



# The Effects of Head-Supported Mass on the Risk of Neck Injury in Army Personnel

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**T**hroughout the evolution of military helmet systems, the amount of head-supported mass (HSM) worn by Army personnel has steadily increased. The U.S. Army Aeromedical Research Laboratory is working to establish recommended limits and guidelines for HSM that can be worn safely by soldiers. To support this effort, crash sled tests were conducted at APL's Impact Biomechanics Facility. These tests, simulating a typical frontal crash environment experienced by soldiers in both ground vehicles and rotary wing aircraft, were used to evaluate the effects of HSM on anthropomorphic test devices. A two-phase experimental design was used to evaluate the influence of independent HSM factors on occupant response. Injury measures, including neck forces and moments, were used to assess potential occupant injury. Test results provided statistically significant correlations between the independent factors and predicted injury, suggesting that future helmet design will be controlled by the selected injury measures.

## BACKGROUND

As military helmet systems have evolved, the amount of head-supported mass (HSM) has steadily increased. This is largely a result of advances in helmet safety and the need for head-supported devices (HSDs). Before the 1950s, aircrew flight helmets were nothing more than leather or cloth caps. Beginning in the 1950s, hard-shell helmets were introduced into service. These helmets provided a dramatic improvement in head impact protection as well as greater sound attenuation and better integral communications equipment. However, these benefits came at the price of increased helmet mass; the mass of these hard-shell helmets was 3 times that of their leather and cloth counterparts.<sup>1</sup> Furthermore,

with the advent of night vision systems during the 1980s, helmets began to serve as mounting platforms for numerous combat-essential HSDs (Fig. 1). Among these devices were night vision goggles, a chemical mask, an oxygen system, a head-up display, a forward-looking IR display, flash blindness and laser eye protection, and weapon aiming systems. With the addition of each of these systems, the necessary HSM for soldiers has steadily increased.

Fixed-wing aviation has followed a similar evolution. The introduction of the F-16 fighter jet in the early 1970s began a new generation of high-performance aircraft. This highly maneuverable aircraft combined rapid



**Figure 1.** Typical Army helmet used as a mounting platform for head-supported devices.

linear acceleration with a high onset rate of centripetal acceleration along the axis of the head and neck ( $G_z$ ). For the first time in aviation history, the pilot had become the limiting factor in the man-machine combination, and studies found neck injury to be a common occurrence among F-16 fighter pilots. In the mid-1980s, the introduction of the F/A-18 Hornet into the Navy and Marine Corps added a multidimensional increase in performance to the Fleet.<sup>2</sup> This carrier-based jet fighter was capable of more than  $9 G_z$  with an onset rate of more than  $18 G/s$ , placing greater load and stress factors on the pilot than had any previous Navy aircraft. The loads exceeded the physiologic capabilities of the pilots and potentially resulted in  $G$  loss of consciousness caused by  $G_z$  effects on cerebral perfusion (blood flow to the brain) as well as less severe, yet more frequent,  $G_z$ -induced neck injury.

The addition of HSDs has increased the magnitude of HSM worn by soldiers. As a result, there is an additional risk of neck injury caused by inertial loads generated during crashes. In a crash environment, the additional mass is likely to intensify stresses and strains on the neck due to acute loading. These crashes can occur in horizontal or vertical directions or a combination thereof. The horizontal crash direction, which is the focus of this research, simulates a frontal crash for a ground vehicle or the horizontal component of an aircraft collision.

## CURRENT RESEARCH EFFORTS

The U.S. Army Aeromedical Research Laboratory (USAARL) at Fort Rucker, Alabama, is conducting research to establish recommended limits and center of gravity (CG) locations of HSM that can be worn safely by soldiers.<sup>3-5</sup> To support this effort, APL has conducted a series of frontal impact sled tests using the Hybrid III mid-sized male anthropomorphic test device (ATD). The ATD was outfitted with an adjustable-weight helmet fixture (the Gladiator) provided by the USAARL, which allows the total helmet weight and weight distribution to be varied over a wide range. The objective of this study was to conduct a series of simulated frontal crashes using the Hybrid III ATD to determine the effects of helmet weight properties on the risk of sustaining a neck injury. Thirty tests were conducted using three helmet weights, three impact speeds, and five different positions for the helmet CG in both the vertical and horizontal directions. Forces and accelerations in the head, neck, and chest of the ATD were recorded and compared to known Injury Assessment Reference Values (IARVs) for the dummy used. These injury values were then used to determine the relative risk of injury for each helmet weight configuration.

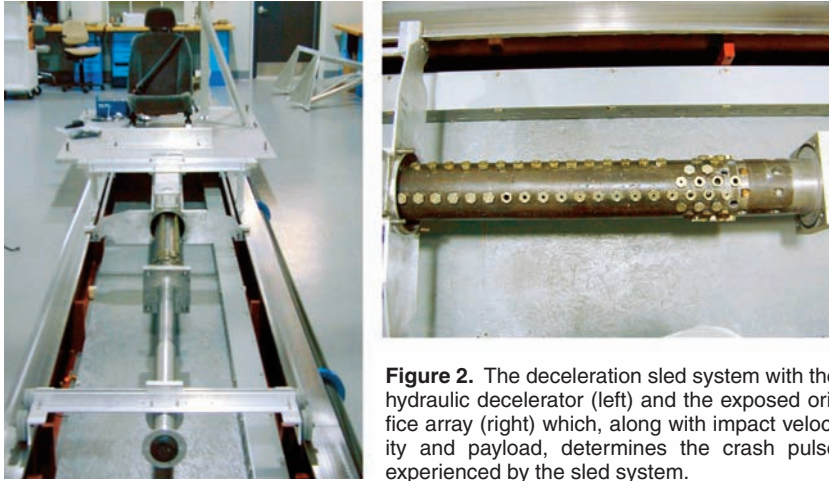
## EXPERIMENTAL METHODS

### Approach

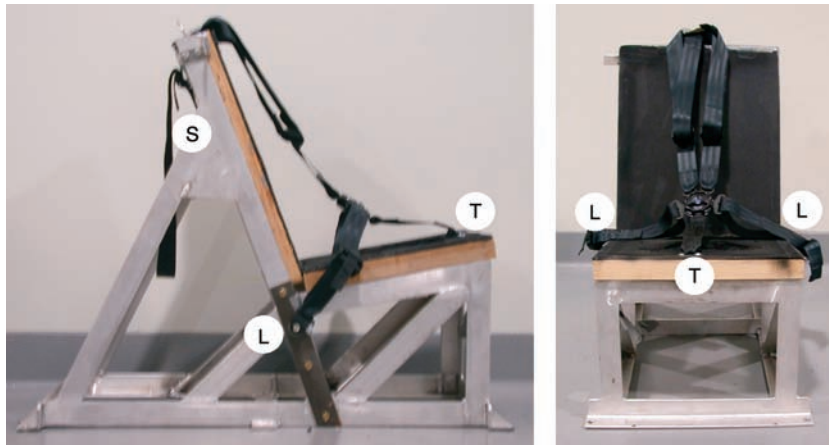
To achieve the stated objective, APL conducted sled tests at its Impact Biomechanics Facility that simulated vehicle crash environments experienced by soldiers. Testing was performed on a Hybrid III mid-sized male ATD outfitted with HSM of various magnitudes and locations. The ATD was placed in a rigid seat for repeatability and restrained with a typical five-point helicopter restraint harness. Considering the innumerable possible HSM configurations and sled impact velocities available for testing, care was taken to design experiments that would allow a representative set of tests to be run that were statistically strong enough to extract ATD response information but did not require testing of each distinct configuration. Recorded responses included neck forces and moments; head, torso, and sled accelerations; and additional measurements related to ATD kinematics.

### Test Setup

The vehicle collision environment was simulated with a VIA Systems Horizontal Impact Test Sled. The APL-operated sled system (Fig. 2) can be tuned to generate a wide range of vehicle crash pulses and can reach impact speeds of more than 50 mph. The system contains a hydraulic decelerator that absorbs the kinetic energy from the sled carriage and payload by forcing fluid through a series of orifices. The selected orifice array determines the crash pulse experienced by the sled system.



**Figure 2.** The deceleration sled system with the hydraulic decelerator (left) and the exposed orifice array (right) which, along with impact velocity and payload, determines the crash pulse experienced by the sled system.



**Figure 3.** Rigid seat system showing attachment points for the shoulder harness (S), lap belts (L), and lap belt tie-down strap (T).

The ATD was positioned in a crashworthy seat system (Fig. 3) designed and fabricated by APL. The system was designed to provide a rigid seat for the mid-sized Hybrid III ATD and mounting locations for the five-point UH-60 helicopter restraint system, which included shoulder restraints and a belt adjuster mounted to the back of the seat as well as a lap belt and adjuster bolted to each side of the seat. The fifth restraint, a lap belt tie-down strap with buckle, was anchored to the seat approximately 15 in. forward of the seatback. The seatback was reclined  $20^\circ$  from vertical, and the seat bottom had a  $10^\circ$  pitch to simulate the typical seating posture within a military aircraft.<sup>6</sup>

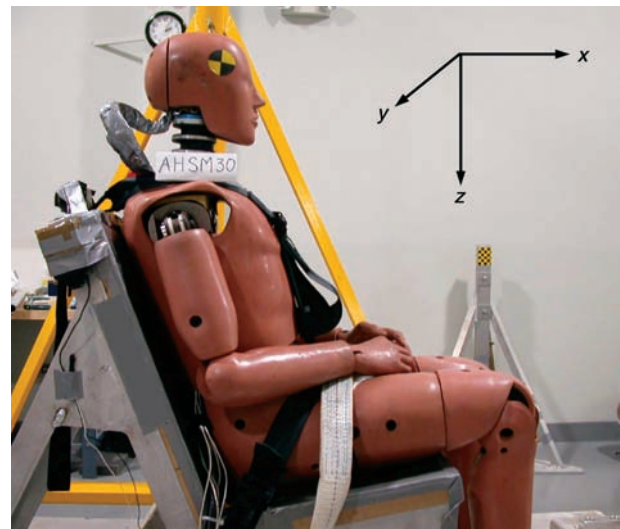
Figure 4 illustrates the coordinate systems used for the ATD and sled. The dummy was instrumented with triaxial accelerometer arrays at the CG of both the head and torso as well as two accelerometers located at the base of the cervical spine (T1). Six-axis load cells, sensing both forces and moments, were installed in the upper and lower neck, and a three-axis load cell was installed in the lumbar spine. Angular rate sensors were attached at the head CG and the spine at T1.

Another accelerometer was mounted to the sled to record the deceleration pulse during impact. All of the designated polarities, filtering, and sampling rates were determined in accordance with recommended practices established by the Society of Automotive Engineers.<sup>7</sup>

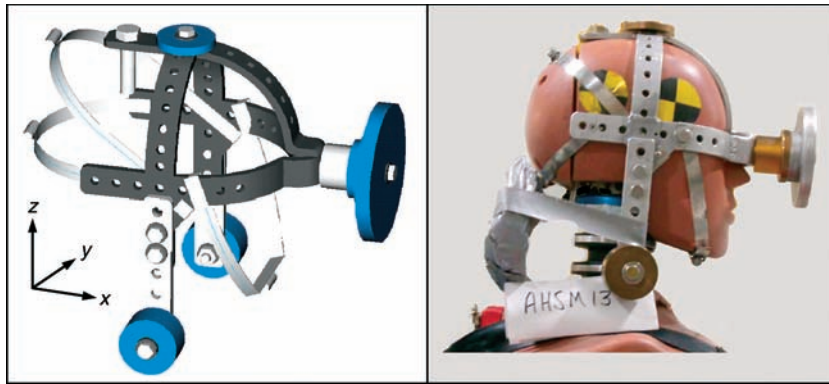
All sensor data were collected at a sampling rate of 10 kHz using an onboard TDAS Pro data acquisition system (Diversified Technical Systems, Inc.) mounted to the sled baseplate. Dummy kinematics were recorded by an onboard IMC Phantom 4 digital video camera (Vision Research, Inc.). The digital video images were recorded at 1000 frames/s at  $512 \times 512$  resolution, which allowed analysis of tracking movement and post-test motion.

The Gladiator adjustable-weight helmet fixture (Fig. 5) allowed the HSM magnitude and location to be varied. The fixture was designed for use with the standard mid-sized male Hybrid III head. It was secured to the head using the existing tapped hole located at the crest of the manikin's head along with three stainless steel straps to prevent slippage. The Gladiator, composed of 0.25-in.-thick aluminum and weighing approximately

2 lb, contains threaded holes at various positions along its contoured surface. Weights were bolted to the fixture



**Figure 4.** Initial position of the ATD in the rigid seat system. The sled and ATD coordinate systems are shown.



**Figure 5.** ProEngineer model (left) and actual test configuration (right) for AHSM13 showing the Gladiator device and added weights. This particular test configuration (no. 4) included 7.5 lb of added weight, with the CG located 3.5 in. forward and 1 in. above the occipital condyles.

in strategic positions to add mass and to alter the HSM's CG. The CG was measured with respect to the occipital condyles (projections from the base of the skull that form the articulation between the skull and spine).

### HSM Configurations

The placement and distribution of weights for specific HSM configurations were determined using a computer model. The geometry and density of the Gladiator frame, individual weights, and hardware were measured and incorporated into a ProEngineer computational model. The amounts and locations of weights for specific HSM configurations were found by adding and removing components to the model. For example, test AHSM13 added 7.5 lb of HSM and had a CG located 3.5 in. anterior and 1 in. superior to the occipital condyles. The resulting Gladiator arrangement is shown in Fig. 5.

### Experimental Design

This project required the study of several factors within certain time and resource limitations. It was therefore decided to evaluate the factors by using a set of statistically designed experiments. This method of experimentation is well known to maximize the information yield while minimizing the number of experiments to be performed. However, it also requires the factors to be studied simultaneously by a judicious choice of the combination of factors. This method is commonly referred to as "Design of Experiment"<sup>8</sup> and can provide substantial experimental cost and time savings.

### Taguchi Screening Test Series

The initial experimental test series used a screening design involving four factors—impact velocity, helmet weight, horizontal CG (*x*) location, and vertical CG (*z*) location—each with three levels (Table 1). As previously mentioned, the CG locations were measured relative to the occipital condyles. Since the CG

was varied with added weights in the *xz* plane only, the CG in the *y* direction remained unchanged at zero. Therefore, the horizontal CG (*y*) location was not a variable examined in this study. Aside from determining the primary effects of the helmet-related factors, this initial screening series was also useful in determining the impact velocity that would be associated with a moderate risk of sustaining neck injuries.

Using a highly fractionated factorial design referred to as the Taguchi experimental design,<sup>9</sup> the four independent variables in Table

1 were tested at the three equally spaced levels with nine independent trials. This design considers the main effects of each independent variable but not two-factor interactions. As part of the experimental design, the test cases were conducted in random order.

### Box-Behnkin Test Series

The initial Taguchi screening test series indicated that an impact velocity of 40 ft/s was necessary to produce a moderate risk of neck injury. It was also found that impact velocity had an overwhelming effect on the dependent variable responses, including the measured forces and moments in the neck. To evaluate the helmet-related factors in more detail, a second test series was designed holding the impact velocity constant at 40 ft/s; the remaining three variables (helmet weight, horizontal CG location, and vertical CG location) were studied with a response surface design. More specifically, the Box-Behnkin design with three center points was chosen,<sup>10</sup> resulting in a design with 16 independent runs

**Table 1.** Taguchi test series matrix.

Test ID	Impact velocity (ft/s)	Helmet weight (lb)	CG ( <i>x</i> ) (in.)	CG ( <i>z</i> ) (in.)
AHSM01	29.6	3.5	1.5	3.0
AHSM02	19.9	5.5	1.5	5.0
AHSM03	19.8	3.5	-0.5	1.0
AHSM04	30.0	5.5	3.5	1.0
AHSM05	20.1	7.5	3.5	3.0
AHSM08	29.8	7.5	-0.5	5.0
AHSM10	39.0	3.5	3.5	5.0
AHSM11	39.2	5.5	-0.5	3.0
AHSM12	39.3	7.5	1.5	1.0

conducted in random order. Each variable was studied at five levels: helmet weight varied from 2.9 to 8.1 lb, the horizontal CG position varied from -1.1 to 4.1 in. relative to the occipital condyles, and the vertical CG position varied from 0.4 to 5.6 in. above the occipital condyles. Figure 6 illustrates the relative position between the helmet CG locations and the CG of the head alone with respect to the occipital condyles. Note that multiple tests with varying helmet weight were conducted at each CG position. The test series matrix is provided in Table 2.

## INJURY CRITERIA

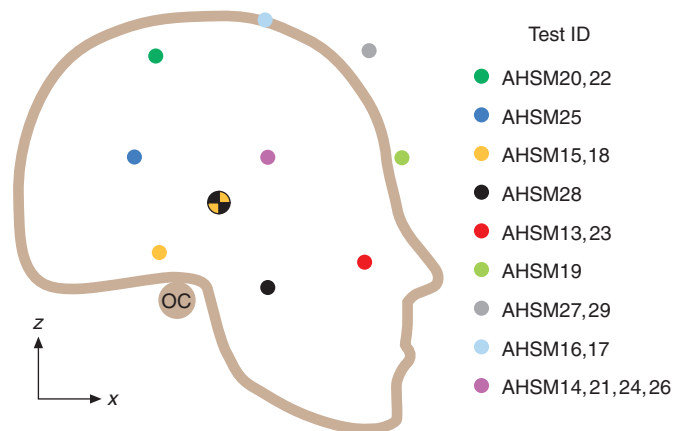
### Injury Assessment Reference Values

As noted earlier, peak upper neck forces and moments from the test results were compared to established IARVs. The IARVs provide a gauge for the probability of injury as a result of a crash. If the peak force or moment value recorded from the occupant during the crash exceeds the corresponding IARV, it is an indication that the occupant may have sustained a serious neck injury. The Abbreviated Injury Scale (AIS), used to classify impact injuries between 1 (minor) and 6 (untreatable), classifies any response exceeding an IARV as  $AIS \geq 3$ . (The referenced IARVs are listed in Tables 3 and 4, which are discussed under “Test Results.”)

### $N_{ij}$ Neck Injury Criterion

The  $N_{ij}$  neck injury criterion measures neck injury risk by considering the combined effects of axial forces and flexion/extension moments. The formula for  $N_{ij}$  calculation is as follows:

$$N_{ij} = \frac{F_z}{F_{int}} + \frac{M_y}{M_{int}}$$



**Figure 6.** HSM CG location for each test in the Box-Behnklin series. Coordinates (x, z) are given with respect to the occipital condyles (OC).

**Table 2. Box-Behnklin test series matrix.**

Test ID	Impact velocity (ft/s)	Helmet		
		weight (lb)	CG (x) (in.)	CG (z) (in.)
AHSM13	40.1	7.5	3.5	1.0
AHSM14	40.2	5.5	1.5	3.0
AHSM15	40.6	3.5	-0.5	1.0
AHSM17	39.9	5.5	1.5	5.6
AHSM18	39.9	7.5	-0.5	1.0
AHSM19	39.8	5.5	4.1	3.0
AHSM20	39.8	7.5	-0.5	5.0
AHSM21	39.6	5.5	1.5	3.0
AHSM22	39.6	3.5	-0.5	5.0
AHSM23	39.6	3.5	3.5	1.0
AHSM24	39.7	2.9	1.5	3.0
AHSM25	39.6	5.5	-1.1	3.0
AHSM26	39.6	8.1	1.5	3.0
AHSM27	39.8	3.5	3.5	5.0
AHSM28	39.6	5.5	1.5	0.4
AHSM29	39.8	7.5	3.5	5.0

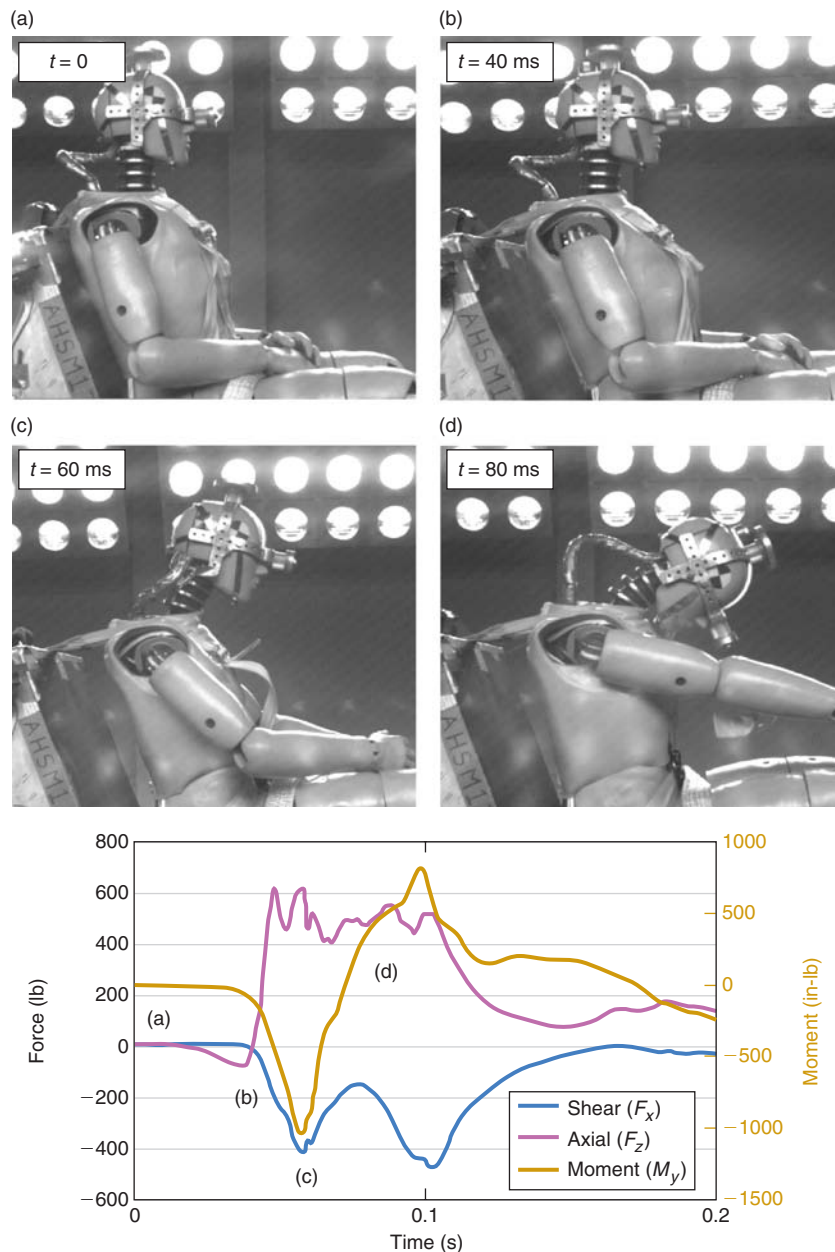
The intercept values ( $F_{int}$  and  $M_{int}$ ) for the 50th percentile male Hybrid III ATD<sup>11</sup> are used along with the data measured by the upper neck load cell ( $F_z$  and  $M_y$ ) to compute  $N_{ij}$ . A characteristic result provides not only the magnitude of  $N_{ij}$  but also the mode in which it occurred. Because of dummy kinematics, the potential injury modes relevant to this study occur when the neck is in a state of tension-extension (TE) or tension-flexion (TF). These injury modes are apparent in Figs. 7c and 7d, respectively. The calculated  $N_{ij}$  values were compared to a reference value of 1.00. The magnitude of the calculated value with respect to the reference is a

measure of the probability of sustaining a serious ( $AIS \geq 3$ ) neck injury. For the upper neck, a value of 1.00 represents a 15% probability of sustaining a serious injury using established injury probability formulas.

## TEST RESULTS

### Taguchi Screening Test Series

The primary objectives of the nine initial screening tests were to better understand the relationships between the independent variables and the dummy response and to determine if any of the four factors overwhelmingly predicted measures of potential injury risk.



**Figure 7.** Captured images from AHSM17 illustrating the ATD movement phases. The plot provides the upper neck forces (shear and axial) and moments for the test. The initial image (a) was taken at the time of impact; the later images (b, c, and d) were taken at the times indicated after impact.

Results from this test series were used to determine the factors for the second series of tests.

Table 1 lists the tests that were run, including the test ID, impact velocity, and HSM configuration. Tests were run at three different impact velocities: 20, 30, and 40 ft/s. The characteristic crash deceleration pulses for the 20- and 30-ft/s impact velocities had peak decelerations of 10 g and 21 g, respectively. Observations from the test conditions at 30 ft/s and anticipated harsh conditions at higher speeds prompted an adjustment

to the deceleration system for the 40-ft/s tests. The result was a 26-G peak deceleration with a duration of approximately 105 ms.

Table 3 provides the peak upper neck forces and moments as well as the corresponding injury measures for the tests run in the Taguchi series. Any peak values exceeding the specified injury thresholds are noted in bold.

### Box-Behnkin Test Series

Again, based on the results of the initial screening test series, all subsequent tests were conducted at a constant impact velocity of 40 ft/s. With four independent factors now reduced to three, a series of 16 tests (1 duplicate) was designed to assess the influence of the remaining helmet-related independent variables on dummy response. Table 2 lists the test ID along with the other impact parameters; Fig. 6 plots the helmet CG location on the surface of the ATD head. Since impact velocity was unchanged, the deceleration profile remained the same for all tests. The peak deceleration was approximately 26 G, with a pulse duration of approximately 105 ms.

Table 4 provides the peak upper neck forces and moments for the tests run in the Box-Behnkin series as well as the corresponding injury measures calculated from the test data. Any peak values exceeding the specified injury thresholds are noted in bold. These data were statistically analyzed for correlations to the three independent factors (helmet weight, CG position in  $x$ , and CG position in  $z$ ). Table 5 lists the results of the statistical analysis

for the upper neck. Any significant correlations ( $p < 0.01$ ) are shown in bold.

### DISCUSSION AND CONCLUSIONS

The main objectives of the Taguchi screening test series were to determine the primary effects of each independent factor on the measured occupant head/neck response and the level of impact velocity that would produce a moderate risk of neck injury. This initial Design of Experiments test series screened numerous

**Table 3. Peak upper neck forces and moments for the Taguchi test series.**

Test ID	Shear force (lb)	Tensile force (lb)	Ext. moment (lb-in.)	Flex. moment (lb-in.)	$N_{ij}$ (TF)	$N_{ij}$ (TE)
AHSM01	301	455	<b>566</b>	1000	0.43	0.55
AHSM02	132	218	204	628	0.23	0.24
AHSM03	125	169	204	425	0.18	0.22
AHSM04	363	439	<b>1044</b>	708	0.34	0.87
AHSM05	168	229	416	584	0.19	0.40
AHSM08	295	629	372	1062	0.63	0.59
AHSM10	438	565	<b>1027</b>	1071	0.60	0.99
AHSM11	540	690	<b>628</b>	1159	0.78	0.67
AHSM12	636	660	<b>1690</b>	929	0.57	<b>1.41</b>
IARV limits	695	740	503	1678	1.00	1.00

Note: Values in bold exceed IARVs.

**Table 4. Peak upper neck forces and moments for the Box-Behnkin test series.**

Test ID	Shear force (lb)	Tensile force (lb)	Ext. moment (lb-in.)	Flex. moment (lb-in.)	$N_{ij}$ (TF)	$N_{ij}$ (TE)
AHSM13	<b>784</b>	<b>823</b>	<b>1595</b>	752	0.52	<b>1.74</b>
AHSM14	532	<b>773</b>	<b>855</b>	518	0.60	<b>1.24</b>
AHSM15	445	622	<b>687</b>	880	0.68	<b>1.00</b>
AHSM17	480	615	<b>740</b>	797	0.73	<b>1.03</b>
AHSM18	550	<b>748</b>	<b>846</b>	873	0.84	<b>1.20</b>
AHSM19	601	715	<b>1154</b>	515	0.50	<b>1.44</b>
AHSM20	519	<b>813</b>	<b>608</b>	1164	<b>1.00</b>	0.87
AHSM21	477	<b>743</b>	<b>775</b>	545	0.58	<b>1.14</b>
AHSM22	416	587	502	958	0.77	0.73
AHSM23	460	528	<b>1040</b>	716	0.45	<b>1.20</b>
AHSM24	426	532	<b>564</b>	665	0.56	0.83
AHSM25	463	637	520	1084	0.84	0.76
AHSM26	542	720	<b>943</b>	574	0.71	<b>1.28</b>
AHSM27	387	544	<b>802</b>	774	0.61	<b>1.03</b>
AHSM28	483	545	<b>1040</b>	695	0.57	<b>1.18</b>
AHSM29	487	671	<b>1066</b>	817	0.71	<b>1.31</b>
IARV limits	695	740	503	1678	1.00	1.00

Note: Values in bold exceed IARVs.

**Table 5. Upper neck correlation of independent factors to dependent variables for the Box-Behnkin series.**

Factor	Shear	Tension	Extension	Flexion	$N_{ij}$ (TF)	$N_{ij}$ (TE)
Helmet weight	<b>0.002</b>	<b>0.001</b>	<b>0.0003</b>	0.225	<b>0.002</b>	<b>0.0001</b>
CG (x)	<b>0.095</b>	<b>0.685</b>	<b>0.0001</b>	<b>0.006</b>	<b>0.0001</b>	<b>0.0001</b>
CG (z)	0.029	<b>0.610</b>	<b>0.0002</b>	0.027	<b>0.002</b>	<b>0.0003</b>

Note: Values in bold are statistically significant predictors ( $p < 0.01$ ) of outcome. Values in red indicate an inversely proportional relationship between the independent and dependent factors. All other relationships are directly proportional.

factors over a range of values with a minimal number of tests and, therefore, provided a large savings in effort and cost. Statistical analysis of the Taguchi data established a strong correlation ( $p < 0.01$ ) between impact velocity and the various neck injury measures. Although weak correlations were found between helmet weight and upper neck forces, and also between CG location and upper neck shear force and moments, these findings were not statistically significant. The domination of the impact velocity on the overall occupant response likely explains why these correlations between the remaining independent factors and dependent variables were not statistically significant. It was for this reason that the impact velocity was held constant for the Box-Behnkin test series.

The Taguchi test series results in Table 3 show that the majority of the measured responses were below the IARVs for serious AIS  $\geq 3$  injuries. The only responses that exceeded the IARV levels were the extension moments for tests AHSM01, 04, 10, 11, and 12. Tests 01 and 04 were conducted at an impact velocity of 30 ft/s, whereas tests 10–12 were all performed at 40 ft/s.

Since the focus of this study was to evaluate the risk of sustaining moderate to serious neck injuries related to impacts with a predominantly  $G_x$  component, the main Box-Behnkin series of tests was conducted at the higher impact velocity of 40 ft/s. For this series of tests, the three independent variables (helmet weight, CG ( $x$ ) location, CG ( $z$ ) location) were varied over their respective ranges (Table 2).

Statistical analysis of the Box-Behnkin data (Table 5) indicated strong correlations ( $p < 0.01$ ) between the independent factors of weight and CG location (both  $x$  and  $z$ ) with  $N_{ij}$  and upper neck extension moment. Both helmet weight and horizontal ( $x$ ), but not vertical ( $z$ ), CG position proved to be strong determinants in lower neck flexion and  $N_{ij}$ . Helmet weight correlated strongly with upper neck shear and tensile forces. The only factor that was a good predictor of upper neck flexion moment was the horizontal position of the CG. No strong correlations were found between the independent helmet-related variables and the lower neck tensile and shear forces.

A review of the results provided in Table 4 indicates that the extension moment IARV was exceeded in all but one test in the Box-Behnkin series, whereas the flexion moment limit was not exceeded in any tests. The tensile force IARV was exceeded in 5 of the 16 tests (AHSM13, 14, 18, 20, and 21). Three of these five tests were with a helmet weight of 7.5 lb, while the remaining two tests were with a helmet weight of 5.5 lb. None of the tests conducted with the 3.5-lb helmet, the lowest level of added HSM, exceeded the individual peak force or moment thresholds for serious injury in the upper neck.

An evaluation of the calculated  $N_{ij}$  values for the Hybrid III dummy indicates that there is a greater

likelihood of sustaining a tension-extension injury than a tension-flexion injury for this crash environment. This can be explained in part by looking at the dummy kinematics for a typical 40-ft/s crash test, as shown in Fig. 7. As the impact event begins, the head translates forward with respect to the torso but does not rotate. The anterior head translation causes localized extension measured by the upper neck load cell. The upper neck tension rises to a peak value and holds relatively constant for the duration of the impact. After 60 ms, the upper neck bending moment transitions from extension into flexion as the chin rotates down toward the chest. The magnitude of the extension moment is greater than the flexion moment, which corresponds to a greater indication of potential neck injury as a result of extension ( $N_{TE}$ ) as compared with flexion ( $N_{TF}$ ).

Examination of the statistical analyses of the test results offers insight into these effects, including significant correlations between HSM CG location and injury measures. Table 5 not only provides the statistically significant outcomes but also shows whether the independent variables are directly or inversely proportional to the injury measures. For example, it is evident that extension is not only significantly affected by HSM position and magnitude, but that the more anteriorly and inferiorly the CG is located on the head, the larger the extension moment becomes. Similarly, the results indicate that neck flexion moment would be expected to increase as the CG moves in the posterior and superior directions.

These results provide evidence that the design of helmet systems depends on the selection of the critical injury measures. Therefore, if neck extension is selected as the critical injury measure, then the data would suggest that, in an effort to mitigate injuries, HSM be shifted to the posterior-superior portion of the head. However, this would contradict the results from the measured flexion moment, which shows an increase in risk of injury as mass is added to the posterior-superior portion of the head. It is important to note that the strength of these correlations varies, with extension being the strongest. The injury criteria can be evaluated in much the same manner. These observations may become an important factor in considerations of future helmet design and the attachment points of HSDs.

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