Computational Assessment of Headborne Equipment: Alteration of Head and Neck Biomechanics During Blast-Induced Accelerative Loading

M. Yates[1], M. Tumperi[1], E. Crane[1], G. Holt[1], L. Voo[1], V. Alphonse[1], D. Drewry[1], and Q. Luong[1]

[1] The Johns Hopkins University Applied Physics Laboratory, 11100 Johns Hopkins Road, Laurel, MD 20723, USA,
Melissa.Yates@jhuapl.edu

Abstract.
Headborne equipment (HBE), such as helmets, maxillofacial protection systems, and visual augmentation systems, alters the biomechanical state of the head and neck potentially influencing the inertial response and associated injury during exposure to blast overpressure. However, due to the complexity and substantial cost of experimentally recreating these events, parametric datasets covering a wide range of blast exposure and HBE combinations do not exist to evaluate injury outcomes. Human computational models can be applied to study the relative effect of changes in risk of injury based on the HBE combinations and assess if additional attention to risk of blast injury is necessary during HBE design. This work investigated the extent to which HBE may alter head and neck biomechanical response during blast-induced accelerative loading in order to help gain insight into the connection between blast loading and HBE combinations. A computational model of the head and neck was validated for kinematic response to impulse from blast overpressure exposure and a sensitivity study was conducted to evaluate the biomechanical effects due to headborne equipment selection, head orientation, and blast overpressure. The predicted biomechanical response data was then evaluated and then compared to established metrics and standards, the Head Injury Criterion (HIC), the Brain Injury Criterion (BrIC), and the Rotational Injury Criterion (RIC). These findings provide insight into the relative importance of assessing how current and future HBE systems influence biomechanics during a blast overpressure event, potentially informing equipment design as well as guidelines on HBE usage. These computational capabilities can provide insight for future testing and evaluation of HBE equipment prior to production of physical hardware. Further work is needed to experimentally capture these biomechanical effects in order to fully verify and validate the outcomes of this study.

1. INTRODUCTION

Combat helmet systems mitigate head injuries by absorbing the energy imparted by blunt impacts and arresting ballistic threats; however, current helmet performance assessments do not evaluate helmet performance for aspects related to blast exposure [1]. Blast exposure effects on the body are typically grouped into five distinct categories, with primary blast exposure induced by two biomechanical effects 1) overpressure loading and 2) inertial loading [3]. Prior efforts have investigated the effect of blast overpressure due to shock wave exposure on brain injury with a helmet system [4,5,6]. However, these studies focused on non-combat helmets, simplified pad systems, or did not evaluate the effect of changes in inertial loading on the head/neck system due to addition of headborne masses and geometries. The addition of helmets and other headborne equipment (HBE) change surface profile surrounding the head and in turn, results in an altered biomechanical state prior to threat exposure, as well as a change in impulse transferred to the head during the blast event. In order to understand if and how this altered biomechanical state affects injury risk, additional research is required. Characterization of the HBE performance during blast overpressure loading is necessary to understand if blast injury risk should be considered during helmet and HBE development, as well as to monitor the risks of augmenting personal protective equipment (PPE) to increase survivability; however, parametric datasets covering a wide range of blast exposure and HBE combinations do not exist to evaluate injury outcomes. *In silico* approaches can be applied to investigate blast-induced biomechanical alterations to the head/neck system and explore potential consequences in terms of injury. Numerous studies have developed numerical models to evaluate blunt injury risk for athletic and automotive applications [19], but a limited set of numerical models have been specifically developed to evaluate blast injury of the head and neck [2,20]. The purposes of this study are to (1) demonstrate a methodology that assesses how addition of HBE can influence the biomechanical response of the head and neck during blast-induced accelerative loading, and (2) evaluate results in the context of established head injury metrics to identify which metrics may warrant further study as potential tools for evaluation of HBE in the context of blast.

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2. METHODS

A human head and neck finite element model (FEM), developed by The Johns Hopkins Applied Physics Laboratory (JHU/APL) was previously validated for blast overpressure loading using published brain displacement and rotation data was leveraged for this study [7]. During this effort, this existing model was refined and incorporated with a current-service combat helmet, fielded HBE surfaces, and blast wave propagation through an open-air environment. This computational setup is summarized in Table 26. These component models were then assembled into the configurations outlined in Table 26 and verified for pressure transmission and kinematic response when loaded by shock waves at incident pressures of 200 kPa (2kg TNT at 2.5m) and 400 kPa (2.27kg TNT at 2.0 m). A simulation dataset for sensitivity analysis was then developed and conducted to evaluate the biomechanical effects due to HBE. Simulations were conducted in an explicit FEM solver based on its established use for spatial and temporal characterization of head and neck biomechanics, as well as capability to simulate blast exposure effects.

Table 26. Summary of computational modeling components and range of study.

Model Component	Description
Helmet	FE model of a representative combat helmet
Mounted Equipment	A representative Night Vision Goggle (NVG) system with mounting hardware
Blast Environment	Blast wave propagation through air developed in LS-DYNA with evaluated incident pressures at 200 and 400 kPa
Human Anatomy	FE model head and cervical spine validated for dynamic motion during blast exposure.

2.1 Model Development and Validation

All blast simulations were conducted using ANSYS LS-DYNA (Livermore Software Technology Corporation, Livermore, CA). A hybridized approach with both Lagrangian and Eulerian approaches capable of material flow and solid deformation, named the Arbitrary Lagrangian-Eulerian (ALE) method, was selected for its appropriateness for evaluation of blast overpressure effects on deformable structures. Briefly, the head and helmet were positioned and oriented with respect to prior experimental tests conducted at JHU/APL with the face oriented toward the oncoming shock wave [7]. For HBE incorporation, geometries were generated by 3D laser surface scans and computed tomography (CT) scans of hardware provided by Program Executive Office Soldier (PEO Soldier) Product Manager Soldier Protective Equipment (PdM SPE). The attaching bracket of the HBE was positioned to be centered on the midsagittal plane and in contact with the brim and shell of the helmet. The night vision goggle (NVG) system was aligned with the eye. If a gap existed between the mount and NVG after positioning, small adjustments were made to ensure an airtight structure during blast exposure.

The head-neck FE model previously developed and validated by JHU/APL for use in evaluating the effect of blast loading was selected for this study [1]. This head model was meshed in TrueGrid to reflect the 50th percentile male and based on the ANSUR II and Visible Human geometries. As shown in Figure 1, the model consists of representations of the brain (336,000 hexahedral elements), the cerebral falx/tentorium (4,600 2D shell elements), skull cortex (45,000 2D shell elements), diploë (143,000 hexahedral elements), frontal and maxillary sinuses (22,600 hexahedral elements), and flesh (297,000 hexahedral elements). In the cervical spine, the ligaments between the individual cervical spine vertebral bodies were modeled as 2D spring elements. The interface between the skull and neck was refined to incorporate previously published rotational stiffness response for the Occipital-C2 joint.

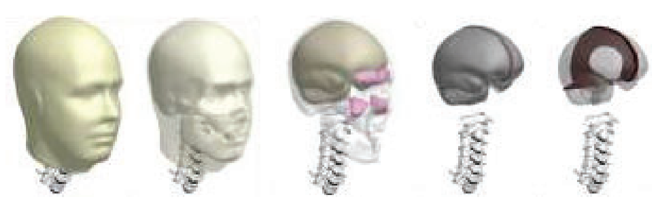

Figure 65. The head and neck FE model structures share nodal connections to enable pressure transmission. Key include flesh (tan), skull (white), vertebral bodies (white), intervertebral discs (black), cerebral spinal fluid (dark tan), sinuses (pink), brain (gray), cerebral falx/tentorium (red).

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Dynamic validation of the human head and cervical spine model was conducted to evaluate kinematic motion in blast loading by simulating the experimental setup of Iwaskiw et. al. [7] and comparing kinematic results. Briefly, a series of front-facing short-duration dynamic overpressure were conducted on four postmortem human surrogates (PMHS) (male, ages 53-67) to characterize dynamic head and neck motion. These specimen were disarticulated in-between the first and second thoracic vertebral bodies, then a fixed boundary condition in the lower cervical spine was created by potting at the 6th and 7th vertebral bodies using poly(methyl methacrylate). Specimen were perfused to achieve physiological intracranial pressure. Authors collected head-neck kinematics, brain motion, and intracranial pressure. The pressure loading applied to the head model was determined using a computational fluid dynamics (CFD) model of the head outer surface as a surface grid embedded in a structured grid at the end of a modeled shock tube as described in the experimental setup. The pressure outputs at the skin surface grid were then mapped to the FE mesh of the flesh. ALE simulations were then run with a fixed sixth cervical vertebra (C6) of the spine to emulate experimental potting at the same location. The error between the mean and the 1st standard deviation of the experimental corridor and simulation predictions were then assessed using CORA (CORrelation and Analysis). CORA has been established as an objective method to evaluate finite element model validation based on error in signal shape, magnitude, and phase [17].

A representative combat helmet FE model was furnished by PEO Soldier PdM SPE and validated in work previously reported directly to PEO Soldier PdM SPE which quantified effects during blast loading. The validation data included axial compression test data for the individual pads of the helmet system and whole helmet axial compression tests at three strain rates with three-dimensional digital image correlation to quantify load displacement response of pads. This helmet model consisted of deformable representations of the helmet shell with edge trim (38044 hexahedral elements), retention system clips (3747 hexahedral elements), retention system straps (848 2D shell elements), and a pad-based suspension system (5248 hexahedral elements). The HBE models were developed from a combination of physical night vision goggle (NVG) and mounting hardware, and supplemented by available computer-aided design (CAD) models. Geometries of a helmet and suspension system were reverse engineered using hand measurements and callipers for initial and approximate shape development with CREO Parametric (PTC). Computed Tomography (CT)scans were then obtained using the X50 (North Star Imaging, Inc.) to complete geometry capture. CREO Simulation (PTC) was used to generate a tetrahedral mesh for each component with a minimum element size of 2.0 millimetres (mm). The meshed NVG was assembled relative to the helmeted head FEM model shown in Figure 66.

Figure 66. APL head-neck model without a combat helmet, with a combat helmet, and with NVG.

ANSYS LS-DYNA can simulate the blast event based on the empirical model outlined in TM5-855 US Army Handbook (ConWep) coded in the Load_Blast_Enhanced keyword and against ConWep and blast pencil-probe data from live fire blast testing [9,10]. This approach models the explosive event, including the charge, the expansion of the detonated explosive, and the near-field physical effects in air. The resulting pressures were then quantified at a distance where near-field effects are not substantial based on experimental data integrated into LS-DYNA. The nonlinearities of the shock wave propagation through the air domain was modelled using a polytrophic equation of state to initialize at atmospheric pressure, 101 kPa. A 100x100x100 mesh was developed using the Structured-ALE (S-ALE) solver. The centre of the blast was oriented to be in the same plane as the head centre of gravity (CG). After the impulse from blast wave was deemed sufficiently transferred (10ms) the ALE domain was removed.

2.2 Sensitivity Study

A simulation matrix was developed to assess the range of biomechanical responses under different conditions including presence of helmet, presence of HBE, incident pressure, and blast orientation. Table 27 below summarizes the configurations evaluated. All simulations were analyzed using LS-DYNA R11.0.0 129956 double precision massively parallel processing (MPP) solver. All simulations described in this matrix were run on a high performance computing system (HPC) running Linux CentOS v6.5.

Simulations were processed in Matlab (The Math Works, Inc.). Pressure propagation in the first 10ms was inspected to confirm sufficient time was permitted to transfer blast wave impulse to the head and that pressures had returned to ambient levels. The resulting head acceleration and rotation were extracted from the simulation and prepared using channel frequency class filters (CFC) 1000 for accelerations and CFC 180 (300 Hz) filter for velocity data, respectively, based on SAE J211, 8.4.1 and ISO 6487.

Table 27. Sensitivity study simulation matrix.

Model ID	Helmet Present	HBE Present	Incident Pressure (kPa)	Blast Direction
1	-	-	200	Frontal
2	-	-	400	Frontal
3	YES	-	200	Frontal
4	YES	-	400	Frontal
5	YES	YES	200	Frontal
6	YES	YES	400	Frontal

In order to assess the relative importance of sensitivity to these potential HBE configurations, the biomechanical data was compared to common metrics and standards in the head injury field. At the time of this work, the thresholds for blast injuries are an existing area of uncertainty and the field lacks a validated neurotrauma injury criteria specific to blast [8]. With the intent of motivating development of these blast specific criteria, this work proposed to place the effects of HBE in context where similarity in mechanisms between blast and blunt injury exists, with acknowledgement that future work will be necessary to develop a validated blast trauma criterion. Selection of injury criteria was based on evaluation of multiple kinematic measures and inclusion of a temporal component to account for the effect of duration on injury. The most commonly considered kinematic injury criteria were developed for blunt impacts. Peak acceleration is a common metric for short duration events such as helmet blunt impacts. For longer duration events such as those observed in automotive crash, a time component can be included such as in Head Injury Criterion (HIC) and Rotational Injury Criterion (RIC). In a primary blast event, high magnitude accelerations can occur over short durations without direct (blunt) impact.

Table 28. Duration of accelerative event is dependent on the loading environment.

Loading Environment	Duration of Accelerative Response
Blunt (helmet impact)	< 10 ms
Blunt (automotive crash)	15 – 30 ms
Blast	< 7 ms

Three injury metrics were selected based on their usage for blast-induced traumatic brain injury (bTBI) and their application feasibility in experimental test and evaluation of HBE using a human surrogate. The metrics are based on kinematic quantities measurable experimentally and include the HIC, the Brain Injury Criterion (BrIC), and RIC in Table 29. The 50% probability of mild traumatic brain injury (mTBI), as identified in the literature to provide context for meaningful injury scores.

Table 29. Summary of injury metrics considered in this study.

Metric	Biomechanical Data Applied	50% Probability of mTBI	Metric Calculation Formula
HIC	Linear Head Acceleration	265 [12]	$HIC = \max_{(t_1,t_2)}\left\{(t_2-t_1)\left[\frac{1}{t_2-t_1}\int_{t_1}^{t_2}\lvert a(t)\rvert dt\right]^{2.5}\right\}$
BrIC	Angular Head Velocity	0.59 [13]	$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xcr}}\right)^2+\left(\frac{\omega_y}{\omega_{ycr}}\right)^2+\left(\frac{\omega_z}{\omega_{zcr}}\right)^2}$
RIC	Angular Head Acceleration	1.03×10^7 [14]	$RIC = \max_{(t_1,t_2)}\left\{(t_2-t_1)\left[\frac{1}{t_2-t_1}\int_{t_1}^{t_2}\lvert \alpha(t)\rvert dt\right]^{2.5}\right\}$

3. RESULTS

3.1 Model Validation

A critical aspect of any simulation study is that the applied model is grounded relative to experimental data or justifiable against accepted theory and literature where experimental data is insufficient. For this study, validation of global head kinematics in blast was targeted. The head CG response and cervical

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spine FEM was compared to experimental head CG kinematics from the experimental shock tube tests at 400 and 550 kPa [7]. Results are shown in the Figure 4 below for Anterior/Posterior (A-P) translation and rotation in flexion/extension. CORA was used to assess the degree of validation as shown in Table 29. Summary of injury metrics considered in this study. While multiple degree of validation scales exist, for this work we consider an excellent degree of validation indicated by a CORA score of 0.94 or greater, good indicated as greater than 0.8, and fair as greater than 0.58 [18]. The cadaveric response has been shown to be more compliant than the *in vivo* response due to the compressive and stabilizing effect of the neck musculature [16]. While the present study does not model the musculature, the resulting response demonstrates stiffness with good agreement to the lower bound standard deviation of the cadaveric experiments. For upright head simulation of blast exposure, it is expected that the APL head-neck model stiffness values more similar to in vivo response.

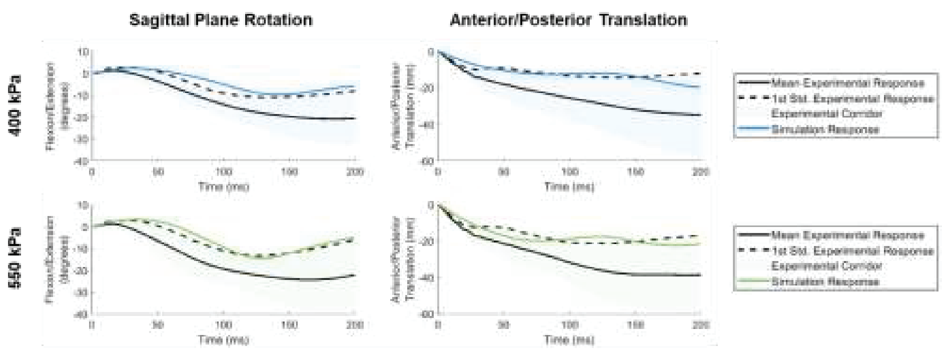

Figure 67. Model demonstrates response in line with stiff cadaveric response in rotation (left) and translation (right) to when exposed to experimental pressures at 400 kPa (top) and 500 kPa (bottom).

Table 30. Model achieves fair degree of validation relative to mean of experimental corridor and excellent when compared to 1st standard deviation of corridor

Incident Pressure	Mean Sagittal Response	1st Std. Dev. Sagittal Response	Mean A-P Response	1st Std. Dev. A-P Response
400 kPa	0.715	0.892	0.743	0.949
550 kPa	0.725	0.968	0.711	0.970

3.2 Headborne Equipment Effect on Head and Neck Biomechanics

The gross biomechanical effect of HBE on the head, when exposed to frontal blast, can be visualized in **Error! Reference source not found.** for each equipment configuration in the 200 and 400 kPa. The d istribution of the overpressure after contact with the human model can be observed in this figure, as well as the resulting head translation and rotation resulting from the blast exposure. The left column for each incident pressure depicts the FE model overlaid with a sagittal slice of the pressure contour map as the blast wave propagates at a single instance of time during the first millisecond. Right columns depict the final position of the head overlaid with the initial position at 200ms to show cumulative head displacement in each configuration. The addition of HBE increase overall head displacement in both blast exposure magnitudes. The biomechanical effect of adding HBE can be quantified by the head peak accelerations and velocities as well as their times at the peak (Table 31). In general, HBE increases head linear velocity magnitude and delay the time to its peak, but it does not increase the linear acceleration peak magnitude. The HBE effects on head rotational velocity and acceleration are more complex and quite different than the linear translational responses. HBE would increase rotational head velocity but could increase or decrease head rotational acceleration. It is worth noting that head rotational kinematic quantities peaked much later than the translational quantities which is an important characteristic of the human head supported by the anatomic multi-joint neck. The translational resultant acceleration peaks can be put in context relative to the acceleration threshold of 150g for combat helmet blunt impact test brain injury criteria as shown in Figure 6, as well as angular acceleration outcomes for flexion/extension and axial rotation. The peak accelerations observed at 200 kPa are consistently less than this test cutoff and the 400 kPa results are consistently greater. While resultant acceleration efficiently captures the peak acceleration event, it is necessary to discretize the rotational response quantities further. The primary

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rotational motion during a frontal blast event is flexion and extension; however, addition of an asymmetric NVG introduces extra head axial rotation which does increase with addition of HBE.

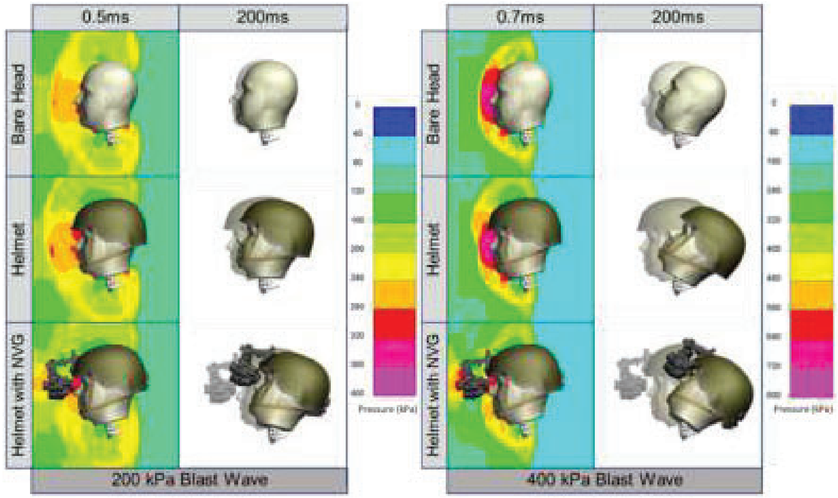

Figure 68. Snapshot of Simulation Results for 3configurations and 2 blast severities. The distribution of pressure after contact is altered by inclusion of the helmet and NVG surfaces.

Table 31. Peak and Time of Peak for Key Kinematic Quantities

Incident Pressure	Configuration	Peak Accel. (g)	Time of Peak (ms)	Peak Velocity (m/s)	Time of Peak (ms)
200 kPa	Bare Head	119.3	0.49	0.38	1.51
	Helmet	109.2	0.56	0.45	3.17
	Helmet with NVG	116.7	0.59	0.56	3.61
400 kPa	Bare Head	234.1	0.45	0.91	1.79
	Helmet	205.5	0.50	1.18	4.02
	Helmet with NVG	194.4	0.52	1.32	4.15
Incident Pressure	**Configuration**	**Peak Accel. (rad/s^2)**	**Time of Peak (ms)**	**Peak Velocity (rad/s)**	**Time of Peak (ms)**
200 kPa	Bare Head	4137.2	30.24	1.98	111.03
	Helmet	6529.7	2.07	2.63	133.48
	Helmet with NVG	6177.5	2.04	4.04	16.37
400 kPa	Bare Head	8572.8	28.06	8.87	83.83
	Helmet	11045.9	1.97	9.16	198.58
	Helmet with NVG	11694.1	17.40	10.18	88.44

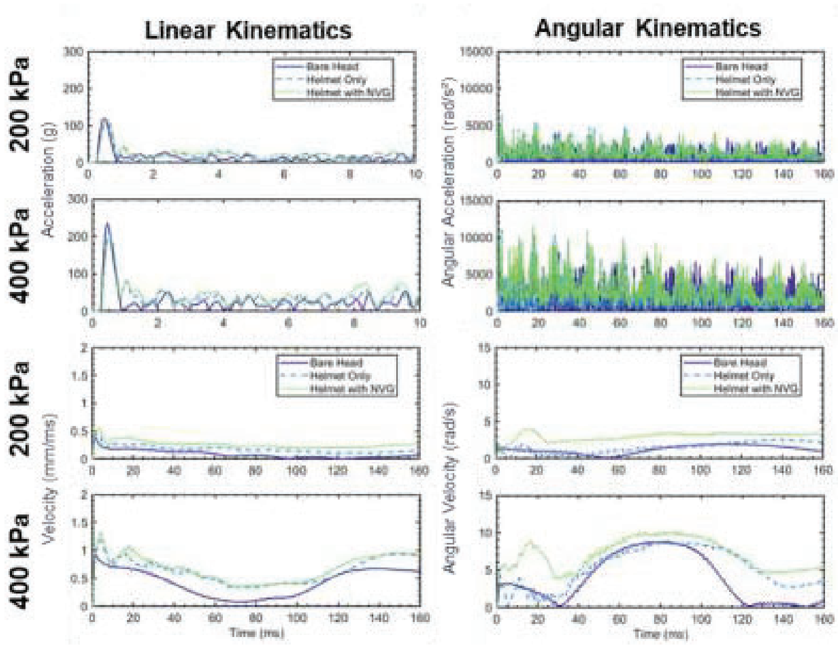

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Figure 69. Resultant Kinematic response scales with increasing blast exposure for all kinematic measures. Linear acceleration peak occurs in first 10 milliseconds and is shown for reduced window.

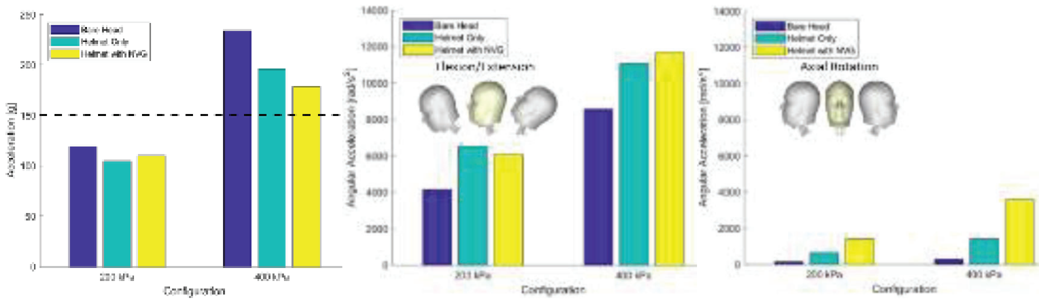

Figure 70. Linear acceleration results compared to acceleration threshold of 150g for combat helmet blunt impact threshold (left). Rotational acceleration results (middle) and axial rotation results (right).

3.3 Calculation of Common Injury Metrics & Standards

In order to relate biomechanical responses to risk of injury, human injury risk curves and metrics have been developed for numerous applications, including peak acceleration threshold metrics for short duration events such as helmet impacts in sports and criteria with temporal components for longer duration events such as automotive crash. In both cases, these metrics were validated for blunt events and have limitations when applied to blast due to differing mechanisms of trauma, though a growing number of metrics have been used to evaluate rotational motion and have been extrapolated to insults such as blast loading [22]. Comparisons with these metrics can motivate future investigations of validated blast metrics for HBE and other helmet mounted equipment if relevance can be established.

The kinematic injury metrics presented in Table 29 were calculated for all simulation conditions using the kinematic data presented in **Error! Reference source not found.**. Results are presented along with t he 50% mTBI threshold to provide context for the relative scales of each metrics. For HIC, a window of 15 ms was selected and applied to the resultant linear acceleration data. The HIC values ranged from 32 – 37 for the 200 kPa condition and 149 – 174 for the 400 kPa condition as shown in Figures 8. All values occur below the 50% mTBI threshold identified for blunt injury. The rotational kinematic metrics, BrIC and RIC are presented in Figure 71. BrIC scores range from 0.04 to 0.1 for the 200 kPa condition and 0.16 – 0.25 for the 400 kPa condition. All values occurred below the 50% mTBI threshold. RIC scores ranged from 0.17×10^7 to 0.56×10^7 for the 200 kPa condition and from 1.37×10^7 to 4.53×10^7 for the 400 kPa condition.

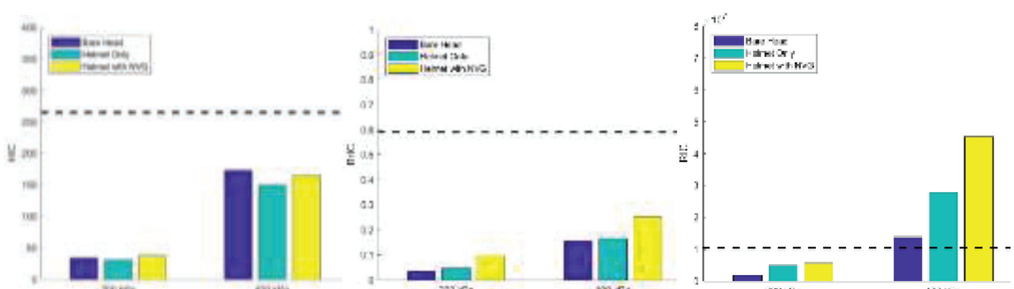

Figure 71. Model predictions of HIC score (left), BrIC score (middle), and RIC score (left). The dashed-black line is provided for context and represents the 50% probability threshold for mTBI

4. DISCUSSION

4.1 Headborne Equipment Effect on Head Kinematic Responses

This work serves as an investigation of whether blast acceleration with the addition of helmets and HBE could affect injury prediction, and therefore only one helmet and type of HBE were examined to make this assessment. The results indicate that there is a difference observed in predicted injury values when additional HBE were added, with introduction of additional axes of rotation when an asymmetric NVG was added. Future work should expand the type of HBE studied to provide a more comprehensive assessment for fielded equipment, as well as examine additional factors that might be affect the aerodynamics of the overpressure and shock wave transmission, such as shape, mass, and CG of HBE. Results demonstrate that total displacement and rotation of the head increases with both increasing blast severity and with addition of HBE. In the helmet with NVG case, we see additional axial rotation of the head due to the asymmetric design of the monocular NVG. For both translational and rotational kinematics, the peak resultant velocity and the time of peak predicted by the model increases both with exposure level and with addition of HBE. Considering all exposure levels and HBE configurations, translational velocity increases by an average of 35% with the addition of HBE and rotational velocity increases by an average of 38%. The percent increase in translational velocity is similar for both exposure levels; however, the percent change in rotational velocity is greater at the lower exposure, 68% for 200kPa, compared to the higher exposure, 9% at 400kPa. These results suggest that while the magnitude of rotational velocity increases at higher exposure levels, the effect of additional HBE is most evident at the lower exposure evaluated by this study.

The peak translational acceleration increased with more severe blast exposure and the addition of HBE decreased the magnitude of acceleration by 10%; however, addition of HBE did not clearly increase or decrease peak translational acceleration. The peak rotational acceleration increases with addition of HBE by an average of 43% and occurs at least 11ms earlier in all exposures. Rotational acceleration results can be further broken into flexion/extension rotation and axial rotation components. While addition of HBE increases flexion/extension rotation, addition of the NVG did not further notably change flexion/extension rotation; however, the NVG addition has a demonstrated effect on axial rotation, with addition of the helmet increasing average peak axial rotation by 383% and with helmet and NVG increasing by 1000%. The decrease in translational acceleration can be attributed in part to the increase in overall system inertia with addition of HBE mass, as well as translation of additional impulse from the blast into rotational acceleration due to addition of a moment from the helmet strap to the head resulting in increased flexion/extension rotation and introduction of an asymmetric cross-sectional area with addition of the NVG resulting in increased axial rotation. These results suggest that placement of additional HBE on the helmet introduces rotational kinematics with lower magnitude than the primary rotational direction. Future work should aim to validate models capable of assessing lateral blast exposure with high fidelity to further understand the effect asymmetric changes to the cross-section of HBE.

4.2 Application of Common Injury Metrics & Standards

This work aimed to study how the head and neck responds with the addition of HBE in the context of the kinematic assessments commonly applied to evaluate personal protective equipment, and to provide motivation for the future consideration of how HBE design could influence kinematic response of the head and neck during a blast event. Assessment of relative effect of HBE on head and neck kinematics was conducted with respect to existing standards and requirements, though these are validated primarily for blunt loading environments. For short duration blunt events, the 150g translational acceleration threshold is commonly applied for combat helmet blunt impact test based on risk of brain injury [15]. Prior work suggests that 275 kPa is an exposure level where instance of mTBI can be identified, and this factored into selection of one higher and one lower exposure level for this study [11]. The simulation results demonstrate translational acceleration peaks that agree with this prior study, with peak translational acceleration below 150g in the 200 kPa simulations and above 150g in the 400 kPa simulations; however, this cut off is not immediately informative of the significance of addition of HBE and this criterion cannot account for duration of the accelerative event which will likely influence injury due to inertial loading. Understanding of inertial events required evaluation of kinematic metrics and standards with a temporal component such as HIC, BrIC, and RIC.

Addition of HBE produced complex head response effects when exposed to a frontal blast wave. The overall head displacement and translational velocity would increase which may not correlate to increased brain injury risk based on the existing injury criteria. It does not increase the translational acceleration magnitude nor its integrated form for brain injury risk assessment. These results are expected from biomechanics perspective, as the added HBE increases the inertial resistance to acceleration without increasing the exposure cross-sectional area. Therefore, HBE would not expect to increase brain injury risk in frontal blast exposure if the head translational acceleration is the dominant factor. HIC measures the likelihood of head injury due to blunt impact, but has been applied more broadly to allow for comparison across insults including blast. This metric incorporates both the effect of head acceleration

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and the duration of this acceleration and can be limited in that severe, short duration accelerations may be assigned similar likelihoods as less-severe, long duration accelerations. Study results indicated less than 50% probability of mTBI for all exposures and HBE configurations and no substantial difference in HIC between Bare Head and with HBE. It has been observed experimentally that short duration (<7ms) peaks in excess of 1000g have the potential to occur. In this study, the filtered resultant accelerations indicate 100 – 250 g peaks with a duration of 1ms with oscillating peaks of 50g decaying across the remaining relevant window for HIC. Results show that HBE increases the head rotational velocity which could increase the risk of brain injury. BrIC considers both the duration and severity of an insult as a function of rotational kinematics, specifically rotational velocity. BrIC prediction increases slightly with addition of helmet and substantially with addition of helmet and NVG, suggesting that while BrIC as an assessment of injury may be limited, the magnitudes of change in angular velocity may be informative to the PPE design community when evaluating new HBE. Considering relative changes in BrIC, the bare head and HBE conditions where the surface geometry is systematic result in similar BrIC scores, while addition of asymmetric surfaces results in double the BrIC score. The RIC score demonstrated the most variation across HBE and mounted equipment conditions, with simulation results ranging from $0.17x10^7$ to $4.53x10^7$. The predicted RIC values are within the relevant range for brain injury based on the proposed criterion, with low exposures falling below the proposed threshold and high exposures above the proposed threshold. Additionally, addition of the helmet increases predicted RIC, which is further increased by addition of an NVG suggesting sensitivity to HBE design and configuration. The increase of head rotational acceleration and its associated brain injury criterion RIC due to HBE, could be an area of caution for PPE design and calls for further investigation. Continued effort is necessary to understand the mechanisms of blast injury and develop validated injury criteria. An area for further investigation could include analysis of the correlation between the head rotational acceleration or RIC with brain strain and/or strain rate which are based on brain tissue responses and potential tissue damage mechanisms.

One may notice that there appears to be a discrepancy of injury risk prediction between HIC/BrIC and RIC criteria. Both HIC and BrIC predicted low levels of injury risk while RIC predicted high levels of injury risk in our simulation results. This discrepancy is largely reflective of the injury criteria equation and the kinematic response characteristics of the head under our simulated blast exposure conditions. Both HIC is time integration of the acceleration signals (Table 4). The linear acceleration results from our study showed very short time pulse duration (typically <1 ms) and well below the common integration time period of 15 ms. Therefore, the peak acceleration magnitude is likely a more relevant injury criteria than HIC for this application. On the other hand, the high RIC results may suggest a greater role of head angular acceleration for brain injury in this application than the head angular velocity on which BrIC calculation is based.

4.3 Assumptions and Limitations

The assessment of head injury is a field of ongoing research. While numerous experimental and biomechanical studies have applied metrics such as HIC, BrIC, and RIC, as well as other tissue-based injury criteria to assess injury risk, the direct attribution of an injury mechanism has yet to be conclusively determined, in particular for more challenging assessments such as injuries occurring after the initial accelerative peak during blast loading. Additional research is needed to make this connection between blast loading and injury. In this study, injury metrics are calculated to provide context to the observed kinematic response. This assumes that the criteria are relevant for the loading based on prior use in the literature or sufficiently informative in providing context to warrant further study to confirm relevance to brain injury risk due to blast inertial loading. The hardware furnished at the time these results were generated during this study did not include fabric components or wires of the system. Higher fidelity HBE models would be necessary to enhance prediction specificity and may be a component of future work. Similarly, additional work is needed to fully validate the coupling of the NVG to the helmet and the helmet to the head. Ongoing work is assessing the stability of these systems, and this work could be incorporated into future studies to validate the coupling of this key load path. Finally, due to limited material data from NVG equipment suppliers, while helmet components have validated deformable material models, HBE components were modeled as rigid surfaces to enable pressure distribution, but cannot capture pressure transmission. This work could include validation with respect to whiplash experiments and incorporation of passive muscle response. Given the current model validation and the proposed future work, the model as applied in this study is sufficient for assessment of relative kinematic effects of HBE, but cannot provide an absolute injury risk assessment.

5. CONCLUSION AND FUTURE WORK

These findings provide insight into the relative importance of assessing how current HBE influences biomechanics during a blast overpressure event during the equipment design process. While there was an increase in blast-induced kinematic effects and a corresponding relative increase in injury metric score with HBE compared to the bare head condition, the addition of HBE showed reduced translation kinematics compared to the helmet-only condition. This demonstrates that there may be limitations in the assessment of risk of injury based on translational injury criteria such as HIC and that the addition of HBE is not a straightforward increase in injury risk. Rotational kinematics may offer more insight, specifically rotational acceleration, but assessment is limited by the need for ongoing fundamental research into the mechanisms of brain injury and further review of the proposed injury threshold. Investigation of HBE parameters such as mass properties and geometry, of helmets and other HBE may offer opportunities to tune designs to reduce head kinematics. This would aid in reducing tertiary injury, but also offer opportunity to reduce primary injury due to blast loading. These computational capabilities can be used to optimize headborne equipment design prior to production of physical hardware, as well as provide insight to end-users and decision makers regarding acquisition and usage of HBE.

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