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Review of Literature Investigating Head-Supported Mass and Acute Neck Injury Risk

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14. ABSTRACT

Military helmet systems have become increasingly complex over the years due to night vision goggles (NVGs), communication devices, and other equipment needed by the Warfighter. The increased mass has led to neck injuries not previously seen by Soldiers in operational environments. The mathematically-derived U.S. Army Aeromedical Research Laboratory (USAARL) Head-Supported Mass (HSM) Curves, based on the aviation environment, serve as well-established acute neck injury guidelines for HSM for the U.S. Army (McEntire & Shanahan, 1997). The curve representing acute injury risk is based on the vertical center of mass (CM) offset of HSM configurations and correlates to a high vertical acceleration exposure. No acute neck injury guidelines for HSM exist for exposures with primarily longitudinal loading or lateral loading. Since the development of the USAARL HSM Acute Injury Curve, several new neck injury criteria (without HSM-effects) have been proposed for both the upper neck and the lower neck; however, there have been no updates to HSM guidelines. The goal of this report is to summarize the findings of research investigating acute injury risk with HSM since 1997 (including the research that is used for the USAARL Acute Injury Curves) and determine gaps in research that are needed in order to create more informative U.S. Army HSM acute injury risk guidelines.

15. SUBJECT TERMS

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Summary

Military helmet systems have become increasingly complex over the years due to the addition of helmet-mounted equipment, such as night vision goggles (NVGs), communication devices, and other equipment needed by the Warfighter. The increased mass has led to neck injuries not previously seen by Soldiers in operational environments. The mathematically-derived U.S. Army Aeromedical Research Laboratory (USAARL) Head-Supported Mass (HSM) Curves, based on the aviation environment, serve as well-established acute neck injury guidelines for HSM in the U.S. Army (McEntire & Shanahan, 1997). The curve representing acute injury risk is based on the vertical center of mass (CM) offset of HSM configurations and correlates to a high vertical acceleration exposure. No acute neck injury guidelines for HSM exist for exposures with primarily longitudinal loading or lateral loading. Since the development of the USAARL HSM Acute Injury Risk Curve, several new neck injury criteria (without HSM-effects) have been proposed for both the upper neck and lower neck; however, there have been no updates to HSM guidelines. The findings from the USAARL HSM Acute Injury Risk Curves were published over 20 years ago and have yet to be updated or expanded beyond the vertical environment even though further neck injury research has occurred. The goal of this report is to summarize research investigating acute injury risk with HSM since 1997 and determine the gaps in knowledge where more research is needed to create more informative U.S. Army HSM acute injury risk guidelines.

A comprehensive literature search identifying research on the relationship between HSM characteristics and neck loading was performed. Research must have focused on establishing the association between HSM characteristics and neck loading with the purpose of evaluating acute injury risk, otherwise it was excluded from our results. Additionally, research must be available in the open literature or as a published technical report.

A total of ten articles, published between 1997 and 2016, were identified that met our inclusion criteria. The studies included human volunteer response (N = 3), post-mortem human subject (PMHS) response (N = 1), mechanical surrogates (e.g., the Hybrid III Anthropometric Test Device [ATD]) response (N = 5), and mathematical or computational methods (N = 4) to estimate injury risk. Three of the ten studies incorporated multiple methods described above. Studies included multiple exposure directions (e.g., axial, frontal, and lateral), a range of mass conditions (i.e., both within the existing USAARL HSM Acute Injury Curve guidelines and exceeding the guidelines), and a range of center of mass offsets (i.e., within and exceeding the USAARL HSM Acute Injury Curve limits).

Overall, there are several major gaps in the literature regarding the contribution of HSM to acute neck injury risk. A significant gap identified was the paucity of PMHS and finite element model research in this area. Only one study assessed the validity of using existing acute neck injury risk criteria (Nij); it noted several concerns about the reliability of the Nij for evaluating injury risk due to the HSM effects on neck kinematics and ATD biofidelity. Furthermore, there has been little research aimed at developing HSM guidelines, but rather HSM effects are assessed using existing neck injury criteria.

Our review has shown that while several efforts have been directed towards investigating HSM effects on acute neck injury in the last 25 years, little progress has resulted towards improved guidance on HSM acute injury risk. Several volunteer studies have been performed,

and while estimated neck loads were below injury thresholds, studies reported that multiple subjects developed pain. However, these studies did not provide sufficient data for the development of updated guidance on defining HSM limits. In addition, multiple studies were conducted with ATDs, which were not designed for HSM wear nor evaluation, over a wide range of dynamic loading conditions. These studies required innovative adaptations to experimentally investigate HSM effects on human biomechanical response. While the ATD studies confirmed a relationship between HSM conditions (mass and CM offset) and potential increases in acute injury risk, the studies focused on the upper neck injury risk. The lack of any lower neck ATD injury assessment reference criteria at the time prevented the reviewed ATD and computational modeling studies from investigating injury risk to the lower neck, which is a region of concern for Service Members wearing HSM. Additionally, all existing acute cervical injury criteria, except one, were not developed for evaluation of HSM. Only Bass et al. (2006) included HSM influence in the development of an acute cervical injury criterion, but the resulting lower neck injury criterion did not provide direct guidance on HSM (mass and CM offset) limits. After review of the limited available studies and with consideration of their respective limitations, the McEntire and Shanahan (1997) HSM guidance, while fundamental in its development, remains the most relevant and only biomechanically based guidance for acute neck injury risk due to HSM (mass and CM offset). More research is needed to develop lower neck injury criteria and improve ATD cervical spine biofidelity. This research would lead to improved HSM guidance for future helmet and helmet mounted device development efforts.

A multifaceted research approach is needed to update HSM guidance that accurately reflects neck injury risk for U.S. Army Aviators and ground Soldiers. PMHS testing using loading profiles representative of aircraft and ground vehicle mishaps with HSM are needed to establish threshold limit values. The PMHS results then need to be compared to matched pair ATD testing to assess the biofidelity of the ATD to create appropriate scaling ratios or Injury Assessment Reference Curves (IARVs) to assess future HSM conditions. Several of the studies reviewed were funded by Department of Defense components and additional contractor reports and data may be available for retrospective analyses to supplement the needed studies. Such a multifaceted approach will provide significant advancement in the understanding of HSM acute neck injury risk. Ultimately, this research would improve our Service Members' protection from acute cervical neck injury.

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Introduction

Military helmet systems have become increasingly complex over the years due advancements in available helmet-mounted technology for operational missions, such as night vision goggles (NVGs), communication devices, helmet-mounted displays (HMDs), and other equipment giving the Warfighter greater capabilities and advantages on the battlefield. The increased mass of the helmet systems due to these additions has led to neck injuries not previously incurred by Soldiers in operational environments (Brozoski et al., 2020; Bass et al., 2006; Shannon et al., 1998; McEntire & Shanahan, 1997; Shanahan & Shanahan 1989). The Ballistic and Blast Protection Community of Practice recognized and documented a need for acute neck injury risk criteria associated with head-supported mass (HSM) (Hoppel et al., 2016); the only current U.S. Army guidance is the mathematically-derived U.S. Army Aeromedical Research Laboratory (USAARL) HSM Curves (Madison, 2019; Estep et al., 2019; McEntire & Shanahan, 1997), which are based on aviation crash exposures.

Since publication of the USAARL HSM Curves in 1997, the U.S. Navy, Product Manager Air Warrior (PdM-Air Warrior), and Combat Capabilities Development Center (CCDC) Command, Control, Computers, Communications, Cyber, Intelligence, Surveillance and Reconnaissance (C5ISR) Center have relied on this information to guide aviation helmet and head-mounted ancillary device development. The USAARL HSM Acute Injury Risk Curve provides guidelines for regions of acceptable risk versus severe injury risk using HSM conditions (in existence at the time), tension failure criteria (Mertz, 1993), and known exposure conditions (based on helicopter seat performance specifications [MIL-S-58095]) (Military Specification, 1971). The USAARL HSM Acute Injury Risk Curve uses a loading vector and magnitude representative of aviation crash environments (combined vertical and horizontal accelerative loading) against a neck tension tolerance to estimate the risk of acute neck injury. The findings from the USAARL HSM Acute Injury Risk Curves were published over 20 years ago and have yet to be updated or expanded beyond the vertical environment even though further acute neck injury research has occurred.

Since the development of the USAARL HSM Curves, several acute neck injury tolerances (e.g., Nightingale et al., 2007a; Nightingale et al., 2007b; Dibb et al., 2006; Chancey et al., 2003) and new acute neck injury criteria (without HSM-effects) have been proposed for both the upper neck and the lower neck. Notably, many of the acute neck injury criteria (upper and lower) incorporate both axial force (e.g., tension or compression) and sagittal plane moment (e.g., flexion or extension) in the prediction of injury (Yoganandan et al., 2020; Chirvi et al., 2015; Eppinger et al., 1999). The inclusion of both modalities enables recent acute neck injury criteria to be more broadly applicable for multiple loading directions. The most widely used recent acute injury criteria for the neck, the Nij criteria developed by Kleinberger et al. (1998), incorporates both axial force and moment to predict upper neck injuries. Although there have been significant efforts in the past to investigate upper neck injuries and associated acute injury criteria, investigations regarding lower neck injury risks have only occurred recently (Chancey et al., 2003; Pintar et al., 2005; Dibb et al., 2006; Nightingale et al., 2007a; Nightingale et al., 2007b; Pintar et al., 2010a; Pintar et al., 2010b; Chirvi et al., 2015; Kang et al., 2016; Yoganandan et al., 2020). One such notable study, developed a new acute injury criteria (LNij) using previous post-mortem human subject (PMHS) testing to create a lower neck acute injury risk assessment (Chirvi et al., 2015). However, none of the cited studies used HSM for the development of acute injury criteria or injury risk curves. Instead of developing HSM-specific acute neck injury risk curves, researchers have used existing criteria to assess HSM conditions (Paskoff & Sieveka, 2004; Merkle et al., 2005; Parr et al., 2012; Gaur et al., 2013).

One study, funded by USAARL, developed an acute injury criterion for the lower neck and included HSM in its development (Bass et al., 2006). The study focused on the lower neck due to evidence showing that the inclusion of HSM may shift the location of injury from the upper neck (as is commonly reported in civilian environments) to the lower neck (Brozoski et al., 2020; Shannon et al., 1998; Shanahan & Shanahan, 1989). The study by Bass et al. (2006) is the only study to incorporate various HSM conditions in the development of their proposed acute neck injury criteria, the Beam Criteria, for lower neck injuries with HSM in aviation environments. However, the study by Bass et al. (2006) does not provide recommended limits for HSM and does not include HSM limits as a covariate in the provided risk curve.

While there has been significant advancement in acute neck injury criteria in recent years, very few studies have investigated the role of HSM related variables (e.g., center of mass [CM] offset or total mass) on acute injury risk. The effects of the mass and CM offset of HSM as a covariate in studies of acute neck injury risk have not been well explored. The goal of this report is to summarize the findings of research that has occurred regarding HSM acute neck injury risk since 1997 (including the research that is used for the existing acute neck injury HSM guidelines) and determine the gaps in research in order to create more informative U.S. Army HSM acute injury risk guidelines.

Methods

A comprehensive literature search identifying research investigating the relationship between HSM characteristics and neck loading was performed. Research must have focused on establishing the association between HSM characteristics and neck loading with the purpose of evaluating acute injury risk, otherwise it was excluded from our results. Additionally, reviewed publications included those available in the open literature or as a public release technical report. The review was restricted to publications that occurred after the establishment of the USAARL HSM Acute Injury Curve (McEntire & Shanahan, 1997). The following keywords were used in the search of the Medical Research & Development Command Library Electronic Access Portal: neck, loading, acute, injury criteria, helmet, weight, mass, and HSM. All loading directions were of interest in our review.

For consistency in reviewing the literature, all studies were compared using common orientations and terminology for directions of forces on the body. Standard terminology for directions of forces on the body, as defined in Volume II of the Aircraft Crash Survival Design Guide (Coltman et al., 1989), were used (Figure 1). The body fixed coordinate system was defined with the positive X-axis coming out of the chest, the positive Y-axis oriented to the right side of the body, and the positive Z-axis coming out of the top of the head. While this is not a right-handed coordinate system, it is in accordance with the Aircraft Crash Survival Design Guide. For complex loading vectors (i.e., those not aligning with a primary axis defined above), all component directions of loading were described with relative angles from primary axes.

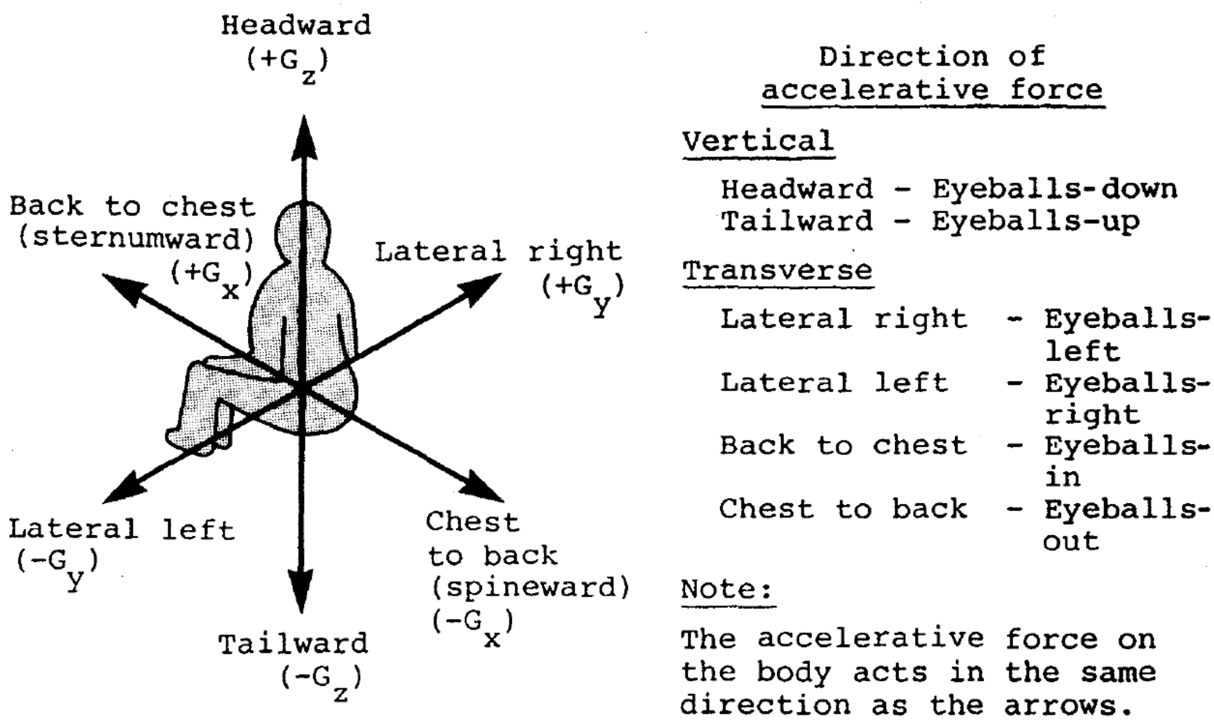

Figure 1. Terminology for directions of forces on the body. Adapted from Volume II of the Aircraft Crash Survival Design Guide (Coltman et al., 1989).

Results

A total of ten articles, published between 1997 and 2016, that met our inclusion criteria were identified. The studies included human volunteer response (*N* = 3), PMHS response (*N* = 1), mechanical surrogates (e.g., the Hybrid III Anthropometric Test Device [ATD]) response (*N* = 5), and mathematical or computational methods (*N* = 4) to estimate risk. Three of the ten studies incorporated multiple methods described above. Exposure conditions included a wide range of severities (from as low as four times the acceleration due to gravity [G] to as high as 35 G) and multiple impact vectors, including side impacts (Gy), frontal impacts (-Gx), and complex loading vectors (e.g., combined vertical [+Gz] and frontal [-Gx] approximating an aircraft crash vector). In addition, exposure conditions included multiple methods for approximating the effects of energy attenuating (EA) seats, such as modeling of the EA seats to determine an analogous pulse for experimental testing, directly estimating the resulting exposure based on theoretical performance of the EA seats (e.g., based on MIL-S-58095 [Military Specification, 1971]), or including existing EA seats in the study directly. HSM conditions ranged from 0 kilograms (kg) to 6.81 kg and added CM offset included both longitudinal (e.g., forward or rearward) offsets and vertical offsets using varied methods for defining the CM position (several studies did not report the HSM offsets used). The identified articles are summarized in the sections below. In addition, a summary of the articles is provided in Appendix A, as well as a comparison of the tested conditions (i.e., exposure severity, HMS mass, and HSM offset) versus the existing USAARL HSM Acute Injury Risk Curve (McEntire & Shanahan, 1997).

Literary Sources

McEntire, B. J., & Shanahan, D. F. (1997). ***Mass requirements for helicopter aircrew helmets*** **(Report No. 98-14). U.S. Army Aeromedical Research Laboratory.**

With increasing mission requirements, the military added additional mass onto aircrew helmets. A study by McEntire and Shanahan (1997) noted that early helmet performance specifications for helmet mass and CM were written vaguely and did not incorporate a scientific justification. In order to create clear recommendations with a scientific justification, the mathematically-derived USAARL HSM Curves were developed. The two USAARL HSM Curves were derived by calculating the theoretical maximum allowable HSM for an acceptable injury risk or performance deficits and establishing its relationship with vertical and longitudinal, respectively, center of mass offset. For the present report, only the USAARL HSM Acute Injury Curve is discussed.

Newton's Second Law was used to determine the maximum allowable HSM acting on the cervical/thoracic spine juncture (C7/T1) (Eq. 1). Through a review of previously current literature, the maximum neck tensile strength was determined to be 4050 Newtons (N). Acceleration was determined by the worst-case accident scenario at the time (35 G directed 31.4 degrees downward from horizontal). The 35 G crash pulse was the resultant of a seat experiencing 30 G longitudinally and stroking with a peak vertical load of 18.3 G combined with a 1.5 over-shoot factor. Simplifying Eq. 1 results in a maximum allowable total mass (m_{total}) of 7.86 kg. Subtracting out the head (m_{head} = 4.32 kg) and neck (m_{neck} = 1.04 kg) mass, the mass of the helmet system (m_{helmet}) became 2.5 kg (Eq. 2).

$$4050\ N = m_{total} \times (35\ \times 1.5 \times 9.81\ m/s^2) \qquad \text{Eq 1.}$$

$$m_{helmet} = m_{total} - m_{head} - m_{neck} \qquad \text{Eq 2.}$$

The vertical CM offset limit was calculated with the assumption of a constant mass moment acting about the C7/T1 juncture. The known mass and CM offset of the AH-1 Cobra helmet, the worst-case fielded helmet system at the time, and the 95th percentile female (85th percentile male) neck length was assumed to be the maximum limit of vertical distance between C7/T1 juncture and CM. The mass (1.74 kg), the vertical CM distance from the tragion notch (5.2 centimeters [cm]) of the AH-1 Cobra helmet, and the vertical distance between C7/T1 and the tragion notch for a 95th percentile female (Z_{neck} = 11.94 cm) were used to calculate the maximum allowable moment (M_{crit}) of 29.8 kg-cm. The AH-1 Cobra helmet was chosen as its inertial properties represented the highest mass and vertical mass offset of U.S. Army helmets being flown at the time of the report. Using the resulting moment, the mass moment equation was rearranged to show the relationship between vertical CM offset (Z_{offset}) and helmet system mass (Eq. 3).

$$Z_{offset} = \left(\frac{M_{crit}}{m_{helmet}}\right) - Z_{neck} \qquad \text{Eq 3.}$$

Through the concepts explained above, a relationship between acute neck injury risk, the vertical CM and mass (Figure 2) was defined. The existing curve uses the maximum offset of 5.2 cm from the worst case helmet CM offset and a mass of 2.5 kg based on published neck tensile strength available at the time of publication as the upper bounds.

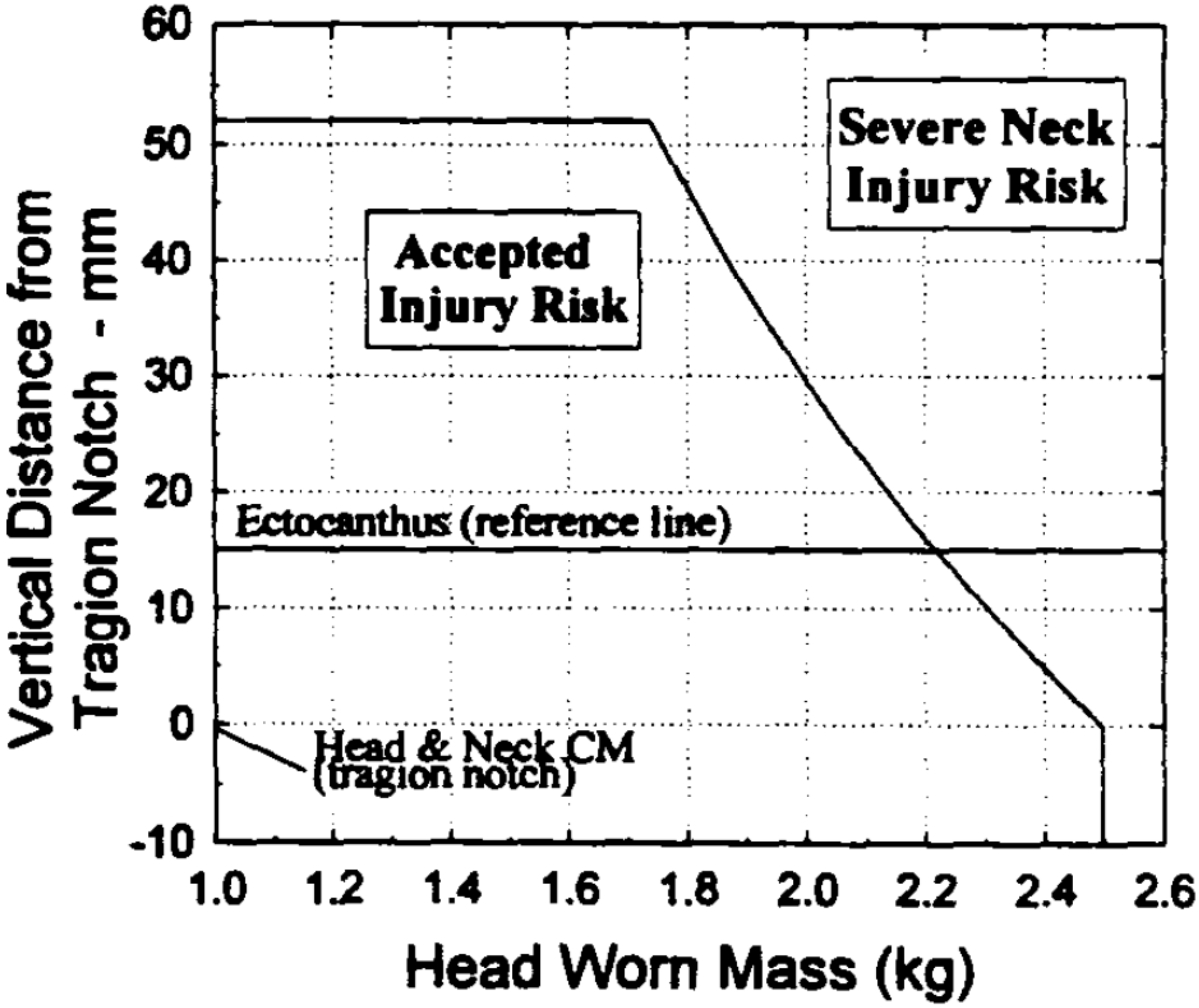

Figure 2. The U.S. Army Aeromedical Research Laboratory Head-Supported Mass (USAARL HSM) Acute Injury Curve depicting the vertical center of mass placement guidance as a function of head worn mass from the original manuscript by McEntire and Shanahan (1997).
Note. The center of mass (CM) offset limit at 52 mm was obtained using the worst case helmet CM offset at the time of the report (the AH-1 Cobra helmet) and the mass limit at 2.5 kg was calculated using maximum neck tensile strength available in literature at the time of the report.

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Perry, C. E., Buhrman, J. R., Doczy, E. J., & Mosher, S. E. (2003). ***Evaluation of the effects of variable helmet weight on human response during lateral + Gy impact.*** **(AFRL-HE-WP-TR-2004-0013). Air Force Research Laboratory.**

The U.S. Air Force designed a study to investigate the effects of HSM on neck loads in lateral accelerative loading (Gy) (Perry et al., 2003). No acute neck injury criteria were calculated as a result of this study; however, the results of the volunteer study were compared to previously conducted research and guidelines (Note: Reported at +Gy impact by coordinate system in Perry et al. (2004), but -Gy according to adopted coordinate system for current review [Figure 1])

A total of 21 male and 10 female volunteers (plus an instrumented, large Advanced Dynamic Anthropomorphic Manikin [ADAM]) were subjected to lateral accelerative loading with varied magnitude and helmet mass. Accelerations ranged from 4 to 6 G and helmet mass conditions were 0 pounds (lb) (0 kg), 3.0 lb (1.4 kg), and 4.5 lb (2 kg). The helmet CM offset was not documented. The human accelerative response in the X-direction (i.e., anterior/posterior) and Y-direction (i.e., lateral) was measured, as well as head rotation about the Y- (i.e., flexion/extension) and Z- (i.e., torsion or twisting) axes. Measured head accelerations were used with the estimated inertial properties of the head and helmet system to mathematically compute neck moments and loads at the occipital condyles (Note: The rotation about the X-axis [i.e., lateral bending)], and therefore the moments about the X-axis, were not calculated due to instrumentation limitations noted by the authors).

Perry et al. (2003) found that head accelerations, neck loads, and neck moments increased with the increase in acceleration magnitude. Head accelerations, neck loads, and neck moments increased when subjects went from a ‘no helmet’ condition to 3 lb (1.4 kg) condition, but decreased when helmet weight increased to 4.5 lb (2 kg). Authors attributed the observed decreases with increase helmet weight to subjects bracing for impact, as subjects were more aware of the increase in helmet weight.

Despite no acute neck injury criteria being calculated, the results were compared to previously conducted research and guidelines. For the neck loads and moments that were calculated, both Y (lateral) shear force (225 pounds force [lbf] [1001 N]) and Y (flexion/extension) moment (240 inch pounds [in-lb] [27 Newton-meter [N-m]]) were higher than the recommended safe guidelines of 90 lbf (400 N) and 133 in-lb (15 N-m), respectively, as previously determined by the Naval Biodynamics Laboratory (Weiss et al., 1989). However, all loadings were tolerated by the study subjects. Measured loads were extrapolated to predict shear loading outcomes at higher magnitudes. These extrapolated results, Y shear force (389 lbf [1730 N]) and Y moment (411 in-lb [46 N-m]), were found to be less than the suggested injury threshold of 1680 in-lb (190 N-m) set forth by Mertz and Patrick (1971). The authors noted that the suggested thresholds published in Mertz and Patrick (1971) were for accelerative loading in the -Gx (frontal) direction and thus not an accurate comparison.

Perry et al. (2004) further included some analysis, discussion, and implications of comparisons between female and male responses, impact of trained response and bracing, and the influence of the headrest on human response to HSM during high accelerative loadings.

Doczy, E., Mosher, S., & Buhrman, J. (2004). The effects of variable helmet weight and subject bracing on neck loading during frontal -Gx impact. *Proceedings of the 42nd Annual SAFE Symposium*, Salt Lake City, UT. Sept 27-28, 186-192.

In order to aid in the development of HSM guidelines, Doczy et al. (2004) measured human volunteer response to frontal loading (-Gx) under various weighted helmet conditions, as well as the effect of bracing for impact on resulting neck loads. Thirty-four (34) volunteers, 18 males and 16 females, were subjected to horizontal accelerations of 6 G, 7 G, 8 G, and 10 G with three different helmet conditions: (1) no helmet, (2) Head Gear Unit Number 56 Personal (HGU-56/P), and (3) variable weighted impact helmet (VWI). Helmet mass conditions were 2.0 lb (0.9 kg), 3.0 lb (1.4 kg), 3.5 lb (1.6 kg), 4.0 lb (1.8 kg), and 4.5 lb (2 kg). All volunteers were instructed to brace their head against the headrest during the sled tests. A subset of volunteers were instructed to use a lighter brace to prepare for impact. Sled velocity, sled accelerations, seat accelerations, subject head and torso accelerations, subject head and torso displacements, and forces developed in the seat and restraint system were collected. Neck forces and moments were calculated about the occipital condyles using the measured accelerations, subject anthropometry, and helmet inertial properties. Experimental and calculated results were used to create a linear analysis of covariance (ANCOVA) model that predicted neck loads about the occipital condyles from seat accelerations and helmet weights. The experimentally derived numerical model was used to estimate neck loads at higher accelerations than tested.

From the experimental results, a slight increase in neck loads was observed with increasing helmet weight while keeping accelerations constant at 8 G except for the 4 lb (1.8 kg) helmet condition where a reduction from the 3.5 lb (1.59 kg) condition was observed for both males and females. Furthermore, pain was reported from both male and female subjects. Three male subjects who experienced pain had an average calculated neck load of 167 lbf (743 N). Maximum calculated neck loads, including results from participants who experienced pain, were 265 lbf (1179 N) for males and 190 lbf (845 N) for females. The study did show that intentionally bracing for impact with HSM did reduce the head accelerations and subsequently the neck forces. The numerical analysis of covariance (ANCOVA) model extrapolated the relationships found between neck loads, acceleration magnitude, and helmet mass from the measured results. In the numerically modeled max condition, 15 G and 5.5 lb (2.5 kg) helmet, loads range from 273 - 350 lbf (1214 – 1557 N). The predicted results were lower than cadaver experiments under similar conditions (Mertz & Patrick, 1971), which, the authors note, could have been due to modeling of experimental results where the participants braced themselves before impact.

The authors noted that while useful information was presented regarding volunteer human subjects' response to HSM and accelerative loading, including reaching the pain threshold for a few subjects, the conditions might not accurately reflect more severe accelerative loading. The findings of this study demonstrate the relationship between helmet mass and neck loads in frontal impacts in human volunteers. The authors further noted that the numerical model extrapolated the results linearly, but it is unfounded if this is the true relationship across the large range of accelerations. Furthermore, it was stated that subjects being able to brace for impact could confound the numerical results and could be why they are not reflective of PMHS results.

Paskoff, G., & Sieveka, E. (2004). Influence of added head mass properties on head/neck loads during standard helicopter impact conditions. *Conference Proceedings 42nd Annual SAFE Symposium.* Salt Lake City, UT. Sept 27-28, 20-40.

The study by Paskoff and Sieveka (2004) used computational modeling results combined with experimental data obtained from multiple ATDs tested on a horizontal sled to determine the effects of varying helmet weight and CM and identify maximum requirements for weight and CM using existing neck injury criteria. The MAthmatical Dynamic Model (MADYMO) model, which was validated during a previous phase of testing, was used to define the pulse for the horizontal sled testing. Computer simulations were used to determine the pulse required to mimic an energy attenuating seat using a rigid seat in order to guide the physical experimentation. However, the authors deemed that the pulse of a stroking seat was too complex to recreate and thus developed a trapezoidal pulse used for the experiment that match the energy from a stroking seat. Simulations that reconstructed the experimental testing were performed for the 4.0 lb (1.8 kg) HSM condition for all ATD sizes. It was concluded that the MADYMO model under predicted peak values for the neck parameters in the low and moderate severity pulse, and peak values occurred earlier than in the sled tests for the high severity pulse.

Experimental data were obtained from a 95th percentile male, a 50th percentile male, and a 5th percentile female Hybrid III ATDs using horizontal sled tests. ATDs were seated in an in-house designed, rigid, crashworthy seat oriented in a recumbent position (i.e., laying on the back). The seat was then rotated 30 degrees (°) up from horizontal and subjected to accelerative loading creating a primary Gz exposure with an additional -Gx component (Figure 3). However, in the high severity case, a forward angle of 10° was added to the torso with the addition of a back pad. This compensated for the significant forward lean the body experienced just before maximum head motion. The ATDs were modified using a custom headform enabling direct mounting of variable masses at pre-defined positions in order to simulate the combined mass and mass offset conditions of a helmeted head (Figure 4A). The modified headform attached to the Hybrid III neck (Figure 4B).

Figure 3. Seating orientation for experimental and computational tests conducted by Paskoff et al. (2004).

All three ATDs were subjected to a low (25 ft/s [7.6 m/s]), medium (31.5 ft/s [9.6 m/s]), and high (50ft/s [15.2 m/s]) pulse severity. The authors note that the medium and high pulses were representative of standard seat qualification crash corridors for a variety of rotary-wing platforms, while low pulses were considered survivable for all occupants. Simulated helmet systems included varying head mass (ranging from 2.8 lb [1.3 kg] to 6 lb [2.7 kg]) and varying

CM offsets defined using polar coordinates with the occipital condyle as the origin. The radius ranged from 2.25 inches (5.70 cm) to 5.00 inches (12.70 cm), while the angles were between 66.7° and 77.5°s off of the horizontal axis (Figure 3b) (Note: The above values were obtained from the target test conditions reported in Paskoff and Sievaka [2004]).

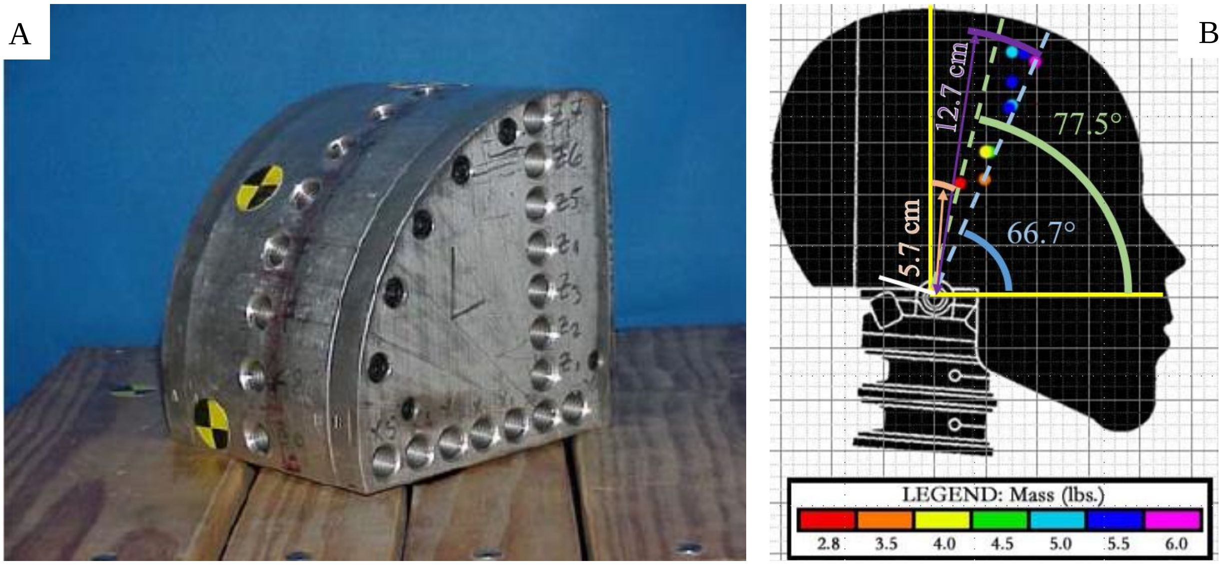

Figure 4. (A) Modified metal headform replacing the Hybrid III headform during experimental tests, and (B) spherical coordinate representation for CM Position as defined in Paskoff et al. (2004).

Axial tension/compression load and flexion/extension moment were measured for each of the test runs. The results were compared to the Mertz Criteria and also used to calculate the Nij, which was used to assess the likelihood of acute neck injury (Table 1). The Mertz criteria relies on the duration of applied loads in addition to peak values (Mertz et. al, 1989). Pulse severity was found to be the factor with the greatest impact on Nij outcomes. All Hybrid III models in all weight conditions were below the Nij threshold for the low severity pulse condition. The 50th percentile and 95th percentile male were below the Nij threshold for the medium pulse severity for all weight conditions, while only the 3.5 lb (1.6 kg) weight condition for the 5th percentile female was under the Nij threshold. The 95th percentile male with the lowest weight condition (2.8 lb or 1.3 kg) was the only ATD and helmet condition combination that did not exceed the Nij threshold when exposed to the high pulse severity. The rest of the tests using the high pulse severity all exceeded the Nij threshold.

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Table 1. Summary of Target Test Configurations and Injury Criteria Results (Adapted from Paskoff & Sieveka [2004])

Test #	ATD (Percentile)	Test Pulse Configuration	HSM (weight and CM Location)	Theta (deg)	Nij Results	Mertz Criteria Results
Low Severity Pulse						
514	5th Female	Max G = 16.25 V (ft/s) = 25.0	W = 4.0 lb R = 3.0 inches	70.1	Pass	Pass
515	5th Female	Max G = 16.25 V (ft/s) = 25.0	W = 5.0 lb R = 4.0 inches	67.5	Pass	Pass
516	5th Female	Max G = 16.25 V (ft/s) = 25.0	W = 5.5 lb R = 4.5 inches	69.5	Pass	Pass
517	5th Female	Max G = 16.25 V (ft/s) = 25.0	W = 6.0 lb R = 5.0 inches	66.9	Pass	Pass
530	50th Male	Max G = 16.25 V (ft/s) = 25.0	W = 4.0 lb R = 3.0 inches	67.7	Pass	Pass
531	50th Male	Max G = 16.25 V (ft/s) = 25.0	W = 6.0 lb R = 5.0 inches	67.9	Pass	Pass
512	95th Male	Max G = 16.25 V (ft/s) = 25.0	W = 4.0 lb R = 3.0 inches	70.6	Pass	Pass
513	95th Male	Max G = 16.25 V (ft/s) = 25.0	W = 6.0 lb R = 5.0 inches	70.7	Pass	Pass
Medium Severity Pulse						
577	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 3.5 lb R = 2.5 inches	67.5	Pass	Pass
578	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 4.0 lb R = 3.0 inches	70.1	Fail	Pass
579	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 4.5 lb R = 3.0 inches	68.7	Fail	Pass
518	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 4.0 lb R = 3.0 inches	70.1	Marginal	Pass
519	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 5.0 lb R = 4.0 inches	67.5	Fail	Marginal
520	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 5.0 lb R = 5.0 inches	72.2	Marginal	Pass
521	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 5.5 lb R = 4.0 inches	69.5	Fail	Marginal
522	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 5.5 lb R = 5.0 inches	69.6	Fail	Fail
523	5th Female	Max G = 18.5 V (ft/s) = 31.5	W = 6.0 lb R = 5.0 inches	66.9	Fail	Pass
524	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 4.0 lb R = 3.0 inches	67.7	Pass	Pass
525	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 4.5 lb R = 4.0 inches	67	Pass	Pass
526	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 5.0 lb R = 4.0 inches	72.2	Pass	Pass
527	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 5.0 lb R = 5.0 inches	66.9	Pass	Pass
529	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 6.0 lb R = 5.0 inches	67.9	Pass	Pass

Test #	ATD (Percentile)	Test Pulse Configuration	HSM (weight and CM Location)	Theta (deg)	Nij Results	Mertz Criteria Results
Medium Severity Pulse, Continued						
590	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 4.0 lb R = 3.0 inches	67.7	Pass	Pass
591	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 5.0 lb R = 4.0 inches	72.2	Pass	Pass
592	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 5.5 lb R = 5.0 inches	69.8	Pass	Pass
593	50th Male	Max G = 18.5 V (ft/s) = 31.5	W = 6.0 lb R = 5.0 inches	67.9	Pass	Pass
509	95th Male	Max G = 18.5 V (ft/s) = 31.5	W = 4.0 lb R = 3.0 inches	70.6	Pass	Pass
510	95th Male	Max G = 18.5 V (ft/s) = 31.5	W = 5.5 lb R = 4.0 inches	70.8	Pass	Pass
511	95th Male	Max g = 18.5 V (ft/s) = 31.5	W = 6.0 lb R = 5.0 inches	70.7	Pass	Pass
High Severity Pulse						
537	5th Female	Max G = 19.0 V (ft/s) = 50.0	W = 3.5 lb R = 2.5 inches	67.5	Fail	Fail
538	5th Female	Max G = 19.0 V (ft/s) = 50.0	W = 4.0 lb R = 3.0 inches	70.1	Fail	Fail
539	5th Female	Max G = 19.0 V (ft/s) = 50.0	W = 2.8 lb R = 2.25 inches	77.5	Fail	Fail
540	5th Female	Max G = 19.0 V (ft/s) = 50.0	W = 2.8 lb R = 2.25 inches	77.5	Fail	Fail
586	50th Male	Max G = 19.0 V (ft/s) = 50.0	W = 2.8 lb R = 2.25 inches	70.6	Fail	Fail
587	50th Male	Max G = 19.0 V (ft/s) = 50.0	W = 3.5 lb R = 2.5 inches	66.7	Fail	Fail
588	50th Male	Max G = 19.0 V (ft/s) = 50.0	W = 4.0 lb R = 3.0 inches	67.7	Fail	Fail
589	50th Male	Max G = 19.0 V (ft/s) = 50.0	W = 4.5 lb R = 4.0 inches	67	Fail	Fail
580	95th Male	Max G = 19.0 V (ft/s) = 50.0	W = 2.8 lb R = 2.25 inches	71.4	Pass	Pass
581	95th Male	Max G = 19.0 V (ft/s) = 50.0	W = 3.5 lb R = 2.5 inches	72.9	Fail	Pass
582	95th Male	Max G = 19.0 V (ft/s) = 50.0	W = 4.0 lb R = 3.0 inches	70.6	Fail	Marginal
583	95th Male	Max G = 19.0 V (ft/s) = 50.0	W = 5.0 lb R = 4.0 inches	72.9	Fail	Pass
584	95th Male	Max G = 19.0 V (ft/s) = 50.0	W = 5.5 lb R = 4.0 inches	70.8	Fail	Fail
585	95th Male	Max G = 19.0 V (ft/s) = 50.0	W = 6.0 lb R = 5.0 inches	70.7	Fail	Pass

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Merkle, A. C., Kleinberger, M., & Uy, O. M. (2005). The effects of head-supported mass on the risk of neck injury in army personnel. *Johns Hopkins APL Technical Digest, 26*(1), 75-83.

Johns Hopkins Applied Physics Laboratory conducted simulated crash testing using a Hybrid III ATD (Merkle et al., 2005). The test set-up was designed to mimic a UH-60 seating configuration, with the seat back reclined 20º from vertical and the seat pan pitched 10º upward. A horizontal sled was used to translate the ATD in the anterior direction and then brake for an applied decelerative load resulting in a primarily -Gx loading vector (Figure 5).

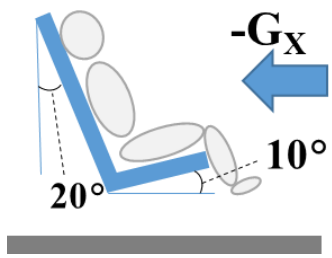

Figure 5. Seating orientation for experimental and computational tests conducted by Merkle et al. (2005).

Several factors were investigated and thus two separate test series were incorporated into this study. Both test series collected peak upper neck forces and moments that were then used with the peak neck force and moment limits and the Nij neck injury criteria to determine acute upper neck injury risk. The first set of experiments followed a Taguchi Screening Test Series[*] to investigate the main factors of impact velocity, helmet weight, mass offset in the x (CMx) and z (CMz) axes with respect to the occipital condyles (Table 2). Through the screening tests, it was determined that an impact velocity of 40 ft/s (12.2 m/s) with a 26 G deceleration, was required to produce a sufficient risk of upper neck injury based on the Nij. The second test series was conducted with the constant impact velocity of 40 ft/s (12.2 m/s) determined in the first series, and assessed three main factors: helmet weight, CMx offset, and CMz offset. A Box-Behnken design[†] was used to statistically test the three main factor effects (Table 3).

Table 2. Taguchi Screening Test Series Limits for Screened Variables

	Low Limit of Screened Variable		**High Limit of Screened Variable**	
Impact Velocity	19.8 ft/s	6.04 m/s	39.3 ft/s	11.98 m/s
Helmet Weight	3.5 lb	1.59 kg	7.5 lb	3.40 kg
CMx	-0.5 in	-1.27 cm	3.5 in	8.89 cm
CMz	1.0 in	2.54 cm	5.0 in	12.7 cm

[*] *The Taguchi Screening Test is a highly fractionated factorial designed used to determine main factor effects with limited number of test runs.*

[†] *The Box-Behnken Design is an experimental design that combines a two level factorial design with an incomplete block*

Table 3. All Test Conditions for the Box-Behnken Experimental Design (Recreated from Merkle et al. [2005])

Test ID	Impact Velocity (ft/s)	Helmet Weight (lb)	CMx (in.)	CMz (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

Results indicate that helmet weight was a statistically significant factor that influences neck shear force, neck tension force, extension moment, Nij tension-flexion, and Nij tension-extension. Horizontal CM position was found to be a statistically significant factor for extension moment, flexion moment, Nij tension-flexion, and Nij tension-extension. Vertical CM position was found to significantly affect extension moment, Nij tension-flexion, and Nij tension-extension.

The injury assessment risk value (IARV) of 503 lb-in (57 Nm) defined in Eppinger et al. (1999) as the critical intercept for extension was exceeded in all but one condition for the Box-Behnkin test series. The lowest helmet weight and smallest CMx offset (Test ID AHSM22) was the only test that did not exceed the extension moment IARV for the Box-Behnkin test series. No test conditions exceeded the IARV for flexion moment or Nij tension-flexion value. All but one test exceeded the IARV for extension moment, and 12 of the 16 tests exceed the Nij tension-extension value. These results suggest that there is a higher likelihood of sustaining a tension-extension injury than a tension-flexion injury in a frontal crash. Furthermore, these results provide insight on the complexity of the increased injury risk problem from HSM. The authors noted, that if HSM is shifted to the posterior-superior portion of the head in an effort to reduce neck extension injuries, an increase in injury risk from flexion moment would occur.

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Bass C.R., Donnellan, L., Salzar, R., Lucas, S., Folk, B., Davis, M., Rafaels, K., Planchak, C., Meyerhoff, K., Ziemba, A., & Alem, N. (2006). A new neck injury criterion in combined vertical/frontal crashes with head supported mass. *International Research Council on Biomechanics of Injury Proceedings*, Madrid, Spain.

The objective of study performed by Bass, et al. (2006) was to develop a new injury risk function for the lower neck with varying HSM and center of mass locations. There were three phases of testing, which included whole body post mortem human subjects (PMHS), ATD, and PMHS head and neck components. Both the Hybrid III and the Test Device for Human Occupant Restrain (THOR) ATDs were included in this study. The first two phases of testing consisted of sled tests using six whole-body PMHS, the Hybrid III ATD, and the THOR ATD. The whole-body PMHS were tested in three different head mass conditions: unweighted, 1.7 kg, and 2 kg. Whole-body surrogates were oriented in a recumbent position with the seatback on the horizontal sled carriage rotated to a 30º angle from the impact vector for each test (Figure 6A). Based on USAARL's recommendation, a triangular deceleration profile pulse of 30 G was used. Matched pair testing was conducted with the whole-body ATDs.

The final phase of testing included head and neck subsystem. The head and neck systems were comprised of the head to below the T4 vertebral body. Specimens were potted using screws in the T3 and T4 vertebral bodies before being oriented within the potting fixture with their normal kyphosis using an epoxy compound filling the fixture. The specimen was suspended upside down on the sled apparatus and a pneumatic impactor applied the pulse in the sagittal plane (Figure 66B). The angle of the neck was varied from -45º off of horizontal to 45º. Specimens were tested with multiple HSM conditions including masses between 2 and 4 kg and vertical CM offset between 0 and 11.8 cm relative to the tragion notch. Viscoelastic padding or a hydraulic decelerator was used to moderate the impact. The first two series of component impacts were short duration, high magnitude impacts to maximize shear force in the neck; the remaining tests used a longer duration and smaller magnitude pulse. Maximum sled velocity ranged from 1.7 m/s to 5.4 m/s. Matched pair ATD testing was conducted with the HIII and THOR for the head and neck complexes as well.

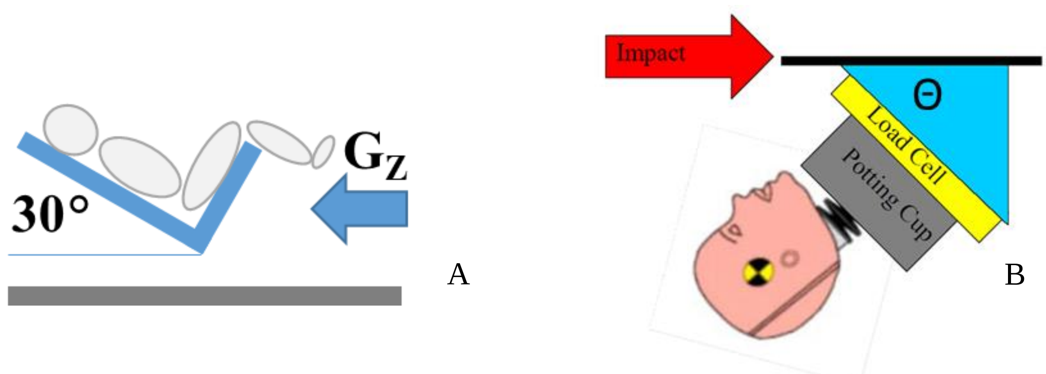

Figure 6. Illustration recreated from Bass et al. (2006) showing (A) the whole body test configuration and (B) the head/neck complex sled test configuration (Θ ranges from -45º to 45º off of horizontal axis).

The data collected through the component tests were used to develop injury criteria while the whole-body tests were used to confirm injury locations (versus published field injury locations in Shanahan and Shanahan et al., 1989), as well as neck kinematics to inform component tests. Frequent injuries documented in the whole-body PMHS included ligament laxity, tears, and transactions. Other injuries involved discs and end plates. Most injuries in component testing were posterior ligaments located between C5 and T2. Laxity, transaction, ligament tears, disc tears, and minor vertebral crushing were all documented.

The injury criterion was developed by modeling the lower cervical spine as a beam (similar to the theory for Nij) and expressed by Eq. 4.

$$BC = \frac{F_z}{F_{zcrit}} + \frac{M_y}{M_{ycrit}} \qquad \text{Eq. 4}$$

Where F_z represents the axial compression/tension force; M_y represents the flexion/extension moment at the C7/T1 intervertebral disc; F_{Zcrit} is a derived constant and was set to 5660 N for axial tension and 5430 N for axial compression; and M_{ycrit} is a derived constant set to 141 N-m for both flexion and extension. The critical values for F_z and M_y were optimized from the motor vehicle injury standards set forth in FMVSS-208 (2003). The critical values (F_{Zcrit} and M_{ycrit}) were specified such that a result of 1.0 from the beam criterion (BC) defined in Eq. 4 corresponds to a 50% risk of Abbreviated Injury Score (AIS) 2+ injury to the human cervical spine.

The authors note that while the shear motion was not a significant factor in the development of the injury risk curve for the BC, it does play a significant role in the PMHS neck kinematics and may influence the resulting bending stiffness. The THOR ATD was stated to be more biofidelic than the Hybrid III; however, it was not able to represent the shear motion joint stop of the cadaveric dynamics. Furthermore, the authors note that through this study the Nij criteria evaluated at the upper neck is not adequate for assessing neck injury with HSM under inertial loading. The authors provide two considerations for why Nij is inappropriate: (1) the Nij criteria are based on the Hybrid III dummy response, which is generally not considered biofidelic for simulating human motion under purely inertial flexion/extension loading and this limitation is exacerbated with HSM included; and (2) upper neck flexion moments are generally lower than lower neck flexion moments, emphasizing the role of the moment arm of the head and HSM system in the dummy response. Finally, the authors note that additional research using computational modeling to investigate more complex loading conditions is required to understand the relative role of the shear forces increasing the stiffness of the neck as well as the combination of lower neck and upper neck loads.

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Manoogian, S. J., Kennedy, E. A., Wilson, K. A., Duma, S. M., & Alem, N. M. (2006). Predicting neck injuries due to head-supported mass. *Aviation, Space, and Environmental Medicine, 77*(5), 509-514.

The effect of HSM on neck injury risk was predicted using the validated The Netherlands Organization (TNO) MADYMO computational model (Meijer et al., 2003). The model was designed to simulate a 50th percentile male subjected to the accelerative loading magnitudes of 5 G, 13.5 G, and 22 G accelerations in seven different loading directions (Figure 7) while varying the HSM from 0 to 3 kg in 1 kg increments. Additionally, the CMx offset ranged from 0.7 cm to 8.7 cm and the CMz offset ranged from 2.3 cm to 10.3 cm (Note: CM offset was defined relative to the occipital condyle for all tests). In order to simulate all of the conditions, a total of 196 iterations were performed. The force and moment data from the simulations were used to calculate the lower neck beam criterion and assess injury risk.

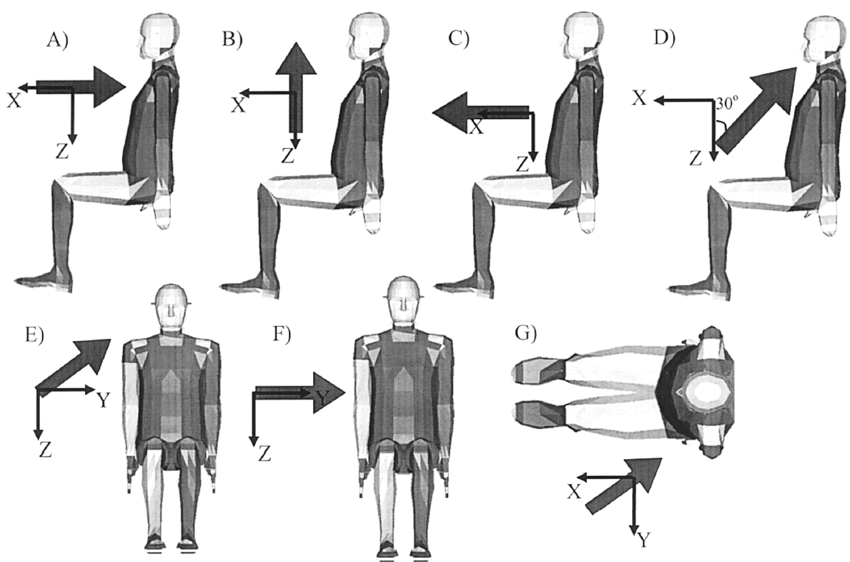

Figure 7. Seven loading directions, A-G, that were modeled in Manoogian et al. (2006) (Figure extracted from Manoogian et al. [2006]).

The simulation results determined that impact magnitude was the primary factor that influenced injury risk, and maximum allowable HSM was dependent on impact acceleration magnitude. The results indicated that for the highest severity tests (22 G), the inclusion of HSM mass and mass offset had a greater effect on the risk of injury above the baseline injury risk (i.e., no HSM and no offset) than for the lower severity tests. However, the discussion further explained that due to limitations, no further conclusions were able to be provided. The authors identify several limitations of the study. In particular, the inclusion of only one sex and size model, fixing the body below T3, and not including any additional constraints (e.g., headrests) to the model were discussed.

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Parr, M. J. C., Miller, M. E., Bridges, N. R., Buhrman, J. R., Perry, C. E., & Wright, N. L. (2012). Evaluation of the Nij neck injury criteria with human response data for use in future research on helmet mounted display mass properties. *Proceedings of the Human Factors and Ergonomics Society Annual Meeting* 56(1), 2070-74).

Parr et al. (2012) performed a preliminary study to assess the use of the Nij injury criteria with previously collected data from volunteer human subjects exposed to non-injurious sled tests to simulate an aviation environment. The focus of the study was to develop the methodology and determine if a new neck injury criteria, similar to the Nij, using human data, could be developed for the aviation environment. Data included -Gx impacts from 23 human subjects wearing a 4.50 lb (2.04 kg) helmet-mounted display while undergoing horizontal sled tests. Horizontal sled tests were designed to mimic the seat deceleration and parachute opening shock phases of a fighter jet pilot ejection. Subjects were seated vertically in standard aircraft seats and were accelerated in the -Gx direction at an acceleration of 8 G for approximately 200 ms. All tests were designed to be non-injurious; however, 20% of the participants reported neck pain. Experimental methods, design, and results were modeled after those reported in Doczy et al., (2004). Neck loads for all trials and Nij values were computed from the resultant head, neck, and body accelerations using a custom software developed by the U.S. Air Force. The authors note that the software includes a correction factor for the bending moments used in ATD tests and that this correction may have overestimated the bending loads in the human data as it would not be required.

All calculated Nij values were below the threshold injury value of 1.0 defined by Eppinger, et al. (1999). Furthermore, all calculated Nij values were below the proposed Aviation cutoff of 0.5 (as proposed by Nichols [2006]). The authors note that the lower value of 0.5 is necessary as pilots and aviators may have to escape and evade following a crash event and the lower threshold value decreases the likelihood of an incapacitating injury. The results from this study suggest that the addition of a helmet under the ejection seat pulse does not substantially increase neck injury risk. However, a substantial amount of variability in the Nij response was identified in the paper. The authors note that the variability in the data presented may be due to several sources, including: sex, neck strength, head size, neck length, and body weight. Furthermore, understanding the relative contributions of each of the above factors would require additional data and statistical analyses. Finally, the authors make a comment that while the Nij provides several advantages for predicting neck injury, additional work is required to develop guidelines to be used in the development of helmets and helmet mounted systems.

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Gaur, S. J., Balakrishna, A., & Aravind, A. S. (2013). Determination of helmet CG (CM) and evaluation of neck injury potentials using "Knox Box criteria" and neck torque limits. *Indian Journal of Aerospace Medicine*, *57*(1).

The study from Gaur et al. (2013) investigated the potential change in injury risk due to the shifts of CM offset and increased mass from wearing a helmet. Five different helmet conditions were examined, including: (1) Mirage (visor stowed), (2) Su-30 (visor stowed), (3) Mig21 (visor stowed), (4) Gallet (no additional equipment), (5) Gallet with NVG and counter weight. A Trifilar pendulum mass property instrument (M/S DEFSYS Pvt Ltd. [Bangalore, India]) was used to derive the helmet mass and CM position relative to a custom ATD head for all five HSM conditions. The ATD head was a custom design made out of wood and aluminum to reflect a 50th percentile Indian male aviator. CM measurements were converted into coordinates along the X- and Z-axis about the reference load cell. The masses of the tested helmets included 1.30 kg, 1.32 kg, 1.61 kg, 1.73 kg, and 2.54 kg and the CM offset ranged from -0.29 to 0.85 cm in the x-axis and 2.45 to 3.22 cm in the z-axis. A Knox Box, which is a function of helmet mass and acceleration level, was overlaid on the CM plot to assess neck injury risk (Gaur et al., 2013). If the Knox Box boundary for a particular acceleration level is exceeded, the helmet is considered outside acceptable limits. CM in the x and z axes was plotted for all five helmet conditions as well as the ATD without a helmet for a reference point. All conditions fell within the Knox Box boundary. The study suggests that the injury risk in ejection with these helmets is negligible based on the Knox Box criteria. Furthermore, the moments were calculated and compared to the USAARL HSM Curves and found that the Gallet assembly, condition 5, exceeded the 29.8 kg-cm limit. The authors did not explain any potential limitations to their study.

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Jadischke, R., Viano, D. C., McCarthy, J., & King, A. I. (2016). The effects of helmet weight on hybrid III head and neck responses by comparing unhelmeted and helmeted impacts. *Journal of Biomechanical Engineering, 138*(10).

The study by Jadischke et al. (2016) investigated the effects of helmet mass on head and upper neck response from helmet to helmet contact. The goal was to study helmeted and unhelmeted impacts that were similar to what is observed in American football. National Football League film of concussive impacts to the head were observed to determine the nine most frequently impacted locations, which were then used in the study (Figure 8B). Two series of head impact experiments using a Hybrid III ATD were conducted. The first series involved a rigid arm pendulum simulating a striking and a struck player (Figure 8A). Tests included a no-helmet condition and three helmeted conditions for the struck player (mounted to a sled). The three helmet weight conditions included the shell and padding only (1.26 kg), the shell and padding plus a facemask (1.92 kg), and the helmet shell with facemask and 350 grams (g) of lead tape (2.27 kg total). The pendulum (striking helmet) was used to produce the impact with speeds of 4.1 m/s and 5.2 m/s. The same helmet (Riddell Revolution) was used for all tests (helmets were inspected for damage between tests). The test protocol was developed to simulate helmet on helmet collisions.

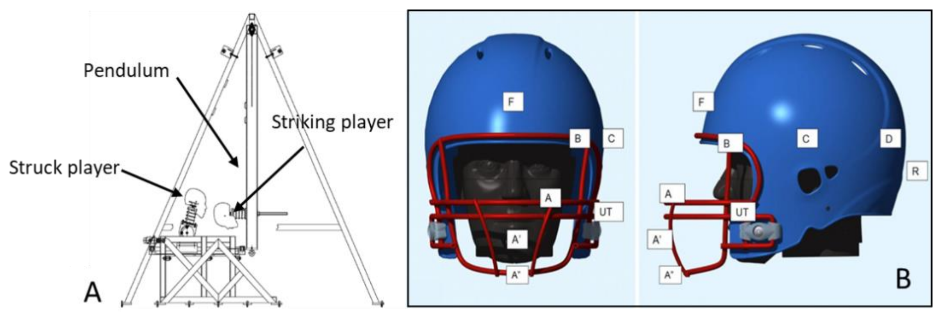

Figure 8. A combined figure from Jadischke et al. (2016) illustrating the (A) test set-up and (B) impact loading locations.

Data were taken from a second series of previously published tests (Viano et al., 2012) that involved impacts with a linear impactor to 17 different makes and models of American football helmets, as well as an unhelmeted condition. Helmeted impact tests were conducted at nominal speeds of 5.5, 7.4, 9.3, and 11.2 m/s. Unhelmeted tests were conducted at speeds of 7.4 and 9.3 m/s. Tests consisted of the helmets being placed on Hybrid III head and neck complex on a linear slide table and impacted with a Biokinetics linear impactor (Biokinetics, Ottawa, ON). Peak headform linear acceleration, linear momentum, delta-V, and peak upper neck forces and moments were recorded for each impact location, condition, and speed. Data were reported for 14 helmets (mass range from 1.5 kg to 2.19 kg) and the unhelmeted test condition with overlapping data and exposure conditions.

The results showed the helmet conditions reduced resultant head acceleration. Additionally, the results showed that increased mass was associated with an increase in the momentum of the headform, along with increases in upper neck forces (especially neck tension) and moments. Findings from this study were not used to calculate neck injury risk and discussion was focused on the role of neck loading in concussion injury risk.

Discussion

A more comprehensive understanding of how increasing HSM mass and shifts in HSM CM offset affect the risk for acute injury has yet to be well established. The current guidelines to limit acute injuries were based on the U.S. Army aviation experience (i.e., the USAARL Acute Injury Curve) and have not been updated since 1997. In the years since publication, several new neck injury criteria and metrics have been developed, as well as advancements in use of statistical methods for estimating injury risk given several factors. The lack of updated guidelines could be due to the lack of research performed in this area. This report reviewed the relevant, published, research from 1997 to present that provides insight on the relationship between HSM and acute injury risk. A summary of the findings from the reviewed literature and a comparison of the published results to the USAARL HSM Acute Injury Curve can be found in Appendix 1.

Three of the ten studies reviewed (Perry et al., 2003; Doczy et al., 2004; Parr et al., 2012) used living human subjects in non-injurious loading scenarios to investigate the effect of HSM on the human loading response. While using non-injurious loading conditions, well below those used in the USAARL HSM Acute Injury Curve, the three studies all showed HSM to be a factor that influenced neck loading response. The study by Parr et al. (2012) was the only study using volunteer human subjects to assess neck injury risk with existing acute neck injury criteria (Nij). The estimated neck loads were found to be below the Nij threshold value, which was expected as the study involved volunteer human subjects and no injuries were reported during the accelerative loading experiments. Perry et al. (2003) were the only group to evaluate accelerative loading in the lateral (i.e., Gy) direction using human volunteers. No injury criteria were used to evaluate the acute injury risk potential of the testing conditions; however, results were used to extrapolate the neck loading response to a 10 G lateral impact with a 1.4 kg HSM. Authors noted that even the extrapolated results were below a proposed neck injury threshold derived from cadaver testing in the -Gx direction by Mertz and Patrick (1971), but did not compare to any lateral neck injury criteria. Perry et al. (2003) went on to note that these results are not comparative due to the different loading directions. Doczy et al. (2004) conducted a similar study to Perry et al. (2003), but used frontal impact (i.e., -Gx) loading direction. Similar to Perry et al. (2003), acute neck injury criteria were not used to assess the volunteer human subject results. Instead, volunteer human subject results were used to develop a computational model to extrapolate estimated neck loads from higher severity tests. The extrapolated results found by Doczy et al. (2004) were then compared to the proposed injury threshold, X-direction moment 1680 in-lb (190 N-m), set by Mertz and Patrick (1971) and found to fall below the limit. Although the volunteer studies established that HSM influences neck loading, the relationship was not clearly defined and the studies do not offer sufficient information to develop updated guidelines for acute injury risk. Furthermore, the degree of error in the calculated neck loads and moments used in the studies by both Parr et al. (2012) and Doczy et al. (2004) is unknown. Both studies state that the kinematics (accelerations and angular rates), anthropometry, and inertial properties were used to calculate neck loads; however, not enough detail is provided for these methods to determine the accuracy of the estimates.

Several studies reviewed used mechanical surrogates to estimate the risk of injury at higher severity conditions. Paskoff and Sieveka (2004), Merkle et al. (2005), and Jadischke et al. (2016) all established that changes in CM and mass influence neck forces and moments using the Hybrid III ATD. Jadischke et al. (2016) investigated both changes in helmet weight and CM offset, but in a helmet to helmet contact loading scenario, and thus the test procedure may not be representative of accelerative loading experienced by Soldiers in an operational environment, particularly for well-restrained personnel in aircraft or vehicle crashes. While test methods from Jadischke et al. (2016) involved direct impact, the study shows similar results to the other studies reviewed with an increase in neck loads corresponding with an increase in helmet mass. Both Paskoff and Sieveka (2004) and Merkle et al. (2005) designed test protocols to investigate typical loading profiles that U.S. Army aviators could encounter with the specific intent to investigate changes in HSM characteristics. Both studies used the Nij to determine acute injury risk from the experiments that assess upper neck injury. Both studies showed that increased mass and offset increased the risk of injury with several test conditions exceeding the injury threshold value for Nij. While important for understanding the relationship between HSM conditions and acute injury risk, the studies were only able to estimate upper neck injury with the Nij criteria as no lower neck criteria existed for use at the time of both studies. Lower neck injuries have been previously linked to results from accelerative loading involving HSM (Bass et al., 2006). Additional studies connecting ATD results with PMHS testing with HSM are needed to fully elucidate the influence of HSM on acute neck injury risk.

Only one study was identified that developed an acute lower neck injury criteria that incorporates HSM, the Beam Criteria (Bass et al., 2006). Both PMHS and ATD experiments were conducted to assess the injury potential of accelerative loading with varied helmet mass and CM offset conditions. Although this is the only study found to incorporate HSM in the development of acute injury criteria, the effects of mass and CM offset on neck loading were not fully defined, and thus recommend HSM guidelines were not provided. Further research is required to define relationships between CM offset, mass, and exposure (direction and severity) during acute neck injury criteria development.

Overall, there are several major gaps in the literature regarding the contribution of HSM to acute neck injury risk. A significant gap identified was the paucity of PMHS and finite element model research in this area. While several studies estimated neck loads through computational modeling using the MADYMO models and software, no studies using more advanced finite element methods were identified. With the advancement of computational processing power and improved finite element models of the neck, future research should investigate the feasibility of estimating neck injury and internal stresses and strains using finite element modeling methods versus rigid body modeling methods. Furthermore, one study that assessed the validity of using existing acute neck injury risk criteria (Nij) noted several concerns over the reliability of the Nij for use in evaluating HSM due to the HSM effects on neck kinematics and ATD biofidelity (Bass et al., 2006). Finally, there has been little research aimed specifically at developing HSM guidelines, but rather HSM effects are assessed with existing criteria, which may not be appropriate for use.

Conclusions

Our review has shown that while several efforts have been directed towards investigating HSM effects on acute neck injury in the last 25 years, little progress has resulted towards improved guidance on HSM acute injury risk. Several volunteer studies have been performed, and while estimated neck loads were below injury thresholds, studies reported that multiple subjects developed pain. However, these studies did not provide sufficient data for the development of updated guidance on defining HSM limits. Additionally, multiple studies were conducted with ATDs that were not designed for HSM wear or evaluation and over a wide range of dynamic loading conditions. These studies required innovative adaptations to experimentally investigate HSM effects on human biomechanical response. While the ATD studies confirmed a relationship between HSM conditions (mass and CM offset) and potential increases in acute injury risk, the studies focused on the upper neck injury risk. The lack of any lower neck ATD injury assessment reference criteria at the time prevented the reviewed ATD and computational modeling studies from investigating injury risk to the lower neck, which is a region of concern for Service Members wearing HSM. Furthermore, all existing acute cervical injury criteria, except one, were not developed for evaluation of HSM. Only Bass et al. (2006) included HSM influence in the development of an acute cervical injury criterion, but the resulting lower neck injury criterion did not provide direct guidance on HSM (mass and CM offset) limits. After review of the limited available studies and with consideration of their respective limitations, the HSM guidance from McEntire and Shanahan (1997), while fundamental in its development, remains the most relevant and only biomechanically-based guidance for acute neck injury risk due to HSM (mass and CM offset). More research is needed to develop lower neck injury criteria and improve ATD cervical spine biofidelity. This research would lead to improved HSM guidance for future helmet and helmet mounted device development efforts.

A multifaceted research approach to update HSM guidance that accurately reflects neck injury risk for U.S. Army aviators and ground Soldiers is anticipated to resolve many of the identified gaps. In particular, PMHS testing using loading profiles that are representative of aircraft and ground vehicle mishaps with HSM are needed to establish threshold limit values. These results then need to be compared to matched pair ATD testing and finite element simulations to assess the biofidelity of the ATD and finite element models and to create appropriate scaling ratios or Injury Assessment Reference Curves to assess future HSM conditions. Such a multifaceted approach will provide significant advancement to understanding HSM acute neck injury risk and progress towards developing a comprehensive guidance for assessing the acute injury risk of newly developed helmets and helmet systems. Ultimately, this research would improve our Service Members' protection from acute cervical neck injury.

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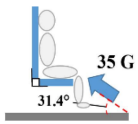
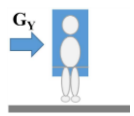
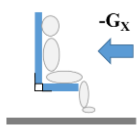
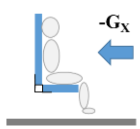
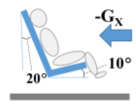
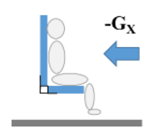
Appendix A. Article Summaries

Table A1. Summary of Experimental Procedures and Results Compared to the USAARL Curves from Literary Sources

Author	Year	Surrogate	HSM Mass	HSM CM offset	Exposures	Exposure Direction	Criteria Used	Article Findings	USAARL HSM Curve Results
McEntire et al.	1997	Mathematically derived	Up to 2.5 kg	Up to 5.2 cm from the tragion notch	Estimated 35 G crash pulse		Neck Tension (Mertz et al. 1993)	Provides guidelines for acceptable HSM limits that have been used for the past 25 years	NA
Perry et al.	2003	Human Volunteers	0.0 kg to 2.0 kg	Not Reported	4 G to 6 G	 Seat pan tilted 13° from horizontal, seat back perpendicular to seat pan	None	Calculated neck loads and moments exceeded safe guidelines of 90 lbf from Weiss et al., (1989); however, all loadings were tolerated by subjects.	• All conditions were below USAARL 2.5 kg threshold. • No CM offset was measured and thus results cannot be evaluated further
Doczy et al.	2004	Human Volunteers	0.91 kg to 2.04 kg	Not Reported	6 G to 8 G, 10 G		None	Conditions were not evaluated for acute injury risk, but 3 subjects reported pain.	• All helmet conditions were below the 2.5 kg weight limit, • CM offset for each condition was not calculated
		Experimentally Derived Computational Model	0.91 kg to 2.04 kg	Not Reported	11 G to 15 G		Compared to PMHS study	Neck loads found to be below suggested threshold for max mass and acceleration condition	• All helmet conditions were below the 2.5 kg weight limit, • CM offset for each conditions were not calculated

Author	Year	Surrogate	HSM Mass	HSM CM offset	Exposures	Exposure Direction	Criteria Used	Article Findings	USAARL HSM Curve Results
Paskoff and Sieveka	2004	MADYMO Computational model *HSM mass, CM offset, and exposures conditions are reported for the simulations compared with experimental data	1.8 kg	Polar representation of CM offset with radius between 6.1 cm and 6.5 cm and angle between 68.6 and 72.2 degrees of horizontal * CM offset is defined relative to the Occipital Condyles	16.25 G 18.5 G 19 G		Nij	Model mainly used to defne experimental test pulse. One condition (1.8 kg) compared with experimental test results and found model either (1) under predicts peak values (in low/medium severity cases) or (2) timing of peak values occurred earlier for the high severity cases	NA
		Hybrid III *Note HSM mass and CM offset are reported as target values	1.3 kg to 2.7 kg	Polar representation of CM offset with radius between 5.7 cm and 12.7 cm and angle between 66.7 and 77.5 degrees of horizontal * CM offset is defined relative to the Occipital Condyles	16.25 G 18.5 G 19 G		Mertz Criteria Nij	All low severity conditions passed Nij; All mid-male and large male medium pulse severity conditions passed Nij	• Tests included mass conditions above and below USAARL limit of 2.5 kg. • CM offset was not compared as article reported distance from Occiptial Condyles using a simulated headform

Author	Year	Surrogate	HSM Mass	HSM CM offset	Exposures	Exposure Direction	Criteria Used	Article Findings	USAARL HSM Curve Results
Merkle et al.	2005	Hybrid III	1.3 kg to 3.7 kg	CGx offset: -2.8 to 10.4 cm; CGz offset: 1.0 to 14.2 cm * CM offset is defined relative to the Occipital Condyles	Peak 26 G		Peak Upper Neck Forces and Moment Nij	Mass, CGx offset, and CGz offset all had correlation with neck injury measures, but the relationship was complex.	• All mass conditions at 2.5 kgs and above fail the USAARL guidelines, • CM offset was not comparible in this study as its based of the Occipital Condyles
Bass et al.	2006	PMHS	0.0 kg to 2.0 kg	0	17 G to 33 G		Beam Criteria	Injuries aligned with Shanahan and Shanahan (1989) with the majority occurring in the lower neck. Injuries were primarily to the ligaments and discs.	• All tested conditions were within the USAARL guidelines; however, injuries were noted.
		Hybrid III THOR	0.0 kg to 2.0 kg	0	17 G to 33 G		Nij	Illustrated the biofidelity concerns for both the Hybrid III and THOR necks when considering inertial loading resulting in primarily flexion and extension.	• All tested conditions were within the USAARL guidelines;

Author	Year	Surrogate	HSM Mass	HSM CM offset	Exposures	Exposure Direction	Criteria Used	Article Findings	USAARL HSM Curve Results
Bass et al. (Cont.)	2006	PMHS Head/ Neck Component	2.0 kg to 4.0 kg	CGz offset: 0 cm to 11.8 cm * CM offset is defined relative to the tragion notch	1.7 to 5.4 m/s	Varied: See Figure 6	Beam Criteria	21 of 36 tests received injury with MAIS scores ranging from 1 to 6. Data were used to derive the Beam Criteria.	• All conditions with added mass of 2 kg where the CMz offset was 2.8 cm or below passed the USAARL guidelines • Remaining conditions failed the guidelines • Injuries occurred for HSM within and outside USAARL guidelines
Manoogian et al.	2006	TNO MADYMO Computational Model	0 kg to 3.0 kg	CMx offset: 0.7 cm to 8.7 cm; CMz offset: 2.3 cm to 10.3cm * CM offset is defined relative to the occipital condyle	5 G 13.5 G 22 G	Varied: See Figure 7	Beam Criteria	Impact magnitude and direction more than HSM affect injury risk. At higher acceleration, mass and CM offset have greater influence on risk.	• All mass conditions above 2.5 kgs fail the USAARL guidelines, • CM offset was not comparible in this study as its based of the Occiptal Condyles
Parr et al.	2012	Human Volunteers	2.04 kg	Not Reported	8 G		Nij	All results were below the Nij threshold value of 1.0 as well as recommended Aviation threshold of 0.5 (Nichols, 2006)	• Helmet condition was below the 2.5 kg weight limit; • CM offset was not specified

Author	Year	Surrogate	HSM Mass	HSM CM offset	Exposures	Exposure Direction	Criteria Used	Article Findings	USAARL HSM Curve Results
Gaur et al.	2013	Wood and Aluminum ATD	1.61 kg to 2.54 kg	CGx offset -0.29 cm to 0.85 cm; CGz offset 2.45 cm to 3.22 cm * CM Offset is defined relative to the ATD CG	None	N/A	Knox Box USAARL HSM Acute Curve	All conditions deemed safe	• One helmet condition (Gallet+NVG+counter weight) exceed vertical distance limit based on mass • Note:ATD CG position is unknown relative to standard ATDs (e.g., Hybrid III) as the study used a custom ATD for their experiemnts.
Jadischke et al.	2016	Hybrid III	0 kg 1.26 kg 1.92 kg 2.27 kg	Not Reported	4.1, 5.2, 5.5, 7.4, 9.3, and 11.2 m/s	Varied: See Figure 8	None	Provided no indication regarding the ability of each condition to prevent injury	• All helmet conditions were below the 2.5 kg weight limit; • CM offset for each conditions were not calculated

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