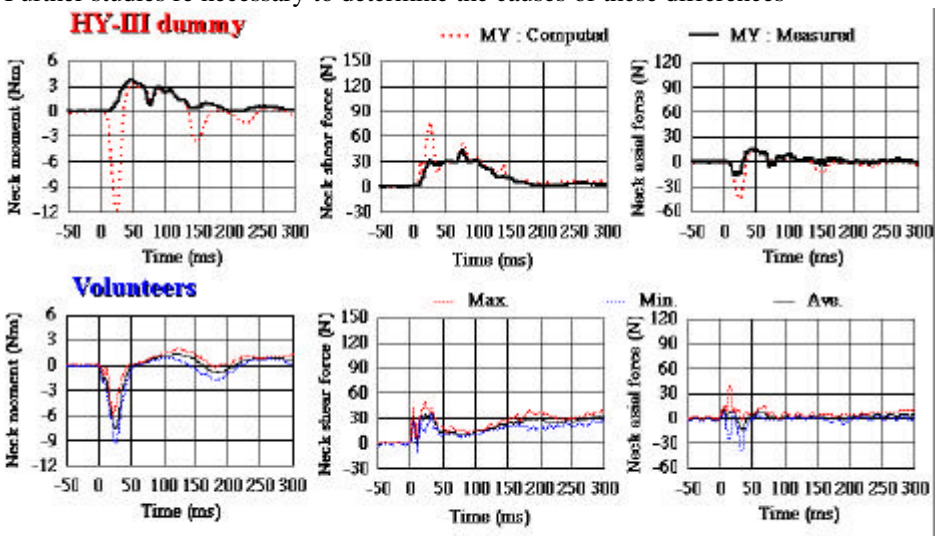


Fig. 10 Comparison of neck moment and forces between HY-III and Volunteers under the condition of rearward load applied to forehead (Head-bacward: U)

DISCUSSION

Various studies have been conducted in the past on the kinematics of the head and neck and neck injury thresholds [9)-14)]. In particular, Mertz and Patrick [14)] conducted frontal and rear impact sled tests using belted subjects including one volunteer and several cadavers. Based on an inertial loading configuration, they proposed neck injury tolerances for bending at the occipital condyles. However, there is currently little known data on the biomechanical response of the human neck under direct head contact loading.

Mertz and Patrick [14)] also developed moment-angle corridors for the flexion and extension response of the neck, where angular rotation of the head was measured with respect to T1. The overall moment experienced by the head is the sum of the moments produced by the OC and the moment produced by the surrounding musculature. The moment in the ligamentous occipital condyle joint is a function of the angle between the head and the top of the neck (Figure 1), not the angle between the head and T1. Forces and moments at the occipital condyles, not the forces and moments transferred through the surrounding musculature, are believed to be responsible for injury to the cervical spine ligaments and spinal cord. The data from this preliminary study will be used to partition the overall moment experienced by the head into the moments at the occipital condyles and the moments transferred the surrounding musculature.

The current study provides preliminary data on the characterization of the human head and neck during direct loading and suggests that head/neck kinematics of the relaxed human volunteers may be somewhat different from that of the HY-III for these low intensity loads. However, there are several issues that need to be addressed. First, it is not known if the differences observed in head/neck kinematics for this low-impact severity study are significant and if these differences will occur for higher-severity impacts associated with air bags. Second, the effects of the initial condition of the neck muscles, i.e. relaxed or tensed, on the head/neck kinematics and kinetics have not yet been analyzed. Third, although the preliminary data seems to indicate that the overall forces at the occipital condyles may be comparable between the human volunteers and the dummy, knowledge of the distribution of the forces and moments between the ligamentous cervical spine and the surrounding musculature is essential for developing a dummy head/neck system that interacts more biofidelically with the environment and for more accurately evaluating the potential for injury to the cervical spine and spinal cord.

Future studies are planned to address these issues and include testing with human volunteers, human cadavers, HY-III and THOR crash test dummies, and computational modeling. The focus of the human volunteer studies will be the load sharing between the cervical spine and surrounding musculature, while the focus of the human cadaveric studies will be neck tolerance under direct head/neck loading conditions. Since cadaveric musculature differs significantly from that of a living human, computational modeling will be used to address this limitation in the cadaveric studies and to better understand muscle activation in the volunteer studies. It is anticipated that the results of these future studies will be used to interpret the performance of current necks and to facilitate their improvement.

CONCLUSIONS

This preliminary study was conducted to determine the kinematics and kinetics of the human head and neck under direct loading scenarios that simulated typical patterns of air bag loading, but at a much lower severity. In general, the three relaxed volunteers had very similar patterns of head rotation with one another for each loading condition. However, because of its incorporation of effects of neck musculature in the design of the nodding blocks and its design for more severe crash loading severity, the HY-III showed a somewhat different pattern of head rotation from that of the relaxed volunteers in this study. Attempts to validate computed versus measured dummy neck moments and forces were successful in only one out of three loading conditions. Future studies are planned to determine the causes of these difference and also include testing with human volunteers, human cadavers, HY-III and THOR crash test dummies, and computational modeling. The focus of the human volunteer studies will be the load sharing between the cervical spine and surrounding musculature, while the focus of the human cadaveric studies will be neck tolerance under direct head/neck loading conditions. Since cadaveric musculature differs significantly from that of a living human, computational modeling will be used to address this limitation in the cadaveric studies and to better understand muscle activation in the volunteer studies. It is anticipated that the results of these future studies will be used to interpret the performance of current necks and to facilitate their improvement.

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