INFLUENCE OF ADDED HEAD MASS PROPERTIES ON HEAD/NECK LOADS DURING STANDARD HELICOPTER IMPACT CONDITIONS

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ABSTRACT

Thirty-degree pitch-down helicopter crash pulses were examined at the Patuxent River Horizontal Accelerator facility. The pulses used were representative of standard seat qualification crash corridors for a variety of rotary wing platforms. The primary objectives of this effort were 1) to quantitatively determine the effect of varying helmet weight and center of gravity (CG) during simulated rotary wing crash scenarios, and 2) to perform data analysis using existing injury criteria to identify maximum requirements for helmet weight and CG for the extremes of the rotary wing aviator population. Quantification of risk was based upon aviator size, helmet mass properties and impact severity. Testing included a Hybrid III 95th percentile male, a 50th percentile male, and a 5th percentile female. In order to achieve the necessary helmet weight and CG values required for the tests, a head-mass fixture was developed and used in place of the manikin's head that allowed weights to be added both forward and laterally on the head to generate a wide array of weight and CG configurations. Modeling of the system was performed using MADYMO, which was used to refine the test matrix for the weight and CG locations that would most likely define the mass properties envelope from the known criteria force/moment limits. A comprehensive analysis of the data was performed using available injury predictors to determine the likelihood of injury to the upper and lower cervical spine. The Nij cervical injury criterion developed by the National Highway and Traffic Safety Administration (NHTSA) was adapted and used to determine the risk of cervical injury. An analysis of variance (ANOVA) with a post-hoc Tukey-Kramer test to determine the source of the difference was conducted on the principal neck parameters to determine whether any of the factors were statistically significant (defined as $p \leq 0.05$). Results indicated that pulse severity was the most dominant variable examined during the study. The effect of gender was more related to the different values for the neck injury limits, which were scaled according to weight. As a result of the study, guidelines based on helmet weight, CG, aviator size and impact severity were generated.

INTRODUCTION

By design, helicopters have some level of intrinsic crashworthiness. When a helicopter crashes in a predominantly vertical impact, the landing gear assembly first compresses to failure. Once the landing gear system has failed, the airframe structure crushes. This crush, transmitted to the floor of the cockpit and cabin, causes the floor to buckle and transmits the force to the seat. The pilot and co-pilot seats are equipped with energy absorbers that attenuate the force to allow 15 Gz to be transmitted to the occupant. The seat strokes and the occupant lags the seat in this event. Once the seat bottoms out the occupant receives a marked acceleration (Gz) overshoot [1]. This force is transmitted up the spine and accounts for the large concern over lumbar and thoracic vertebral fracture. The crash event, however, is not limited only to the vertical axis: the horizontal and lateral crash loads can be extreme as well. Most commonly the horizontal loads are of concern. After bottoming out the occupant begins pitching forward in the seat. The restraint systems, typically a four or five-point restraint, allow forward movement due to the stretch of the webbing material. Once this stretch is complete the occupant's thorax is abruptly halted, but the head and neck continue to translate and rotate. This motion of the head and neck results in cervical tension and shear that approach and can exceed injurious levels [2].

Helmet mounted devices (HMD's) that are worn on the head are used for air-to-air and air-to-ground targeting, night vision, forward looking infra-red views, and flight control information. As the number of features increases, the total weight of the system increases. Earlier systems that provided a number of features, in addition to the basic functions of a helmet, weighed close to 6 pounds. Advances in modern materials have reduced this overall weight to 4.8 pounds or less. While the total weight is important, the major factor in the injury potential is the location of the center-of-gravity (CG) of the head and helmet system combined. Typically the CG is placed forward of the occipital condyles and results in high pitch moments about the occipital condyles (C0/C1) and C7/T1 which are consistent with cervical injury. The neck injury

Report Documentation Page

Form Approved
OMB No. 0704-0188

1. REPORT DATE **SEP 2004**	2. REPORT TYPE **N/A**	3. DATES COVERED **-**
4. TITLE AND SUBTITLE **Influence of Added Head Mass Properties on Head/Neck Loads During Standard Helicopter Impact Conditions**		5a. CONTRACT NUMBER 5b. GRANT NUMBER 5c. PROGRAM ELEMENT NUMBER
6. AUTHOR(S)		5d. PROJECT NUMBER 5e. TASK NUMBER 5f. WORK UNIT NUMBER
7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES) **NAVAIR Escape Systems Engineer, Computer Modeling Engineer Escape & Crashworthiness Division 48110 Shaw Road, Bldg 2187, Suite 1280 Patuxent River, MD**		8. PERFORMING ORGANIZATION REPORT NUMBER
9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)		10. SPONSOR/MONITOR'S ACRONYM(S) 11. SPONSOR/MONITOR'S REPORT NUMBER(S)
12. DISTRIBUTION/AVAILABILITY STATEMENT **Approved for public release, distribution unlimited**		
13. SUPPLEMENTARY NOTES **Published in the Proceedings of the Forty Second Annual SAFE Association Symposium, Held in the Salt Lake City, Utah, September 27-28, 2004. SAFE Assocation, Post Office Box 130, Creswell, OR 97426-0130. http://www.safeassociation.org, The original document contains color images.**		
14. ABSTRACT		
15. SUBJECT TERMS **Safe**		

16. SECURITY CLASSIFICATION OF:			17. LIMITATION OF ABSTRACT	18. NUMBER OF PAGES	19a. NAME OF RESPONSIBLE PERSON
a. REPORT **unclassified**	b. ABSTRACT **unclassified**	c. THIS PAGE **unclassified**	**SAR**	**21**	

Standard Form 298 (Rev. 8-98)
Prescribed by ANSI Std Z39-18

potential in helicopter crashes is due to force transmission from the vertical acceleration component, extreme head and neck motion which places the neck in hyperflexion and distraction, and neck compression and shear due to head contact with the crew station or cabin structures, sighting systems, and flight controls [3,4].

In addition, female pilots are a part of combat missions, where historically they have not been. Crashworthiness design criteria based on a male population is not applicable to females. Recent findings in vertebral gender differences indicate that males and females of approximately the same size are at equal levels of risk to thoracic and lumbar injuries, but a gender difference may exist in the cervical spine. At this region, females are predicted to have a 13% decrease in vertebral strength versus same-sized males [5]. With the added weight and increased use of HMD's, the potential for injury is increased.

Currently, there is a lack of comprehensive data that can effectively predict the probability of cervical injury during helicopter impact for varying added head masses and all aircrew sizes. Modern systems are being fielded without a complete understanding of the risks that the aviator will be subjected to in the event of a crash. Guidelines based on helmet weight, CG, aviator size and impact severity are required by helmet and HMD manufacturers to develop helmet systems that may safely be worn by rotary wing crewmembers in the event that an impact occurs.

OBJECTIVES

The primary objectives of this effort were:

a) To quantitatively determine the effect of varying helmet weight and center of gravity (CG) during simulated rotary wing crash scenarios using the MADYMO models. The MADYMO models were used to make predictions of inertial loads upon the neck as a result of the helmet mass properties and to develop the mass/CG test matrix used for the system level testing.
b) To use system level Horizontal Accelerator testing to validate the predictions of the computer simulations and provide quantitative data regarding baseline and advanced helmet configurations.
c) To perform data analysis using cervical injury criteria to identify maximum requirements for helmet mass and CG for the entire rotary wing aviator population. Recommendations of risks were based upon aviator size, helmet mass properties and impact severity.

TEST METHODOLOGY

Analysis of Naval Helicopter Mishap Data

Information was compiled on available Naval helicopter mishap data [2,6-9]. Data regarding Naval aircraft impact conditions was analyzed and plotted as cumulative frequency versus resultant velocity for both land and water impacts (Figure 1). Only impacts that were deemed survivable were included on the figure. Significant survivable accidents were defined as those in which substantial structural damage occurred and one or more major injuries to the occupants. Non-survivable impacts were defined as those in which the impact acceleration environment exceeded the known limits of human tolerance, and/or the occupied volume was compromised such that survival would have been unlikely. The rotary wing platforms that were included in the data were: AH-1, UH-1, H-2, H-3, H-46, and H-53.

From this data it was possible to determine the frequency of occurrence for each of the three crash pulse velocities that were used during the testing (Table 1). For example, the low severity pulse represents 63% of all survivable Naval helicopter mishaps. This means that 63% of the reported mishaps had a resultant velocity that was at or below this level. Or, in other words, 37% of survivable accidents occurred above this resultant impact velocity. These pulses were originally derived from dynamic structural crashworthy seat requirements tied to specific aircraft platforms. However, it is unclear where these requirements were derived.

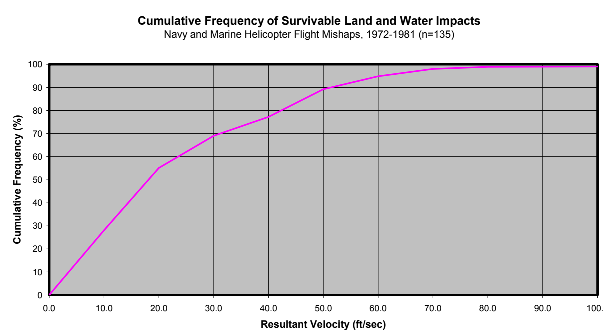

Figure 1: Cumulative Frequency of Survivable Land & Water Mishaps

Table 1: Cumulative Frequency of Survivable Impacts
Navy and Marine Helicopter Flight Mishaps (1972-1981)

Pulse Severity Classification	Proposed Test Crash Pulse	Frequency of Survivable Impacts (%)		
	Resultant Vel. (Ft/sec)	Land	Water	Total
Low	25.0	65%	58%	63%
Medium	31.5	74%	66%	71%
High	50.0	93%	84%	89%

Computer Modeling

A computer model characteristic of a seated figure was validated using the MADYMO (MAthmatical DYnamic MOdel) modeling software (Figure 2). The MADYMO model provided numerous advantages over traditional testing methods such as repeatability, cost and versatility. The model was validated using data from an earlier phase of testing from this effort in which sled runs were performed with an HGU-84/P helmet. After model validation, head weight, center of gravity and moments of inertia were varied to determine the effect of these mass properties on the dynamic response and joint forces/moments of the occupant. An analysis of neck injury was conducted on the results. Once the models had identified approximately where the tolerable limits of helmet weight and center of gravity appeared to be (for a given occupant size and pulse severity), systems level horizontal accelerator tests were performed at those limits. The data from the 4.0 lbs. added head mass were compared with the results of the simulations, as this was the only test configuration that was common to all occupant sizes and test pulses used during the program. The mass properties of the test fixture, used for the computer models, using the 4.0 lbs. added head mass configuration are shown in Table 2.

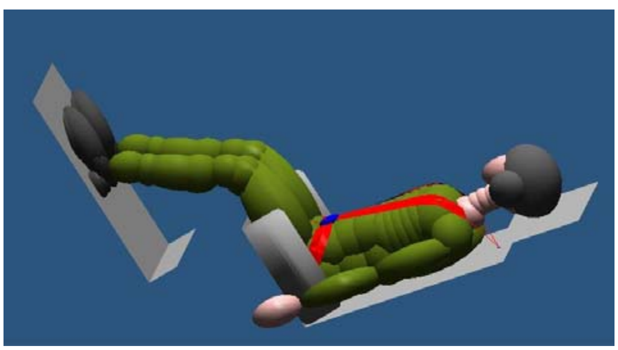

Figure 2: Seated MADYMO model

Computer modeling was also used to define the crash pulses used during the Horizontal Accelerator testing. It was determined that the rigid crashworthy seat imparted a much greater acceleration to the seated occupant than an actual stroking seat would. Cost and schedule prohibited using stroking seats for all of the tests. However, simulations were used to determine the acceleration profile that a typical stroking seat would 'see' during a standard helicopter impact test and this profile was applied to the rigid seat. Since it was not possible for the Horizontal Accelerator to match this pulse exactly, an approximated "rounded" pulse was used and simulations were conducted to determine whether this pulse approximation would adversely affect the occupant dynamics. In addition, a back pad was used to rotate the torso 10 degrees into a more upright position for the highest severity crash pulse. The intent of this pad was to increase the forward component of the acceleration vector such that measured upper body accelerations would more closely match those of an individual seated in an actual stroking seat. The result was a much closer correlation of the rigid seat simulation to the stroking seat model for the high severity tests.

Table 2: Manikin Head Fixture Mass Properties with 4.0 lbs. Added Head Mass

Manikin Size	Total Head Mass (lbs)	CG Distance from OC (inch)	(angle)	MOIxx ($lbs\text{-}in^2$)	MOIyy ($lbs\text{-}in^2$)	MOIzz ($lbs\text{-}in^2$)
Small Female	12.11	2.55	70.1°	91.96	80.33	89.93
Mid-Male	13.55	2.42	68.6°	111.77	103.81	103.82
Large Male	14.72	2.39	72.2°	141.53	111.82	107.20

* Note: Mass properties include head and added head mass.

Horizontal Accelerator Testing

Standard 30-degree pitch-down helicopter crash pulses were examined. The pulses used were representative of standard seat qualification crash corridors for a variety of rotary wing platforms. A crash-repeatable rigid seat with a fixed generic restraint was used in all of the tests. The input pulse supplied to the rigid seat was representative of the acceleration observed by the seat pan of an energy-absorbing seat structure with a 30-degree pitch-down for the rotary wing platforms specified in Table 3. The only exception to this was the third crash pulse configuration. This pulse was representative of a lower level impact that was expected to be survivable for all occupant sizes and added head masses. Yaw and roll conditions were not examined as part of this effort for several reasons: 1) an analysis of mishap data demonstrated that nearly 60% of all impacts have no roll component, and 80% of all impacts have no yaw component, 2) introducing these variables into the testing increases the test complexity, and 3) neck injury criteria have only been validated and accepted for the sagittal plane. By adding yaw and roll components into the tests, the measured forces and moments in the sagittal plane are decreased, consequently jeopardizing the validity of the results.

Table 3: Proposed Crash Pulses

	Structural Airframe Crash Pulse		Crash Pulse as Experienced by Seat Pan of Stroking Seat (derived from Computer Simulation)	
Rotary Wing Platform	**Crash Pulse (Peak G)**	**Crash Pulse ΔV (ft/sec)**	**Crash Pulse (Peak G)**	**Crash Pulse ΔV (ft/sec)**

SH-60, UH-1, V-22, H-53	45-50	45-50	19.0	45-50
AH-1Z	38	31.5	18.5	31.5
All	25-28	25	16.25	25

A total of 43 successful tests were conducted as part of this study. Table 4 describes the tested added head mass conditions for each crash pulse simulated during this test program. These conditions are also graphically described by Figures 3, 4 and 5. Calibration of the HYGE acceleration pulse signature was performed prior to conducting the dynamic tests. Peak accelerations and velocities were maintained plus or minus ten percent.

Table 4: Test Matrix for Horizontal Sled Tests

All tests were 30-Degree Pitch-Down

Test #	Manikin (%ile)	Test Pulse Configuration[1]	Added Head Wt & CG Location	Theta (deg)
Low Severity Pulse				
514	5th Female	Max G = 16.25 V (Ft/s) = 25.0	W = 4.0 lbs R = 3.0 inches	70.1
515	5th Female	Max G = 16.25 V (Ft/s) = 25.0	W = 5.0 lbs R = 4.0 inches	67.5
516	5th Female	Max G = 16.25 V (Ft/s) = 25.0	W = 5.5 lbs R = 4.5 inches	69.5
517	5th Female	Max G = 16.25 V (Ft/s) = 25.0	W = 6.0 lbs R = 5.0 inches	66.9
530	50th Male	Max G = 16.25 V (Ft/s) = 25.0	W = 4.0 lbs R = 3.0 inches	67.7
531	50th Male	Max G = 16.25 V (Ft/s) = 25.0	W = 6.0 lbs R = 5.0 inches	67.9
512	95th Male	Max G = 16.25 V (Ft/s) = 25.0	W = 4.0 lbs R = 3.0 inches	70.6
513	95th Male	Max G = 16.25 V (Ft/s) = 25.0	W = 6.0 lbs R = 5.0 inches	70.7
Medium Severity Pulse				
577	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 3.5 lbs R = 2.5 inches	67.5
578	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 4.0 lbs R = 3.0 inches	70.1
579	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 4.5 lbs R = 3.0 inches	68.7
518	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 4.0 lbs R = 3.0 inches	70.1
519	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 5.0 lbs R = 4.0 inches	67.5
520	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 5.0 lbs R = 5.0 inches	72.2
521	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 5.5 lbs R = 4.0 inches	69.5
522	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 5.5 lbs R = 5.0 inches	69.6
523	5th Female	Max G = 18.5 V (Ft/s) = 31.5	W = 6.0 lbs R = 5.0 inches	66.9
524	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 4.0 lbs R = 3.0 inches	67.7
525	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 4.5 lbs R = 4.0 inches	67.0
526	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 5.0 lbs R = 4.0 inches	72.2
527	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 5.0 lbs R = 5.0 inches	66.9
529	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 6.0 lbs R = 5.0 inches	67.9
590	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 4.0 lbs R = 3.0 inches	67.7
591	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 5.0 lbs R = 4.0 inches	72.2
592	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 5.5 lbs R = 5.0 inches	69.8
593	50th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 6.0 lbs R = 5.0 inches	67.9
509	95th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 4.0 lbs R = 3.0 inches	70.6
510	95th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 5.5 lbs R = 4.0 inches	70.8
511	95th Male	Max G = 18.5 V (Ft/s) = 31.5	W = 6.0 lbs R = 5.0 inches	70.7
High Severity Pulse				
537	5th Female	Max G = 19.0 V (Ft/s) = 50.0	W = 3.5 lbs R = 2.5 inches	67.5
538	5th Female	Max G = 19.0 V (Ft/s) = 50.0	W = 4.0 lbs R = 3.0 inches	70.1
539	5th Female	Max G = 19.0 V (Ft/s) = 50.0	W = 2.8 lbs R = 2.25 inches	77.5
540	5th Female	Max G = 19.0 V (Ft/s) = 50.0	W = 2.8 lbs R = 2.25 inches	77.5
586	50th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 2.8 lbs R = 2.25 inches	70.6
587	50th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 3.5 lbs R = 2.5 inches	66.7
588	50th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 4.0 lbs R = 3.0 inches	67.7
589	50th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 4.5 lbs R = 4.0 inches	67.0
580	95th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 2.8 lbs R = 2.25 inches	71.4
581	95th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 3.5 lbs R = 2.5 inches	72.9
582	95th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 4.0 lbs R = 3.0 inches	70.6
583	95th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 5.0 lbs R = 4.0 inches	72.9
584	95th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 5.5 lbs R = 4.0 inches	70.8
585	95th Male	Max G = 19.0 V (Ft/s) = 50.0	W = 6.0 lbs R = 5.0 inches	70.7

* NOTE: The CG location from the OC includes the added mass only.

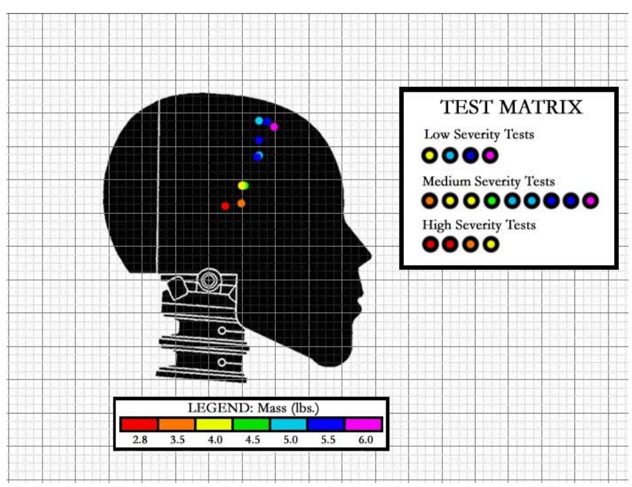

Figure 3: Small Female Added Head Masses Tested

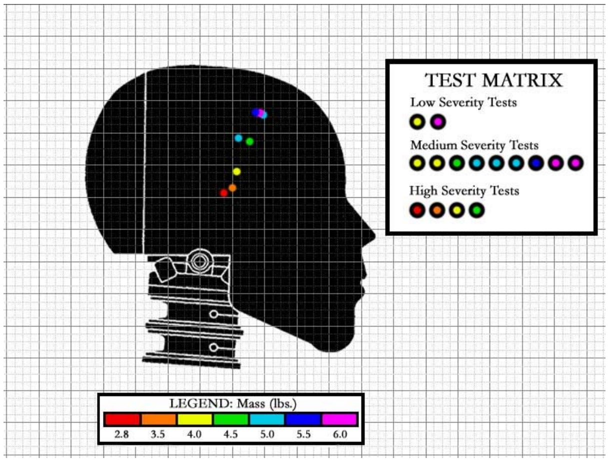

Figure 4: Mid-Male Added Head Masses Tested

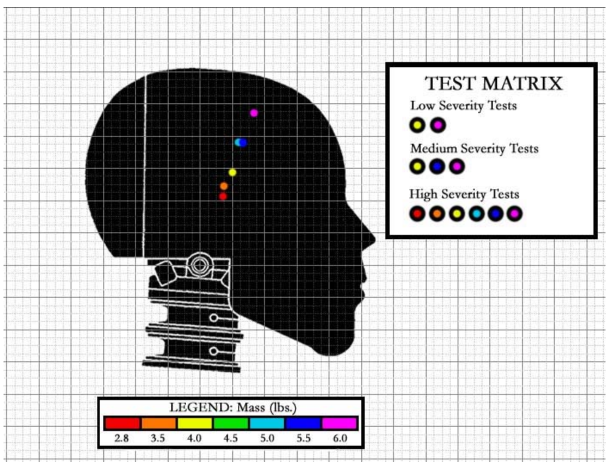

Figure 5: Large Male Added Head Masses Tested

Computer simulations determined that the magnitude of importance of the variables used to define the crash pulse were ΔV (ft/sec), Peak Acceleration (G), G-Onset Rate (G/sec), and Pulse Duration (Δt), respectively. The Horizontal Accelerator facility was successfully able to accommodate the first two variables of the pulse while needing to sacrifice the last two. The consequence of this trade-off resulted in pulses that were typically longer in duration with lower onset rates than the simulation study proposed. This was deemed acceptable to meeting the requirements of the program. Examples of the sled pulses are shown in Figure 6.

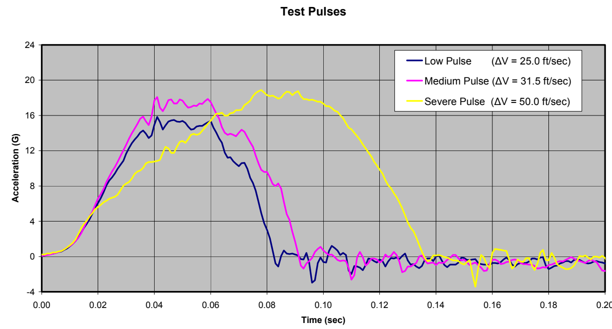

Figure 6: Pulses used for Horizontal Accelerator testing

Horizontal Accelerator Facility

The Horizontal Accelerator (HA) Facility simulates typical decelerative crash forces associated with aircraft mishaps by reversing the orientation of the test article and subjecting the test article to accelerative forces from an initial velocity of zero. The Horizontal Accelerator consists of three main assemblies: 1) the accelerating mechanism, 2) the test sled, and 3) the guide rails. The accelerating mechanism is a twelve-inch HYGE actuator that consists of a stainless-steel cylinder, divided into two twelve-foot long chambers. The energy required to produce the impact acceleration is generated within the actuator cylinder by means of differential gas pressures acting up on a thrust piston. The rear chamber contains compressed air used as the firing pressure. The front chamber is filled with pressurized nitrogen, used to apply braking to the thrust assembly.

Upon actuation, air is introduced into the front chamber, accelerating the thrust assembly forward. A metering pin, located between the two chambers, controls the acceleration-time profile applied to the sled. A maximum force of 225,000 pounds of gross thrust can be generated. This force is reacted by a reinforced concrete block weighing 75 tons. The result is a smooth transition of energy from the cylinder to the test sled. The length of the power stroke is variable, controlled by the volume of hydraulic fluid within the front chamber.

The test sled is twelve feet long and four feet wide. It is attached to guide rails that allow the sled to move away from the accelerator with minimum friction. After the accelerating stroke is completed, caliper brakes mounted on the sled are automatically activated to grip the rails and decelerate the sled to a smooth stop. The total length of the rails is 100 feet. Key characteristics of the Horizontal Accelerator Facility are:

Maximum Acceleration:	50 Gs
Maximum Velocity:	100 feet/sec
Maximum Payload:	5,000 lbs. at 20 Gs

Power Stroke: 10 feet
Pulse Shape: Variable

Rigid Crashworthy Seat System

The rigid crashworthy seat is an in-house fabricated test fixture that represents the seated position of a rotary wing aviator, while allowing many uses of the seat structure. The system is comprised of an aluminum frame seat pan and seat back and was mounted at a 30-degree angle in the sagittal plane for all tests. Attachment points for the restraint system are representative of an actual crashworthy seat. A generic five-point restraint system was used on all the tests. Integrated into the restraint system yoke, in place of a standard H. Koch and Sons MA-16 inertia reel, was a 3,000-lb Denton 1914 load cell to measure belt loading during the impact (Figure 7). The intent of the load cell was to attempt to pre-tension the restraint system consistently for each test. Pre-test pre-load of the restraint system ranged from 36 to 44 lbs. throughout the entire test series. The restraint system was replaced frequently during the testing. One-inch thick rate dependent foam was used as a seat cushion. A back pad was used to rotate the torso 10 degrees into a more upright position for the highest severity crash pulse. The intent of this pad was to increase the forward component of the acceleration vector such that measured upper body accelerations would more closely match those of an individual seated in an actual stroking seat. The effect of this condition was deemed insignificant for the low and medium severity tests and was validated from simulation analysis.

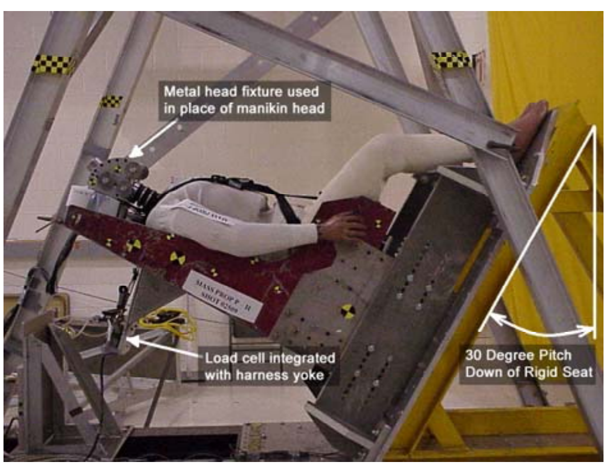

Figure 7: Detailed Test Setup

Anthropomorphic Test Dummy (ATD)

Three ATD's were used for this study: 1) a 95th %ile male Hybrid III (224 lbs.), 2) a 50th %ile male Hybrid III (184 lbs.), and 3) a 5th %ile female Hybrid III (118 lbs.). The ATD's were dressed only in thermal underwear. Pre-test positioning of the manikins complied with typical flight posture and was constant during all tests. An in-house designed and fabricated aluminum head fixture (Figure 8) was used in place of each ATD's normal head. The fixture was designed to mount directly to the Hybrid III neck and allowed weights to be bolted vertically (holes designated Z1-Z7), horizontally (holes designated X-1, X1-X5) and radially (holes designated F1-F9) from the head pivot pin as well as at the pivot pin (hole designated OC). Weights were added to the head fixture such that the added head weight varied between 2.8–6.0 lbs., and the CG (referenced from the head pivot pin) varied between 2.25–5.0 inches (not including the mass of the head). The angle, theta (θ), that the CG made with the head pivot pin ranged from 66.7°–77.5° for all configurations tested. It was desired to keep this angle between 65°–75° degrees, as this was felt to be representative of the CG of existing HMD systems. An Excel spreadsheet was used to mathematically determine CG location (radius and angle) for each configuration. The minimum and maximum mass properties data (2.8–6.0 lbs. added head mass) for each manikin type are listed in Table 6. The 6 lbs. configurations are shown in Figures 9, 10 and 11.

Figure 8: Metal head test fixture

Table 6: MEASURED Min/Max Manikin Head Fixture Mass Properties with Added Head Mass

Manikin Size	Added Head Mass	CG Distance from OC		MOIxx	MOIyy	MOIzz
	(lbs)	(inch)	(angle)	(lbs-in^2)	(lbs-in^2)	(lbs-in^2)
Small Female	2.79	3.71	79.4º	35.28	8.49	42.41
Small Female	5.92	5.67	68.4º	32.32	23.34	93.78
Mid-Male	2.87	3.18	53.2º	50.83	23.67	29.03
Mid-Male	6.00	5.66	65.1º	20.2	51.99	47.51
Large Male	2.89	2.85	47.6º	56.19	2.15	43.35
Large Male	6.01	5.44	66.9º	84.66	40.65	62.73

* Note: Mass properties ONLY include added head mass.
* Note: Mass properties in this table were measured, versus calculated.

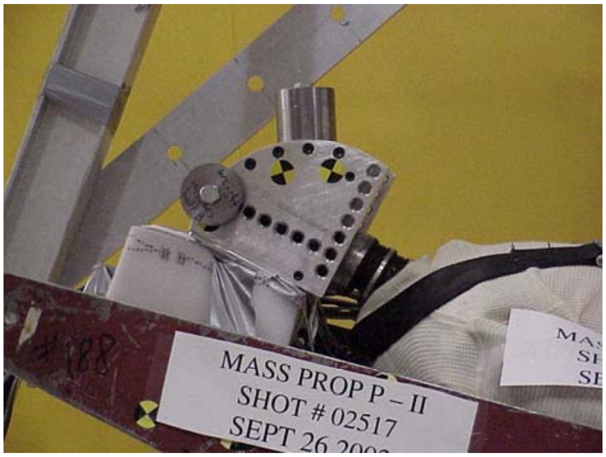

Figure F-9: Small Female with 6 lbs. added head mass

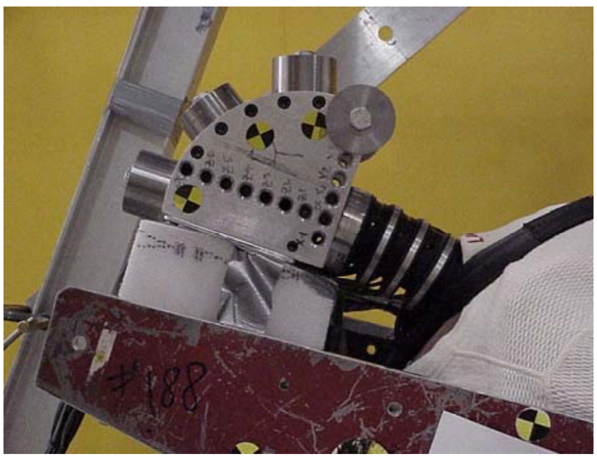

Figure F-10: Mid-size Male with 6 lbs. added head mass

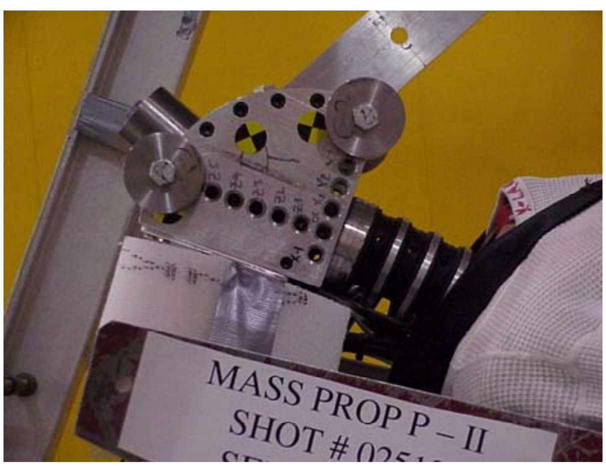

Figure F-11: Large Male with 6 lbs. added head mass

The instrumentation incorporated into the horizontal accelerator test sled and Hybrid III manikin included linear and angular accelerometers, and force/moment load cells at the top and bottom of the neck, thorax, and pelvis.

All data was collected at 1000 Hz. Data was filtered at 100 Hz using an 8-pole, zero-shift Butterworth filter. The upper neck load cell used in all three manikins was the Denton model J-1716. In order to apply the cervical injury criteria it was necessary to translate the measured y-moment values from the load cell to the head pivot pin. This location on the manikin approximately mimics the occipital condyles in a human. The equations used to accomplish this transformation are shown in Equation Set 1. In addition, a similar procedure was followed for the lower neck load cell to translate the forces and moments to the base of the neck. The lower neck load cell used in the manikins for the Low and Medium severity tests was the Denton model J-1794. However, it was determined that the y-moment capability was being exceeded in the High severity tests. Consequently, this load cell was replaced with the Denton model J-5832 which featured a more robust range for recording the y-moment. The equations used to transform the forces and moments in the lower neck load cells are shown in Equation Set 2 for the J-1794 and Equation Set 3 for the J-5832.

+Fx :	Head/Neck rearward, Chest forward
+Fy :	Head/Neck leftward, Chest rightward
+Fz :	Head/Neck upward, Chest downward
+Mx:	Left Ear toward Left Shoulder
+My:	Chin toward Sternum
+Mz:	Chin toward Left Shoulder

$$Mocx = Mx + Fy * OD$$
$$Mocy = My - Fx * OD$$

Where OD = 0.700 inches (for all manikin sizes)

Equation Set 1

$$Fx = Fxm$$
$$Fy = Fym$$
$$Fz = Fzm$$
$$Mx = Mxm - Fym*Dz$$
$$My = Mym + Fxm*Dz + Fzm*Dx$$
$$Mz = Mzm - Fym*Dx$$

Where Dx = 1.75” (small female) and 2.00” (mid- and large male), and Dz = 1.125” (all manikin sizes tested).

Equation Set 2

$$Fx = Fxm * \cos\Phi + Fzm * \sin\Phi$$
$$Fy = Fym$$
$$Fz = Fzm * \cos\Phi - Fxm * \sin\Phi$$
$$Mx = Mxm * \cos\Phi + Mzm * \sin\Phi - Fym * (1.72'') * \cos\Phi - Fym * (2.5'') * \sin\Phi$$
$$My = Mym + Fxm * (1.72'') + Fzm * (2.5'')$$
$$Mz = Mzm * \cos\Phi - Mxm * \sin\Phi - Fym * (2.5'') * \cos\Phi + Fym * (1.72'') * \sin\Phi$$

Where $\Phi = 11.95^{o}$ and represents the angle setting on the lower neck.

Equation Set 3

* NOTE: Fxm, Fym, Fzm, Mxm, Mym, Mzm are the forces and moments measured at the lower neck load cell and Fx, Fy, Fz, Mx, My, and Mz are the forces and moments at the center of the base of the neck.

Cervical Injury Analysis

A comprehensive analysis of the data was performed using available injury predictors to determine the likelihood of injury to the upper and lower cervical spine. These criteria have been developed and established by the automotive and military environments over the past four decades [10-16]. These criteria reflect the most current knowledge and understanding of cervical injury limits and have been jointly agreed upon by the USN and USAF for use in such programs as the Joint Ejection Seat Program (JESP) [17]. While these are non-crashworthy programs, the injury criteria are still valid for use in this study. Critical moment values were specified for flexion and extension about the y-axis and force duration curves were generated for axial tension, compression, lateral bending and shear (as described in detail in Appendix B). It is important to point out that only specific crash pulses and orientations were modeled and tested. Testing was conducted only in the X-Z plane, even though it was understood that helicopters may impact at any orientation, because this was expected to maximize the forces (Fx and Fz) and moments (My) recorded by the manikin in the sagittal plane. Cervical injury criteria have not been adequately developed for lateral and rotational moments about the neck (Mx and Mz).

In addition, it was assumed through the course of this effort that the aviator would be properly restrained in the seat during the crash event. Out of position occupants were not modeled as a course of this effort (i.e. occupant looking back over their shoulder at the time of impact). Occupants who have their heads turned at the time of impact will have much greater risk of cervical injury. The result of this study is a probability of cervical injury, based on aviator size, helmet mass and impact severity, to be used as design criteria for helmets and helmet-mounted systems for rotary wing aircraft. The results of this analysis are subject to change following any changes or modifications to the dynamic properties of the seat (i.e. improved crashworthiness) or restraint system (i.e. neck load mitigation devices), interior cockpit structures or configuration, structural crash properties of the helicopter, aviator population, or development of improved neck injury criteria. In addition, the results of this analysis are to be used solely as a baseline for helmet and helmet-mounted system development. Final system configuration may have unforeseen effects that could not be represented in the study. Therefore, systems level testing would still be required with any new helmet or helmet-mounted system.

RESULTS

Computer Model Validation

Rigid Crashworthy Seat vs. Actual Stroking Seat

The rigid crashworthy seat was used during the course of the testing. However, the energy transmitted to the occupant through a rigid seat was drastically different from the energy transmitted to the occupant through a stroking crashworthy seat. It was intended that the tests mimic the head/neck response of an occupant in a stroking seat as closely as possible. Therefore, it was determined that modeling would be used to determine what the correct pulse applied to a rigid seat would need to be to 'simulate' the acceleration profile exhibited by a stroking seat. The MADYMO simulation code was set up for both the stroking seat and fixed seat cases. The stroking seat simulation represented the performance of a crashworthy seat subjected to an actual crash condition, and the fixed seat model represented the HA test condition.

The structural dynamic pulse requirements were run through the stroking seat simulation and the resulting seat accelerations were used as the input accelerations for the rigid seat model. Ultimately, the Y-axis neck moments, both upper and lower, were the target variables of interest, but the effort to achieve equivalence began with the pelvic acceleration. Since the interaction between the seat and the pelvis drove the rest of the body's response, it was felt that a reasonable match to the pelvic acceleration was the key to overall agreement. In the first iteration, the X-axis acceleration of the stroking seat was used as the driving acceleration of the fixed seat. This produced the pelvic acceleration shown in Fig. 12. The agreement between the fixed and stroking models appeared reasonable, but the fixed model over-predicted the pelvic acceleration initially and under-predicted it later on. This is believed to have been due to the fact that the seat does not stroke horizontally.

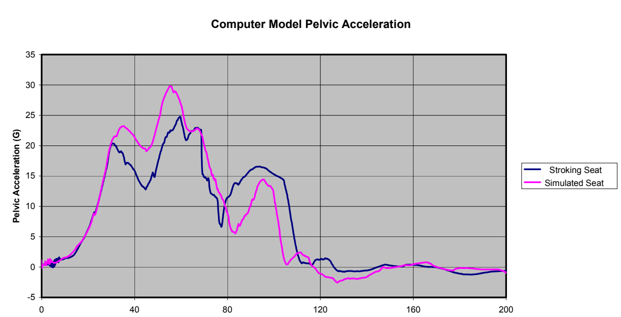

Figure 12: Computer Model Validation (Stroking Seat vs. Rigid Seat)

The seat was rotated 30° from horizontal to mimic the 30° pitch-down test condition used in qualifying rotary wing crashworthy seats. Thus, as the seat stroked, it moved down as well as forward, as shown in Fig. 13. This led to a vertical component of acceleration, A_Z, which in turn had a component normal to the seat bottom, $A_Z \cdot \sin(30^o)$. This component reduced the pelvic acceleration early in the event, but caused an increase at the end. A correction factor, shown at the bottom right of Fig. 13, incorporated into the fixed seat pulse, provided a better overall match to the pelvic acceleration, as shown in Fig. 14.

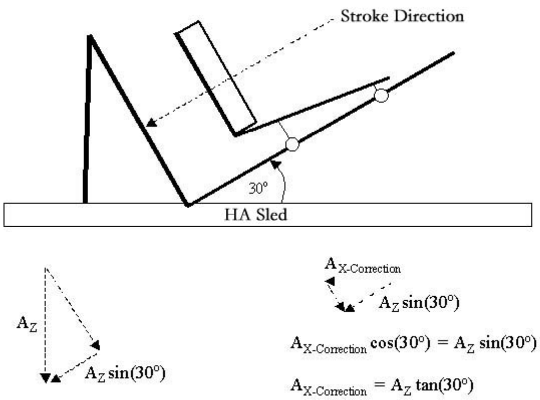

Figure 13: Derivation for Rigid Seat X-Accel. Correction Factor

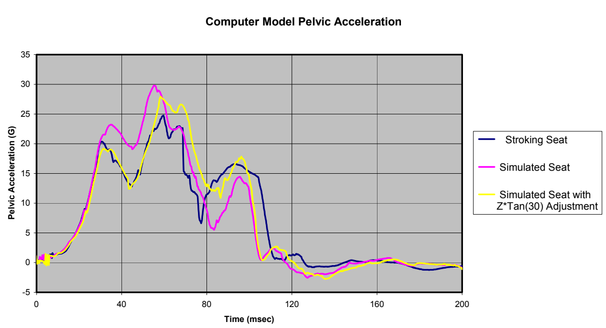

Figure 14: Computer Model Validation (Stroking Seat vs. Rigid Seat with Correction Factor)

The stroking seat acceleration was a complex pulse that the Horizontal Accelerator facility was not able to duplicate. Therefore, simulations using approximations of this pulse (with the same total energy) were compared against the results of simulations using the actual pulse to determine the acceptability of altering the pulse profile.

Three stroking seat equivalent pulses were generated that represented low, medium, and high severity impact conditions. Good correlation was obtained for the low and medium severity pulses. It was determined that a 10-degree back wedge was required on the high severity tests to obtain the necessary correlation between models of the principal measurement parameters. Results from the computer simulations using the simplified pulse were compared with results from the test data set under identical conditions.

Trapezoidal Pulse vs. "Actual" Pulse

The equivalent pulses were integrated to obtain the Delta-V (ΔV) for each. Because the derived, equivalent pulses were too complex to be reproduced on the HA, the ΔV's were used to generate trapezoidal pulses with the same amount of energy. These, in turn, were "rounded" to produce target pulses that could be closely approximated on the HA (Figure 15). Runs with the original equivalent pulses, the trapezoidal pulses, and the rounded pulses showed little change where the critical neck moment parameters were concerned. This was probably because the peak values for the neck parameters occurred after the pulse was over. Delivering the right amount of energy, over approximately the correct time period appeared to be sufficient to obtain reasonable neck results. The rounded pulses were then used to approximate the centerline of the pulse corridor for the actual HA pulses.

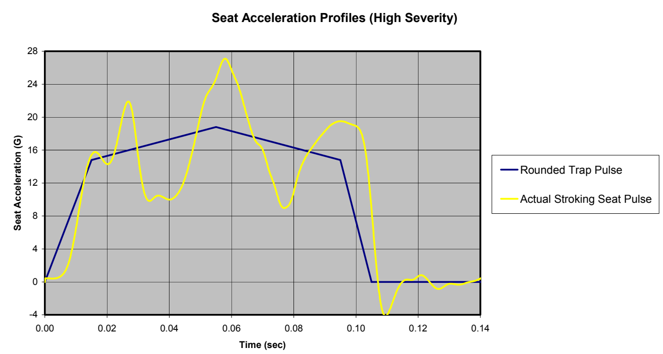

Figure 15: Actual vs. Approximated High Severity Input Profile

10-Degree Back Pad Validation

Subsequent comparison of the neck moments for the low and medium severity pulses showed good agreement between the fixed and stroking seat simulations (Figures 16 and 17). The high severity case, however, demonstrated large variations between the two models. This discrepancy was due mainly to the extended time-span and seat-stroke distance in these runs. This allowed the body to reposition itself with a significant forward lean, prior to the time of maximum head motion. To compensate for this with the fixed seat, an initial, forward angle of 10° was added to the torso of the manikins in the high-energy simulations. This produced much better neck moment agreement, as shown in Figure 18. Additional angles were investigated, but 10-degrees appeared to provide the optimal fit of the rigid seat approximation of the stroking seat model.

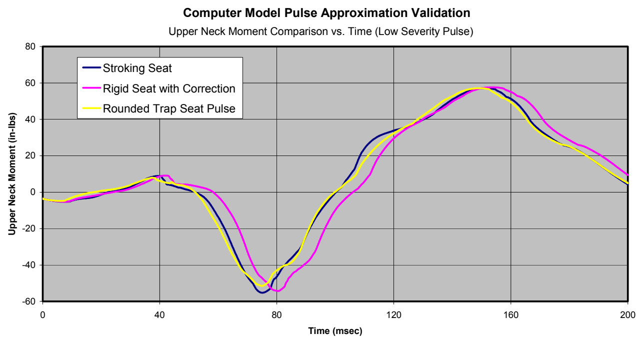

Figure 16: Computer Model Pulse Approximation Validation (Low Severity Pulse)

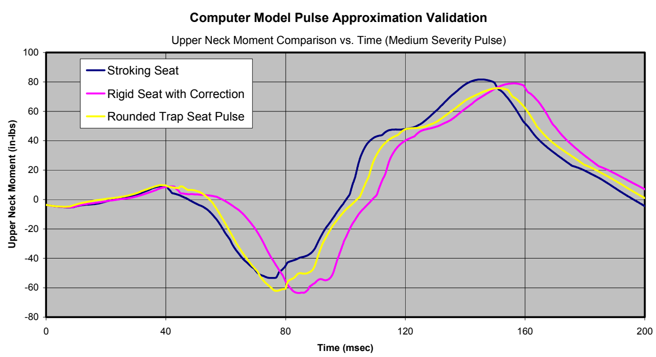

Figure 17: Computer Model Pulse Approximation Validation (Medium Severity Pulse)

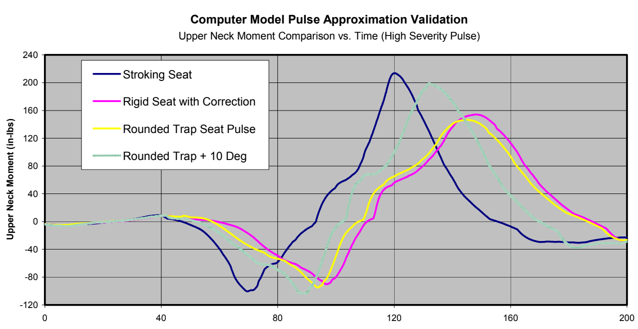

Figure 18: Computer Model Pulse Approximation Validation (High Severity Pulse)

<u>Comparison with 4.0 lbs Added Head Weight Test Data</u>

The computer model predicted results of the 4.0 lbs. added head weight were compared with the test data using the same configuration. Note that the computer model used for comparison used the 'rounded' trapezoid pulse (with the 10-degree back pad for the high severity cases). In general, the trends predicted by the model agreed very well with the test data results. However, the model tended to under-predict the peak values for most of the neck parameters in the low and medium severity cases for all the manikin sizes (Figures 19 and 20). In the high severity cases, the peak values usually occurred earlier in the model than in the sled tests for all manikin sizes (Figure 21).

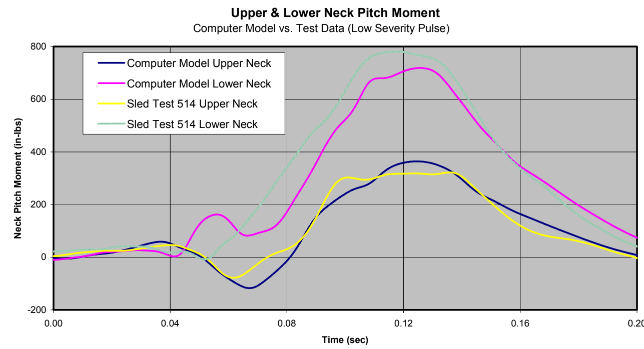

Figure 19: Upper & Lower Neck Moments
Computer Model vs. Test Data (Small Female – Low Severity Pulse)

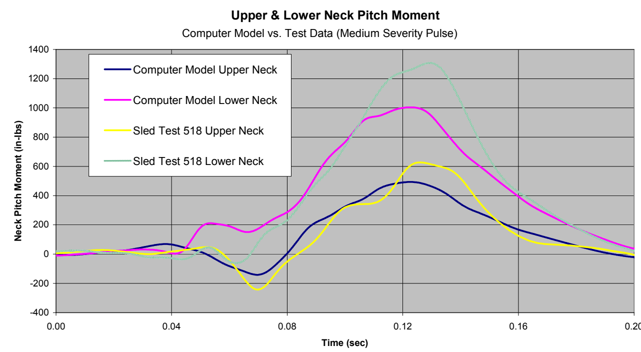

Figure 20: Upper & Lower Neck Moments
Computer Model vs. Test Data (Small Female – Medium Severity Pulse)

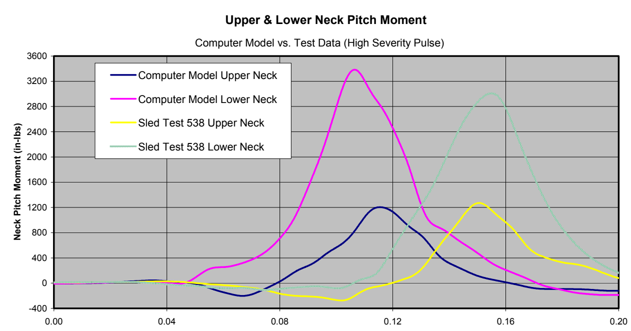

Figure 21: Upper & Lower Neck Moments
Computer Model vs. Test Data (Small Female – High Severity Pulse)

Horizontal Accelerator Tests

The NHTSA validated Head Nij criterion (Appendix C) was used to evaluate the potential for neck injury due to combined loading effects. Critical values for Tension, Compression, Flexion Moment, and Extension Moment were scaled to represent approximately a 5% probability of injury for the different anthropometric sizes tested (Nij = 0.5). The Nij criterion was evaluated at the base of the neck. Values above 1.5 may be representative of injury to the lower neck. However, an injury criteria at this location of the neck has never been validated and only subjective assessments may be made.

Pulse severity was undeniably the most dominant variable examined during the study. This was evident by the relatively small slope in all of the peak value plots when added head mass was increased. The effect of aviator size was more related to the different values for the neck injury limits which were scaled according to weight. Therefore, despite the fact that the cervical forces and moments were lowest in magnitude in the small female tests, because the injury criteria limits were also reduced, the benefits were negated.

Plots of the cervical injury analysis resulting from the tests are given in Appendix A (Figures A-1 – A-9). The peak values of the recorded data are listed in Table A-1. Trends with added head mass were more noticeable during the lower severity pulses, when the total energy experienced by the occupant was not so severe. This was evident by the amount of scatter present in the data for the higher severity pulses. A summary of the results of the neck injury analysis is shown in Table 6. This table may be used as a guideline when determining the risk to the aviator for a particular helmet system. All of the aviator sizes tested passed the low severity tests for all of the added head weights tested (up to six pounds). For the medium severity tests, only the mid-male and large male passed the neck injury criteria for all of the added head weights tested (up to six pounds). The small female reached the limit at approximately four pounds. For the high severity tests, the small female and mid-male failed for all of the added head weights tested. The large male passed the neck injury criteria only for the lowest added head weight (2.8-3.0 pounds). It is important to point out that nearly 90% of all survivable Naval helicopter mishaps occur at an impact level that is equal to or less than the high severity pulse (50 ft/sec) that was used for these tests.

TABLE 6: Test Results Indicating Pass/Fail Status Regarding Neck Injury Criteria for all Tests

Test #	Pulse Severity	Added Head Weight (lbs)	Pass = green; Fail = red; Marginal = yellow Nij Criteria	Load Duration
Small Female				
514	Low	4.0 lbs	Pass	Pass
515	Low	5.0 lbs	Pass	Pass
516	Low	5.5 lbs	Pass	Pass
517	Low	6.0 lbs	Pass	Pass
577	Medium	3.5 lbs	Pass	Pass
518	Medium	4.0 lbs	Marginal	Pass
578	Medium	4.0 lbs	Fail	Pass
579	Medium	4.5 lbs	Fail	Pass
519	Medium	5.0 lbs	Fail	Marginal
520	Medium	5.0 lbs	Marginal	Pass
521	Medium	5.5 lbs	Fail	Marginal
522	Medium	5.5 lbs	Fail	Fail
523	Medium	6.0 lbs	Fail	Pass
539	High	2.8 lbs	Fail	Fail
540	High	2.8 lbs	Fail	Fail
537	High	3.5 lbs	Fail	Fail
538	High	4.0 lbs	Fail	Fail
Mid-Male				
530	Low	4.0 lbs	Pass	Pass
531	Low	6.0 lbs	Pass	Pass
524	Medium	4.0 lbs	Pass	Pass
525	Medium	4.5 lbs	Pass	Pass
526	Medium	5.0 lbs	Pass	Pass
527	Medium	5.0 lbs	Pass	Pass
529	Medium	6.0 lbs	Pass	Pass
590	Medium	4.0 lbs	Pass	Pass
591	Medium	5.0 lbs	Pass	Pass
592	Medium	5.5 lbs	Pass	Pass
593	Medium	6.0 lbs	Pass	Pass
586	High	2.8 lbs	Fail	Fail
587	High	3.5 lbs	Fail	Fail
588	High	4.0 lbs	Fail	Fail
589	High	4.5 lbs	Fail	Fail
Large Male				
512	Low	4.0 lbs	Pass	Pass
513	Low	6.0 lbs	Pass	Pass
509	Medium	4.0 lbs	Pass	Pass
510	Medium	5.5 lbs	Pass	Pass
511	Medium	6.0 lbs	Pass	Pass
580	High	2.8 lbs	Pass	Pass
581	High	3.5 lbs	Fail	Pass
582	High	4.0 lbs	Fail	Marginal
583	High	5.0 lbs	Fail	Pass
584	High	5.5 lbs	Fail	Fail
585	High	6.0 lbs	Fail	Pass

An analysis of variance (ANOVA) with a post-hoc Tukey-Kramer test to determine the source of the difference was conducted on the principal neck parameters to determine whether any of the factors were statistically significant (defined as $p \leq 0.05$). The results of the statistical analyses indicated that for each of the manikin sizes tested, added head mass did not have a significant effect on the magnitude of the forces or moments measured when contrasted with test variability. However, forces and moments were still observed to increase as added head mass was increased. There was a significant effect of the severity of the applied pulse for all parameters except lower neck tension. Since pulse severity was the dominant factor, as compared to weight or manikin size, ANOVA for the effects of these two factors were also conducted for each individual pulse type. Size was a significant factor (1) in the low severity tests for all parameters except the upper neck flexion moment, (2) in the medium severity tests for all parameters except the lower neck tension, and (3) in the high severity tests for all parameters except the upper neck tension. It is important to note that the results of these statistics were based upon a small sample size with very few repeated measures. Additional testing under identical test conditions would be required to improve the statistical confidence.

CONCLUSIONS & RECOMMENDATIONS

The crash pulses used in the course of this study, in conjunction with the rigid crashworthy seat, ranged from an expected survivable impact to a very severe impact. The 30-degree pitch down configuration was chosen because it was believed to represent a worst-case scenario in terms of loading vectors and kinematics. Injury criteria that were developed for axial loading were deemed not applicable due to the orientation of the applied loads. Modified injury criteria were used that were based upon historical aviation test and mishap data and approximated a 5% probability of injury. The results of this study regarding allowable added head mass are subject to change in the event that the injury criteria used to analyze the data are updated.

It is recommended that further testing be conducted to improve the statistical significance of each tested configuration. Additional testing would provide greater confidence in quantitatively establishing the injury threshold for all aviator sizes given changes to added head weight and impact condition. In addition, it is recommended that further testing examining the effects of roll, pitch and yaw on the observed cervical forces and moments be conducted and compared with current neck injury criteria. Finally, the mass properties guidelines may be revised based upon this improved dataset.

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BIOGRAPHIES

Glenn Paskoff is a Biomechanics Engineer for the Naval Air Warfare Center, Aircraft Division, Patuxent River, MD. Mr. Paskoff received his B.S. in Mechanical Engineering from Penn State in 1991 and his M.S. in Biomedical Engineering from the University of Virginia in 1995. He received a Research Assistantship at the Automobile Safety Laboratory in Virginia where he gained extensive experience in injury analysis, subject preparation and testing protocols. Since beginning his career with the Navy, Mr. Paskoff has been involved in all aspects of escape systems research, development, design, testing, computer simulation, analysis and evaluation, management, and support related to the protection of aviators from injury resulting from high-speed ejection and crash.

Dr. Sieveka holds a B.S. in Physics and Computer Science from the College of William and Mary, and an M.S. in Physics from Louisiana State University. In 1983, he earned a Ph.D. in Engineering Physics from the University of Virginia. He remained at UVA as a research scientist in the Mechanical Engineering Dept., and became heavily involved in the computer modeling of human body crash dynamics related to automobile mishaps, frequently working in conjunction with the University's Automobile Safety Laboratory. In 2001, he moved to Maryland and joined the Crashworthy Escape Systems team at the Patuxent River Naval Air Station. In this new position, Dr. Sieveka guides a growing program that is using occupant dynamics modeling to enhance aircrew safety for a wide variety of Naval aviation mishap scenarios involving both fixed-wing and rotary-wing aircraft.

APPENDIX A: TEST RESULTS

Test Conditions (Manikin - Pulse)	Test #	Add Head Wt (lbs)	CG Rad (in)	Delta V (ft/s)	Head (G)	Thorax (G)	Head +Fx (lbs)	Head +Fz (lbs)	Head -Fz (lbs)	Head +My (in-lbs)	Head -My (in-lbs)	Neck +Fx (lbs)	Neck +Fz (lbs)	Neck -Fz (lbs)	Neck +My (in-lbs)	Neck -My (in-lbs)	Head Nij	Neck Nij
Small Female - Low Pulse	514	4.0	3.0	25.08	18.7	21.1	90.5	139.7	-209.4	322.4	-106.5	113.9	103.9	-292.4	781.4	-150.1	0.29	0.66
Small Female - Low Pulse	515	5.0	4.0	25.30	23.3	26.0	96.0	167.4	-304.6	493.0	-189.9	108.6	141.2	-402.9	902.8	-161.2	0.39	0.77
Small Female - Low Pulse	516	5.5	4.5	24.94	26.3	28.0	94.1	179.0	-360.4	543.3	-236.2	117.9	165.0	-465.6	930.0	-161.8	0.49	0.80
Small Female - Low Pulse	517	6.0	5.0	24.98	26.7	29.7	99.6	177.6	-362.0	545.4	-217.4	128.5	160.7	-473.7	920.8	-180.8	0.48	0.78
Small Female - Medium Pulse	577	3.5	2.5	32.41	25.3	23.1	118.1	154.5	-232.7	377.3	-162.2	102.5	160.4	-309.7	921.9	-122.6	0.36	0.82
Small Female - Medium Pulse	518	4.0	3.0	31.60	34.5	36.2	153.3	239.6	-422.6	627.6	-242.7	155.0	202.2	-556.6	1307.5	-192.3	0.52	1.16
Small Female - Medium Pulse	578	4.0	3.0	31.99	28.3	29.0	136.7	249.6	-327.8	511.7	-305.2	138.6	288.1	-429.4	1116.7	-111.6	0.63	1.00
Small Female - Medium Pulse	579	4.5	4.0	32.34	30.7	31.2	129.1	269.3	-363.7	578.4	-386.2	165.2	304.3	-473.0	1109.7	-80.7	0.77	0.97
Small Female - Medium Pulse	519	5.0	4.0	31.48	35.3	37.4	129.3	192.1	-483.0	630.7	-267.8	140.4	177.5	-611.5	1100.8	-174.5	0.61	0.95
Small Female - Medium Pulse	520	5.0	5.0	31.70	34.9	36.9	108.8	213.8	-445.8	694.2	-243.5	140.2	182.9	-581.9	1186.1	-171.8	0.53	1.03
Small Female - Medium Pulse	521	5.5	4.0	31.71	35.8	37.9	134.8	234.7	-489.4	754.9	-263.0	168.1	188.4	-630.2	1354.9	-185.4	0.61	1.21
Small Female - Medium Pulse	522	5.5	5.0	31.64	41.7	42.6	134.8	229.9	-548.6	813.5	-264.4	166.8	194.0	-695.3	1380.2	-187.4	0.69	1.22
Small Female - Medium Pulse	523	6.0	5.0	31.62	33.6	37.7	140.3	238.3	-452.6	822.3	-241.6	195.9	179.7	-563.9	1436.9	-202.1	0.65	1.27
Small Female - Severe Pulse	537	3.5	2.5	50.28	43.9	26.2	326.6	576.7	-139.1	1310.4	-238.6	402.1	589.6	-352.9	2774.7	-274.1	1.02	2.40
Small Female - Severe Pulse	538	4.0	3.0	50.05	56.6	27.6	373.1	788.6	-139.1	1270.9	-276.3	475.7	812.3	-408.2	3009.0	-355.3	1.04	2.66
Small Female - Severe Pulse	539	2.8	2.3	50.34	56.6	31.7	352.9	706.2	-143.8	1202.6	-256.2	437.0	714.3	-378.0	2729.8	-530.3	0.96	2.41
Small Female - Severe Pulse	540	2.8	2.3	50.05	61.5	33.7	395.4	764.6	-155.5	1213.8	-299.6	408.8	761.2	-404.9	3020.0	-580.3	1.02	2.66
Mid-Male - Low Pulse	530	4.0	3.0	25.23	26.8	27.7	123.1	191.8	-356.8	484.9	-292.7	159.4	170.8	-435.7	1156.7	-341.1	0.35	0.44
Mid-Male - Low Pulse	531	6.0	5.0	24.74	27.2	27.0	111.8	189.9	-420.3	579.3	-336.4	155.9	191.3	-488.7	1217.2	-312.6	0.39	0.48
Mid-Male - Medium Pulse	524	4.0	3.0	31.52	21.5	23.7	206.5	308.1	-265.5	798.4	-232.1	267.1	326.6	-357.5	2051.2	-316.3	0.33	0.81
Mid-Male - Medium Pulse	525	4.5	4.0	31.08	30.1	33.0	166.6	269.6	-424.9	741.1	-335.3	220.7	288.7	-550.2	1788.6	-320.4	0.41	0.70
Mid-Male - Medium Pulse	526	5.0	4.0	31.32	26.5	27.4	170.5	264.6	-377.7	740.2	-299.1	205.5	272.0	-470.3	1802.1	-249.9	0.42	0.68
Mid-Male - Medium Pulse	527	5.0	5.0	31.37	24.3	25.6	210.3	319.7	-335.2	1070.1	-275.8	224.6	348.6	-422.3	2343.3	-254.4	0.44	0.95
Mid-Male - Medium Pulse	528	5.5	5.0	31.36	26.6	28.2	138.3	239.8	-350.4	693.3	-326.2	176.3	266.5	-425.5	1347.5	-298.1	0.44	0.55
Mid-Male - Medium Pulse	529	6.0	5.0	31.09	30.5	31.2	163.0	257.9	-484.4	825.2	-402.1	195.5	288.7	-596.3	1793.5	-402.3	0.45	0.70
Mid-Male - Medium Pulse	590	4.0	3.0	32.01	19.3	23.8	175.9	230.3	-253.3	744.9	-192.4	285.0	84.8	-281.4	2007.4	-327.7	0.28	0.87
Mid-Male - Medium Pulse	591	5.0	4.0	31.67	20.3	22.2	172.2	225.7	-302.3	803.0	-286.3	262.9	66.6	-306.1	1971.4	-294.0	0.30	0.85
Mid-Male - Medium Pulse	592	5.5	5.0	31.55	21.0	23.9	150.1	241.4	-321.9	843.5	-326.5	268.2	85.2	-339.1	1891.7	-274.2	0.32	0.80
Mid-Male - Medium Pulse	593	6.0	5.0	31.54	23.9	25.1	150.6	233.5	-375.3	903.6	-252.5	286.2	72.8	-376.5	1874.6	-266.8	0.35	0.79
Mid-Male - Severe Pulse	586	2.8	2.3	51.06	62.4	31.5	472.8	697.3	-134.1	1586.7	-384.3	833.8	220.1	-565.3	5830.9	-1240.1	0.68	2.52
Mid-Male - Severe Pulse	587	3.5	2.5	51.16	65.4	31.9	425.1	804.9	-134.2	1671.2	-430.6	980.6	239.6	-501.0	5397.2	-1289.8	0.70	2.30
Mid-Male - Severe Pulse	588	4.0	3.0	51.16	53.9	28.9	479.6	758.6	-168.6	1923.0	-388.8	842.4	217.0	-522.4	6060.3	-958.1	0.84	2.58
Mid-Male - Severe Pulse	589	4.5	4.0	51.38	53.0	27.3	487.0	828.5	-188.0	2043.3	-295.0	973.3	232.3	-514.2	6370.9	-1149.8	0.84	2.69
Large Male - Low Pulse	512	4.0	3.0	25.09	18.3	20.0	142.1	279.8	-263.6	731.1	-390.5	220.7	263.2	-353.9	1465.4	-356.9	0.30	0.40
Large Male - Low Pulse	513	6.0	5.0	24.94	18.2	19.2	132.5	279.5	-303.3	707.0	-391.4	186.8	229.0	-386.4	1444.3	-268.0	0.29	0.42
Large Male - Medium Pulse	509	4.0	3.0	31.05	20.5	23.3	157.2	307.6	-268.4	673.9	-332.2	181.5	280.5	-372.8	1399.9	-318.2	0.29	0.41
Large Male - Medium Pulse	510	5.5	4.0	31.27	17.9	21.9	162.8	302.6	-286.5	646.9	-339.4	205.7	244.4	-379.5	1760.5	-425.3	0.26	0.49
Large Male - Medium Pulse	511	6.0	5.0	30.99	20.9	22.8	178.1	389.1	-347.8	793.7	-484.7	238.2	294.1	-453.3	1861.0	-373.4	0.38	0.55
Large Male - Severe Pulse	580	2.8	2.3	50.03	51.2	26.0	439.0	658.9	-195.5	1276.4	-577.6	369.6	179.4	-302.1	4927.8	-1050.0	0.46	1.52
Large Male - Severe Pulse	581	3.5	2.5	50.42	47.1	26.9	446.3	747.1	-183.9	2090.5	-363.7	444.3	152.4	-442.6	5981.7	-917.0	0.62	1.87
Large Male - Severe Pulse	582	4.0	3.0	50.32	60.1	29.3	516.9	893.1	-176.4	1943.3	-617.5	354.8	179.5	-352.8	6453.4	-1064.3	0.63	1.96
Large Male - Severe Pulse	583	5.0	4.0	50.39	51.9	28.5	516.4	858.1	-191.6	2263.9	-584.5	416.3	173.7	-479.7	6920.2	-1110.9	0.68	2.15
Large Male - Severe Pulse	584	5.5	4.0	50.19	56.4	30.3	541.6	981.8	-199.5	2510.4	-447.1	307.2	270.7	-406.0	7431.7	-1002.0	0.75	2.26
Large Male - Severe Pulse	585	6.0	5.0	50.44	47.0	25.4	472.2	939.9	-211.0	2485.0	-536.1	420.1	175.9	-421.3	6702.9	-1005.4	0.72	2.05

Table A-1: Max Values of Recorded Data for all Tests

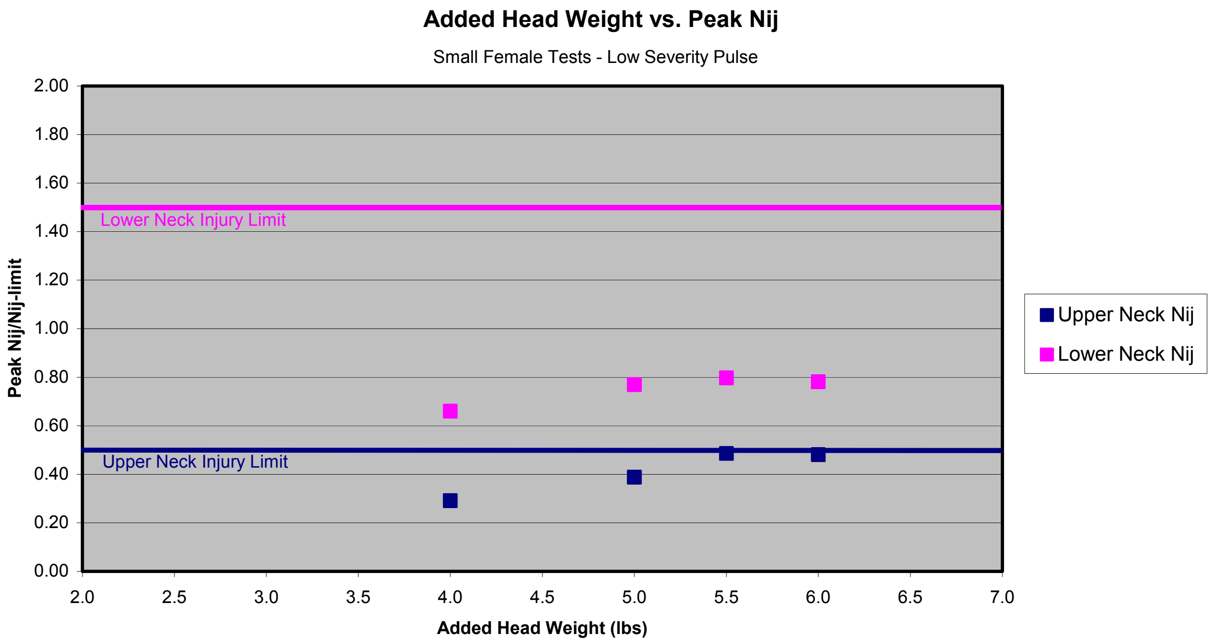

Figure A-1: Peak Nij Values (Small Female, Low Severity Pulse)

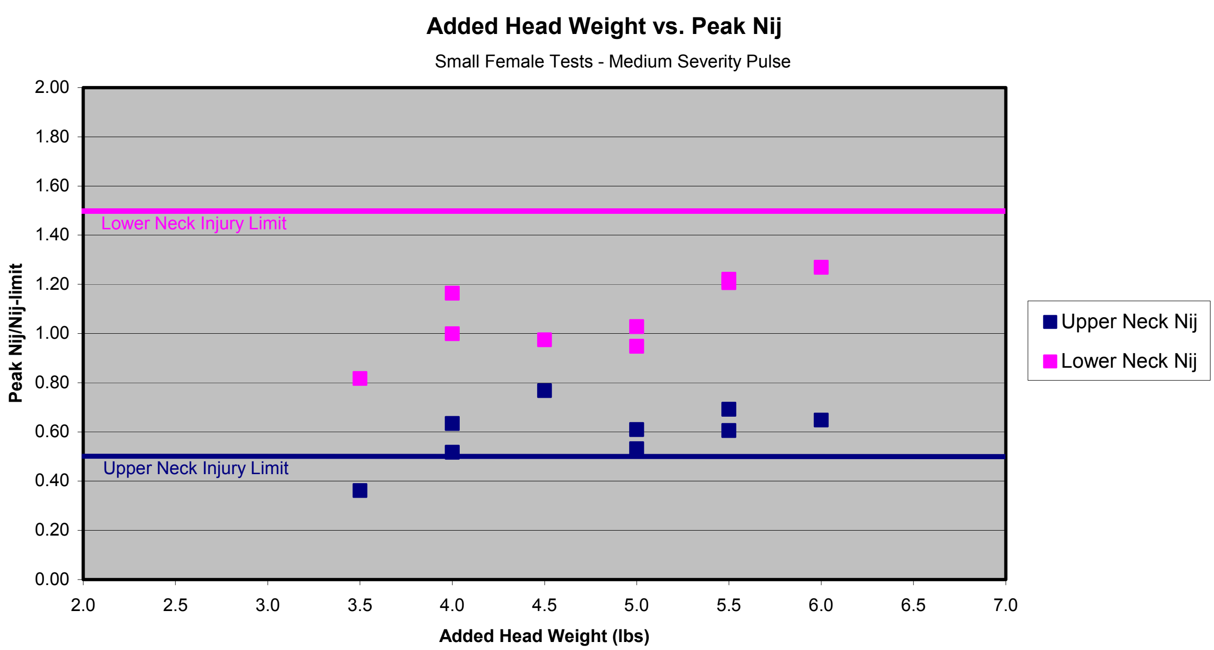

Figure A-2: Peak Nij Values (Small Female, Medium Severity Pulse)

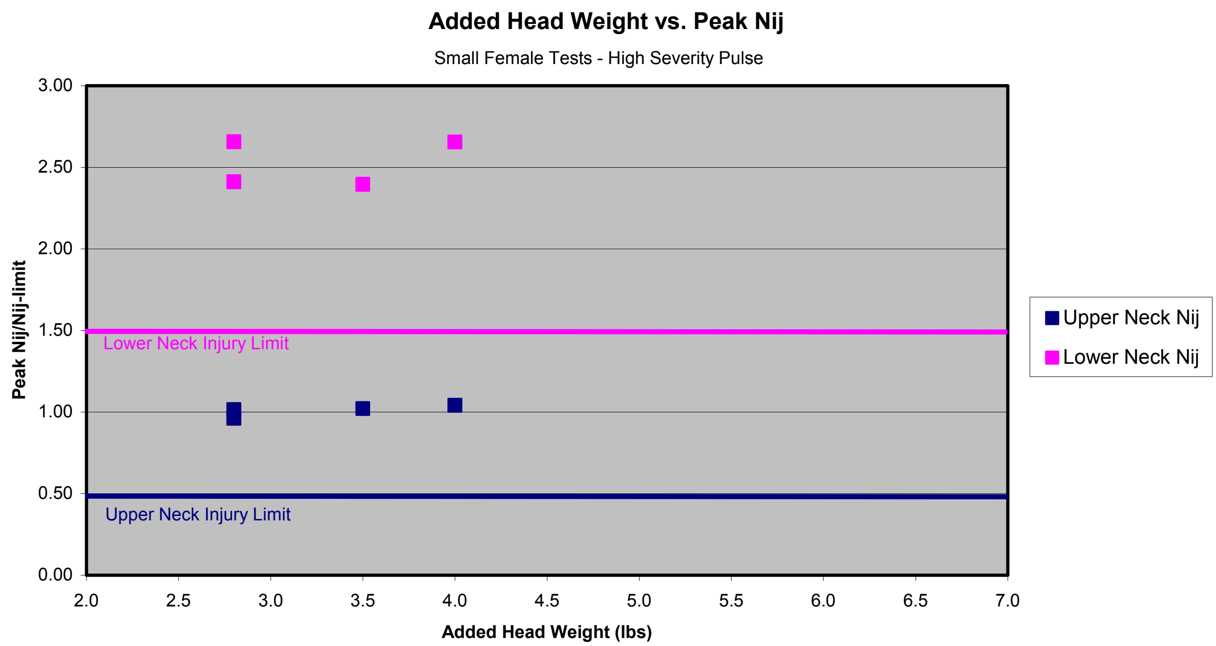

Figure A-3: Peak Nij Values (Small Female, High Severity Pulse)

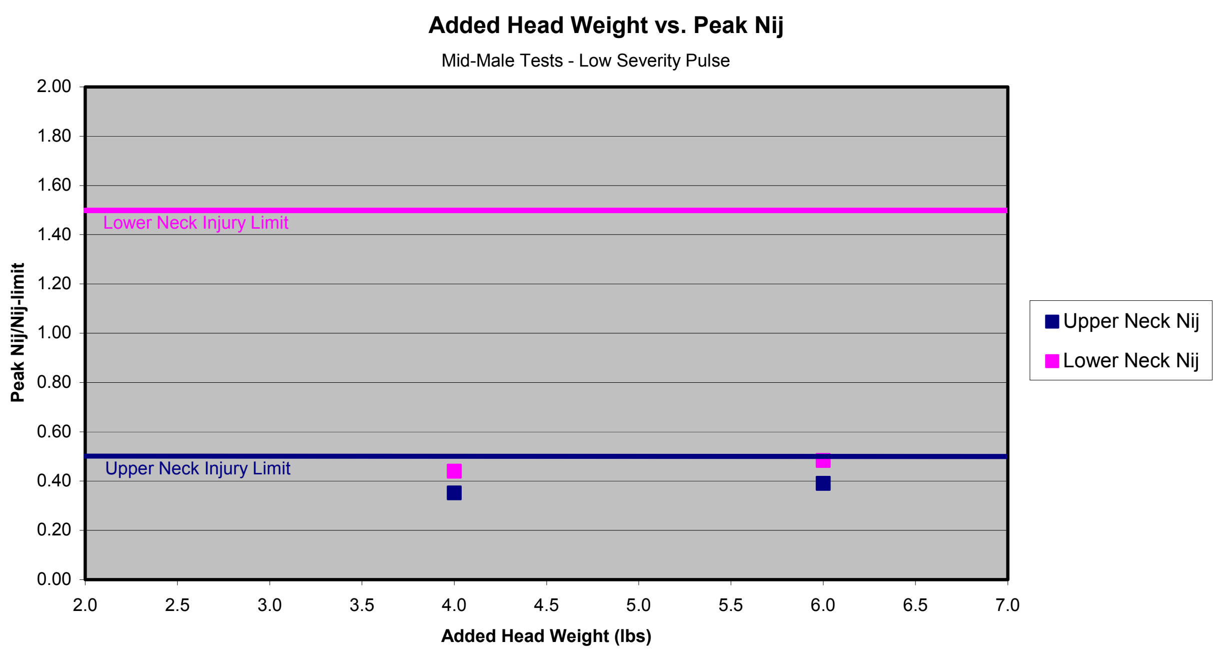

Figure A-4: Peak Nij Values (Mid-Male, Low Severity Pulse)

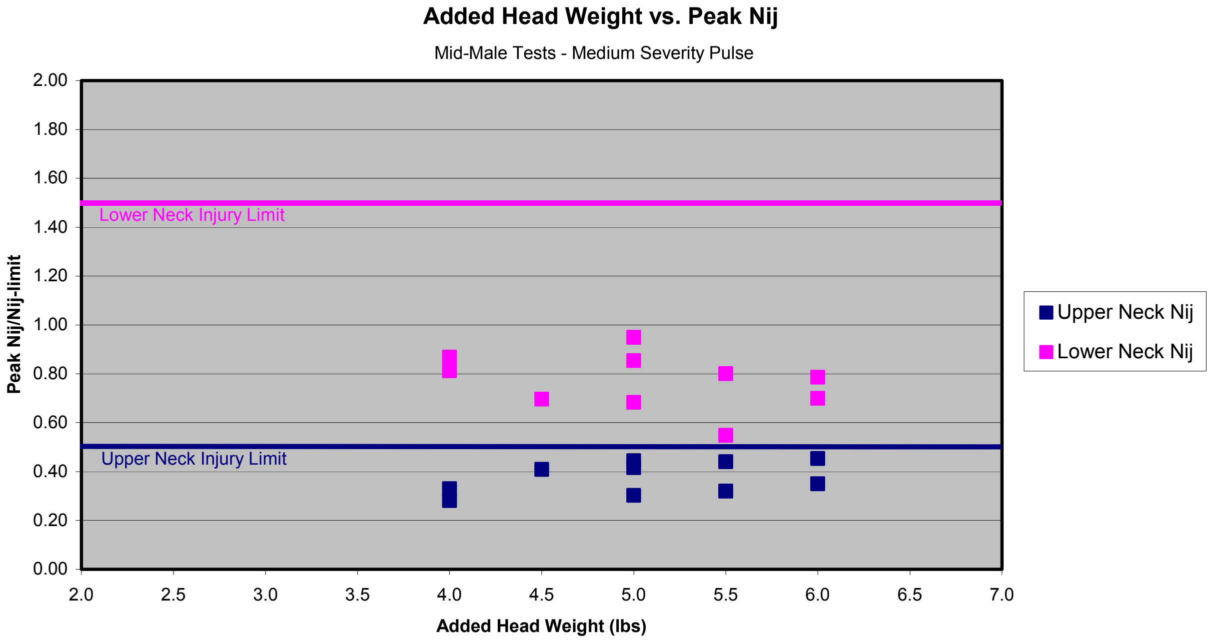

Figure A-5: Peak Nij Values (Mid-Male, Medium Severity Pulse)

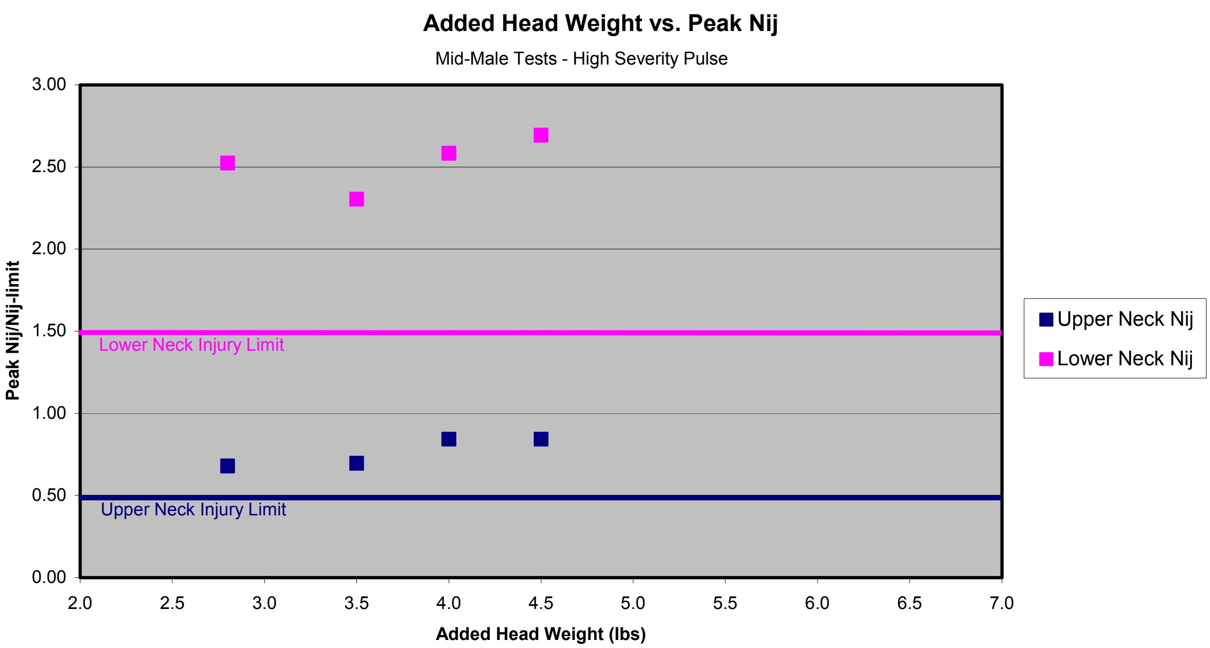

Figure A-6: Peak Nij Values (Mid-Male, High Severity Pulse)

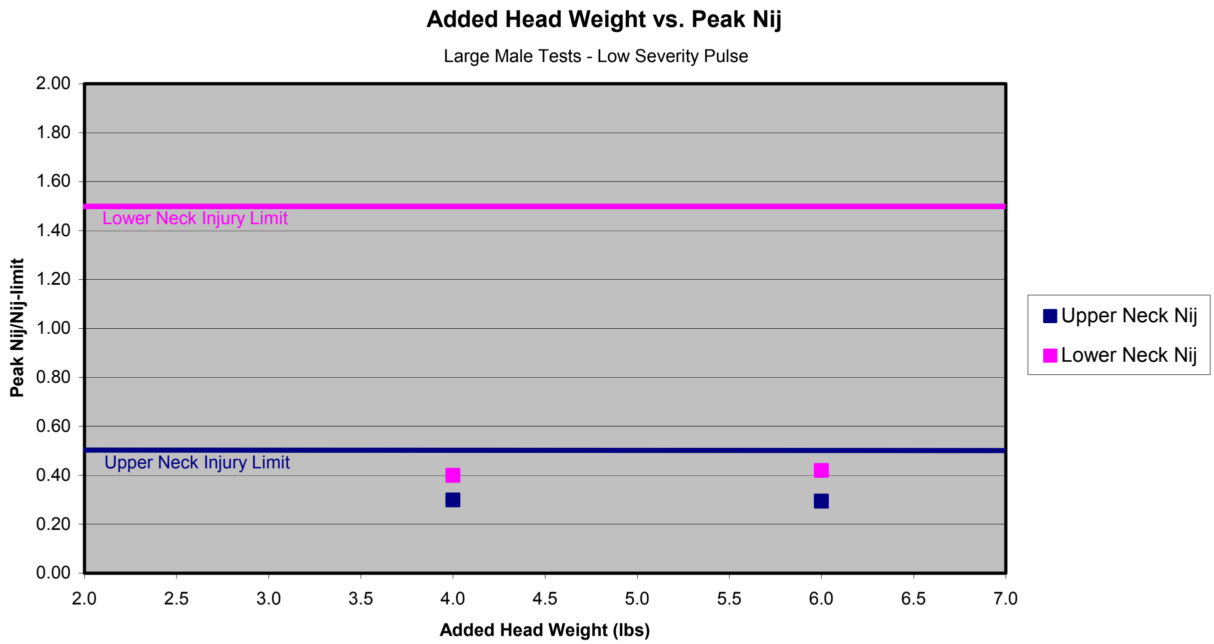

Figure A-7: Peak Nij Values (Large Male, Low Severity Pulse)

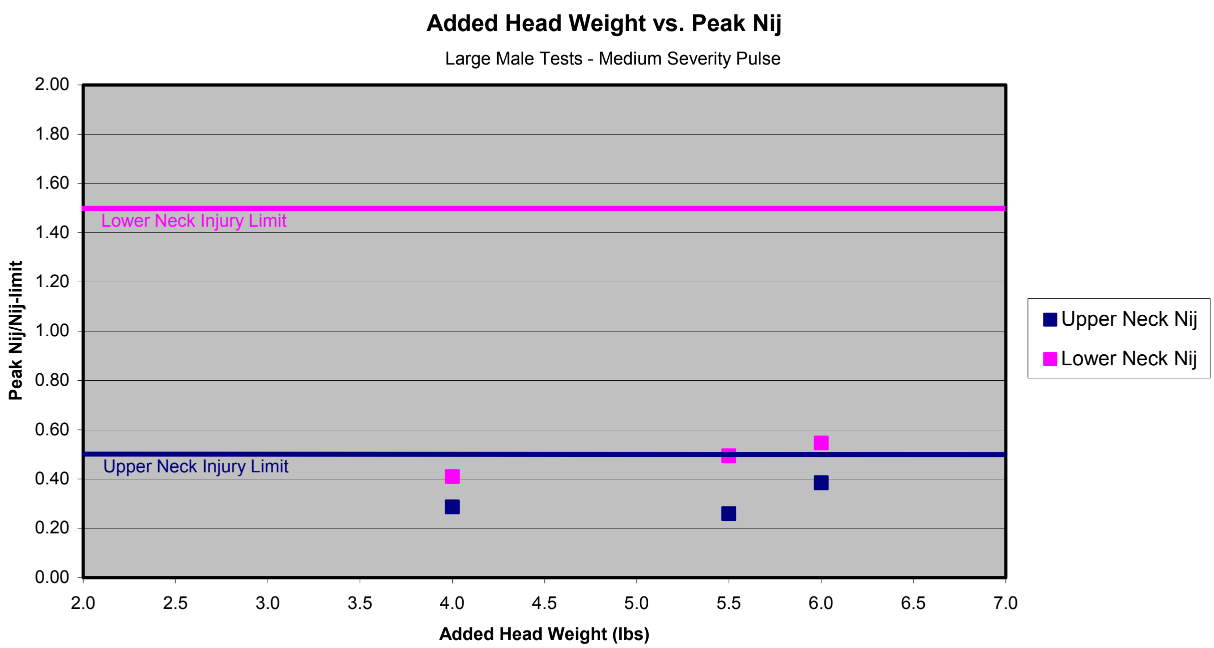

Figure A-8: Peak Nij Values (Large Male, Medium Severity Pulse)

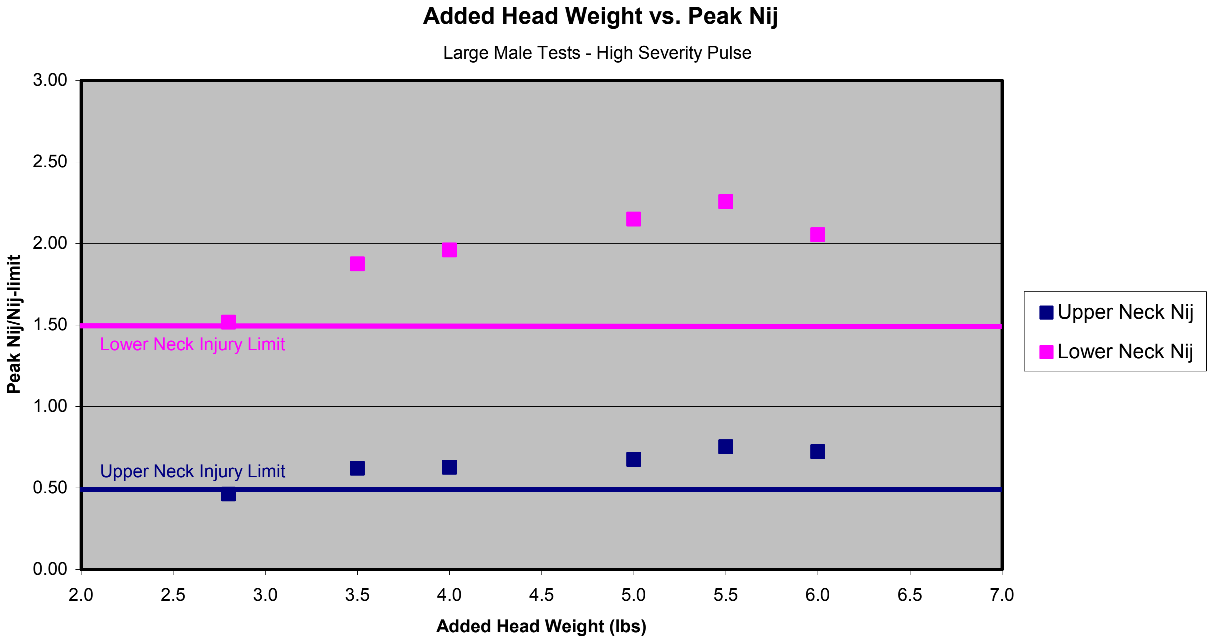

Figure A-9: Peak Nij Values (Large Male, High Severity Pulse)

APPENDIX B: INJURY CRITERIA DESCRIPTION

It is understood that the availability of head/neck injury criteria developed for specific aerospace applications (e.g. ejection seats and helicopter crashworthy seating) is limited, and requires a significant level of further basic research. Human bone and tissue response is highly non-linear and dependent upon the rate of acceleration, the magnitude of acceleration, and the total energy applied to the body. Applying injury criteria to different applications can be potentially misleading and therefore give rise to misinterpretation of results. Specifically, the load profiles and total energy seen in an automotive crash test are substantively different from what is experienced in aerospace environments. Nevertheless, the automotive head/neck criteria represent the only established basis for evaluating head/neck injury. Two such criteria currently exist: the Mertz criteria [12-15] and the Nij [10, 11]. The Mertz criteria relies on the duration of applied loads to tension, compression and shear. The Nij is a combined loading criteria that incorporates peak loads in tension, compression and moments in flexion and extension into a single injury predictor. The criteria each have critical parameters that are scaled according to mass and represent a likelihood (probability) of injury based upon that mass.

Both head/neck injury criteria were deemed inappropriate as they were defined for the automotive environment. Therefore, the critical intercepts were modified to more closely represent the aviation environment and provide a more acceptable level of risk (~5% probability of injury). An extensive analysis of historical ejection and helicopter manikin data was conducted to relate measured forces and moments with observed cervical injuries within the Naval mishap database.

Neck Tension Duration Limits

The maximum acceptable neck tension (lifting force) limits measured at the occipital condyles (C0-C1, upper neck) and cervical vertebrae (C7-T1, lower neck) are defined in Table B.1. These limits represent the maximum allowable load that can be sustained for a given duration.

Table B.1: Neck Tension Force Duration Limits for a Given Occupant Size					
Small Female Hybrid III Type Manikin (96 to 118 lbs)		**Mid-Size Male Hybrid III Type Manikin**		**Large Male Hybrid III Type Manikin** (200 to 245 lbs)	
Time (ms)	**Tension at C0-C1 & C7-T1 (lbs)**	**Time (ms)**	**Tension at C0-C1 & C7-T1 (lbs)**	**Time (ms)**	**Tension at C0-C1 & C7-T1 (lbs)**
5	414	5	618	5	761
31	414	35	618	37	761
40	200	45	320	48	450
80	200	80	320	80	450

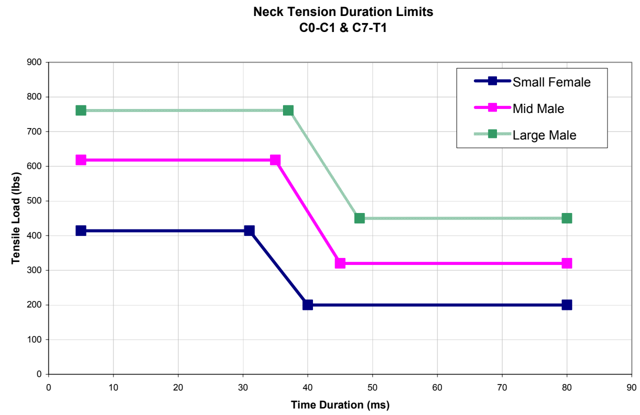

Figure B-1: Neck Tension Duration Limits for Different Manikin Sizes

Neck Compression Duration Limits

The maximum acceptable cervical compression force limits are defined in Table B.2.

Table B.2: Neck Compression Force Duration Limits for a Given Occupant Size					
Small Female Hybrid III Type Manikin (96 to 118 lbs)		**Mid-Size Male Hybrid III Type Manikin**		**Large Male Hybrid III Type Manikin** (200 to 245 lbs)	
Time (ms)	**Compress. at C0-C1 & C7-T1 (lbs)**	**Time (ms)**	**Compress. at C0-C1 & C7-T1 (lbs)**	**Time (ms)**	**Compress. at C0-C1 & C7-T1 (lbs)**
5	519	5	790	5	979
27	200	30	320	32	450
80	200	80	320	80	450

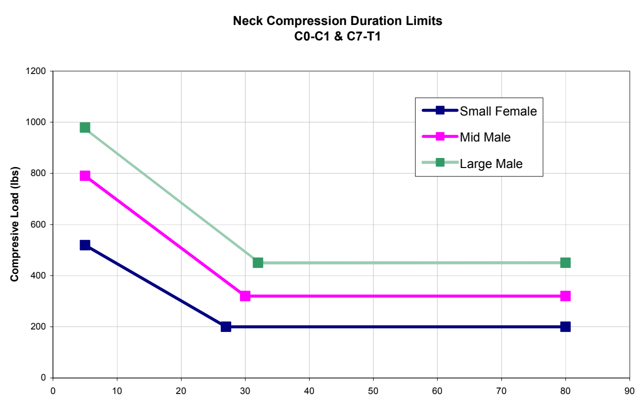

Figure B-2: Neck Compression Duration Limits for Different Manikin Sizes

Neck Shear Force Duration Limits

The maximum acceptable cervical shear force limits are defined in Table B.3. Note that the limit for the lower cervical spine is estimated to be double the limit for the upper cervical spine.

Table B.3: Neck Shear Force Limits for a Given Occupant Size					
Small Female Hybrid III Type Manikin (96 to 118 lbs)		**Mid-Size Male Hybrid III Type Manikin**		**Large Male Hybrid III Type Manikin** (200 to 245 lbs)	
Time (ms)	**Resultant Shear at C0-C1 (lbs)**	**Time (ms)**	**Resultant Shear at C0-C1 (lbs)**	**Time (ms)**	**Resultant Shear at C0-C1 (lbs)**
5	405	5	625	5	777
20	225	25	337	28	414
29	225	35	337	39	414
37	165	45	247	50	304
80	165	80	247	80	304
Time (ms)	**Resultant Shear at C7-T1 (lbs)**	**Time (ms)**	**Resultant Shear at C7-T1 (lbs)**	**Time (ms)**	**Resultant Shear at C7-T1 (lbs)**
5	810	5	1250	5	1554
20	450	25	674	28	828
29	450	35	674	39	828
37	330	45	494	50	608
80	330	80	494	80	608

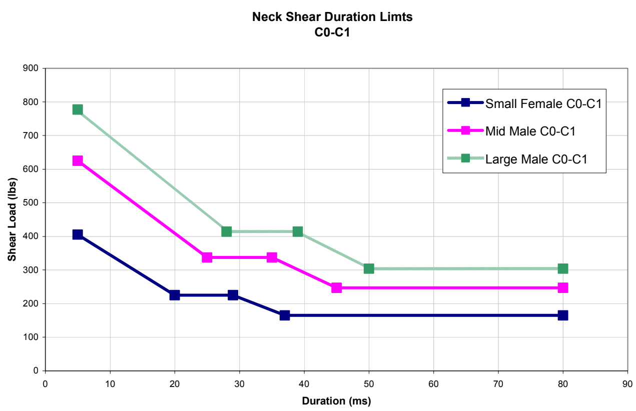

Figure B-3: Upper Neck Shear Duration Limits for Different Manikin Sizes

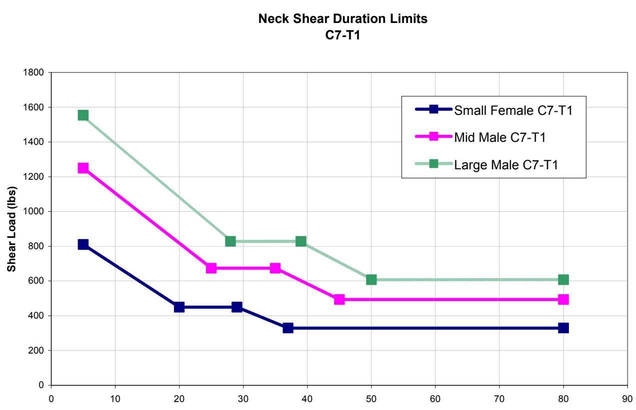

Figure B-4: Lower Neck Shear Duration Limits for Different Manikin Sizes

Combined Neck Moment and Peak Load Limits (Nij)

The maximum combined-cervical-force-and-moment limit, expressed as Neck Injury Criteria (Nij), is 0.5, as measured at the occipital condyles (C0-C1). The maximum Nij as measured at the lower neck (C7-T1) is 1.5. Nij is not applied for pure tension or compression. Nij is calculated from the following equation:

$$N_{ij} = \frac{F_z}{F_{\text{int}}} + \frac{M_y}{M_{\text{int}}} \qquad \textbf{Equation B.1}$$

where:

F_z is the axial tension/compression load.
F_{int} is the critical intercept load (defined in Table B.4).
M_y is the flexion/extension bending moment.
M_{int} is the critical intercept moment (defined in Table B.4).

Table B.4: Critical Intercept Values for N_{ij} Calculation at C0-C1 for a Given Occupant Size			
Critical Parameter (Forces = lbs., Moments = in-lbs.)	**Small Female Hybrid III Type Manikin** (96 to 118 lbs)	**Mid-Size Male Hybrid III Type Manikin**	**Large Male Hybrid III Type Manikin** (200 to 245 lbs)
Tension (+Fz)	964	1530	1847
Compression (-Fz)	872	1385	1673
Flexion (+My)	1372	2744	3673
Extension (-My)	593	1195	1584

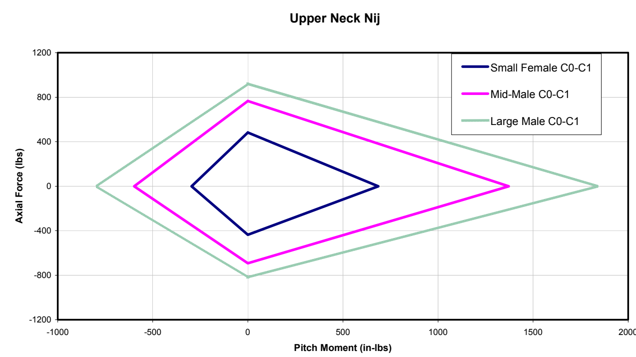

Figure B-5: Upper Neck Nij Limits for Different Manikin Sizes

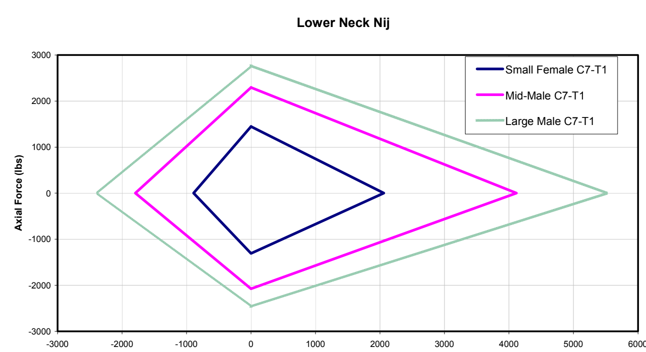

Figure B-6: Lower Neck Nij Limits for Different Manikin Sizes