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Evaluation of a Combat Helmet Under Combined Translational and Rotational Impact Loading

by Ryan Neice and Thomas Plaisted

NOTICES

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Traumatic brain injury (TBI) is a health threat to military and civilian populations. Recent biomechanics research suggests that rotational kinematics are a significant cause of brain deformation during blunt impact events to the head. Military helmet blunt impact test methods evaluate protective capacity in terms of peak translational head acceleration using a drop impact format. This study investigated the utility of a different impact format based on a pneumatic ram to induce combined translational and rotational head motion to evaluate blunt impact protection in combat helmets. This test method has been adopted for certification tests related to professional and amateur football helmets. Impacts were performed on a Hybrid III head and neck assembly in the bareheaded and helmeted condition for a total of 126 impacts. Translational and angular acceleration was recorded along with high-speed video for each impact. A multibody model that relates the head kinematic response to brain strain, referred to as Diffuse Axonal Multi-Axis General Evaluation, was used to estimate the extent of injury resulting from impact. Results indicated that, on average, the Advanced Combat Helmet reduces headform translational and rotational kinematics compared to the response of the bareheaded case. Impacts to the rear of the head, for both bareheaded and helmeted impacts, resulted in the greatest predicted brain strain. Furthermore, rotational velocity showed a stronger correlation to predicted brain deformation than rotational acceleration under these impact conditions.</td></tr>
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This research was accomplished under Cooperative Agreement No: W911NF-19-2-0246. The views and conclusions contained in this document are those of the authors and should not be interpreted as representing the official policies, either expressed or implied, of the DEVCOM Army Research Laboratory or the US Government. The US Government is authorized to reproduce and distribute reprints for Government purposes notwithstanding any copyright notation herein.

1. Introduction: Military Traumatic Brain Injury (TBI) and Rotational Head Loading

In the United States, human traumatic brain injury (TBI) is a significant societal health problem, as evidenced by the estimated 1.7 million annual occurrences, of which 80% lead to emergency room visits, 16% require hospitalization, and 3% result in death (Faul et al. 2010). Brain injuries in civilian populations often occur during sports and recreational activities, automotive collisions, and accidents. TBI is also a significant threat to the Warfighter, especially when considering the demanding physical requirements of training and the dangers of combat. The Defense and Veterans Brain Injury Center (DVBIC) reports that there were over 400,000 TBIs diagnosed within the US military from 2000 through 2019, as shown in Fig. 1. Around 60% of those TBI diagnoses occurred within the Army alone (DVBIC 2019). Worth noting is that these statistics are solely from medical diagnoses and do not account for unreported brain injuries.

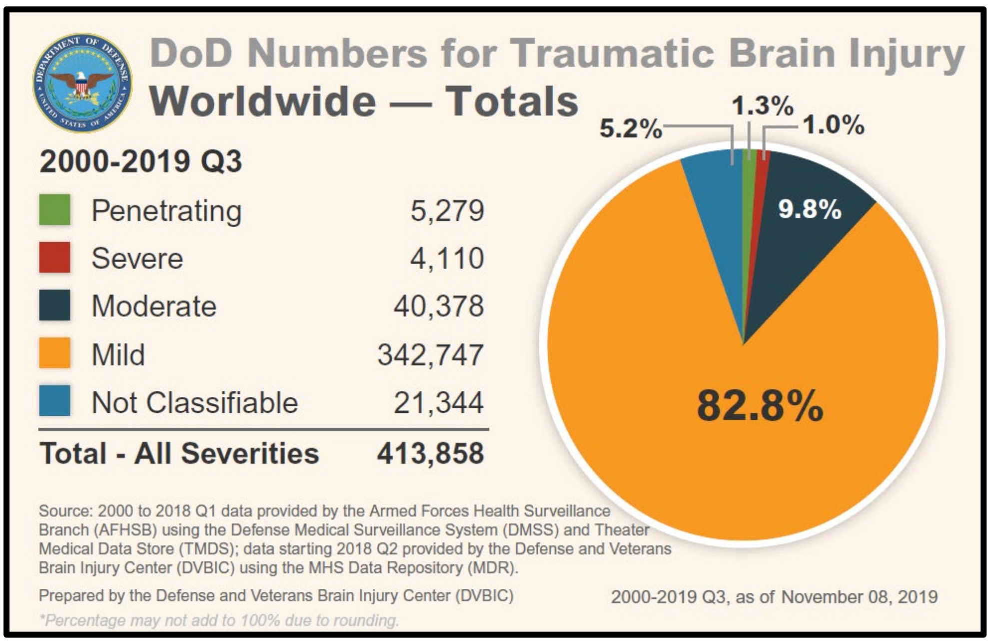

Fig. 1 Worldwide (foreign and domestic) medical diagnoses of TBI occurring in US military populations from 2000 to 2019 (DVBIC 2019)

Classification of diagnosed US Department of Defense (DOD) TBI injuries is achieved through symptomatic and clinical diagnostics. Of the hundreds of thousands of diagnosed DOD TBIs, over 80% were described as being mild (mTBI), often referred to as a concussion. Symptoms of mTBI include confusion, disorientation, memory loss, and short-term loss of consciousness (Table 1). In this reporting, penetrating injuries are mutually exclusive to TBI diagnoses, meaning that if there is a penetration injury, it is immediately assigned its own classification separate from the TBI-only cases.

Table 1 Classifications of TBI as reported by DVBIC (2019)

Brain injury descriptions (DVBIC)		
Classification	**Symptoms**	**Brain imaging**
Penetrating	Scalp, dura mater, or skull penetrated	x
Severe	Confusion or disorientation more than 24 h Memory loss lasting more than 7 days Loss of consciousness greater than 24 h	Abnormal
Moderate	Confusion or disorientation more than 24 h Memory loss lasting between 1 and 7 days Loss of consciousness between 30 min and 24 h	Normal or abnormal
Mild	Confusion or disorientation less than 24 h Memory loss lasting less than 24 h Loss of consciousness for up to 30 min	Normal

Biomechanics research suggests that rotational kinematics are a significant contributing cause of diffuse brain injury, wherein damage to the white matter is distributed throughout the brain tissue, often leading to mTBI, or in more severe cases, diffuse axonal injury. Diffuse brain injury results from excessive motion of the brain with or without a direct impact to the skull. This theory was first proposed in 1943 by Holbourn and has grown over the decades with support from numerous research investigations utilizing a variety of tools such as human cadavers, animal subjects, clinical diagnostics, and finite element modeling (Holbourn 1943; McKee and Daneshvar 2015; Antona-Makoshi 2016; Alshareef et al. 2017; Giudice et al. 2018). At the material level, the shear modulus of brain tissue is several orders of magnitude less than its bulk modulus, creating an increased propensity to deform under rotation compared to pure translational motion (Kleiven 2013; Budday et al. 2019). This mechanical behavior is similar to that of an incompressible material, owing to the high water content of the brain. Deformation of the brain has been proposed as a tissue-level metric that can be related to the probability of a TBI based on available injury data (Takhounts et al. 2013). Many kinematic-based metrics have been proposed, as reviewed by Sanchez et al. (2017), including injury risk thresholds based on angular velocity, linear (translational) or angular acceleration, and combinations of different kinematic measurements. The advantage of kinematic-based injury metrics is that head motion can be more easily measured than brain deformation, particularly on living human subjects. Improved understanding of the link between head kinematics and TBI risk is essential for creating effective performance standards to evaluate helmet protective equipment, for both civilian and military applications.

The US Army fields a variety of combat helmet models to protect the Soldier from ballistic and blunt impact threats. Blunt impact protection is assessed by a drop impact format that measures peak linear acceleration experienced by the headform based upon a modified version of the Federal Motor Vehicles Safety Standard 218 (FMVSS 218 [DOT 2011]). As described in the Purchase Description for the Advanced Combat Helmet (ACH), testing involves fitting the helmet on an instrumented magnesium Department of Transportation (DOT) headform, which is dropped from an elevation, guided by a monorail, such that impact occurs onto a rigid, stationary anvil with hemispherical shape at a prescribed velocity (CO/PD-05-04 [PEO Soldier 2012]). The standard requires that helmets must limit translational headform acceleration (measured in the vertical direction) to less than 150 g's (g-force, 9.81 m/s^2) when impacted at 10 fps at the crown, front, rear, sides, and nape regions, as shown later in Fig. 3. The ACH meets this performance standard with the 7-pad Team Wendy Zorbium Action Pad (ZAP) suspension system. Combat helmets are not evaluated for mitigating rotational kinematics under this standard.

The purpose of this study is to investigate a different impact format based on a pneumatic ram to induce combined translational and rotational head motion to evaluate blunt impact protection in combat helmets. The linear acceleration response of the headform is compared to the drop test evaluation method. Rotational kinematics are assessed in terms of a metric that uses a multibody analytical model to relate the head kinematic response to brain strain. This research could lead to enhanced helmet evaluation methods that are more representative of the blunt impact events experienced in combat and training, while promoting innovations toward improved protection technologies for reducing TBI risk for the Warfighter.

2. Pneumatic Ram Test Method

The pneumatic ram test method, also known as the linear impactor format, has been used in recent years to evaluate the performance of professional and amateur football helmets. The National Football League (NFL) has made the test part of its helmet certification requirement (Helmet Test Protocol [Biocore 2019]). Additionally, the National Operating Committee on Standards for Athletic Equipment (NOCSAE), the organization that certifies all football helmets in the United States, has implemented new standards based on this method (NOCSAE DOC (ND) 081 [NOCSAE 2019]). The pneumatic ram testing apparatus used in this research study is manufactured by Biokinetics and Associates, LTD (Ottawa, Canada) and is shown in Fig. 2. The machine uses compressed air to drive a ram (mass = 16 kg) toward a surrogate headform and neck, which are in turn secured to a table that is free to translate upon impact. The surrogate head and neck consists of a Hybrid III 50th percentile male, which is used to measure the impact kinematics for the pneumatic ram test setup. This head–neck complex is initially at rest and has a free boundary condition to slide along the direction of the impact vector (x direction, as labeled in Fig. 2). The total mass of the head–neck and sliding table assembly is 17.7 kg.

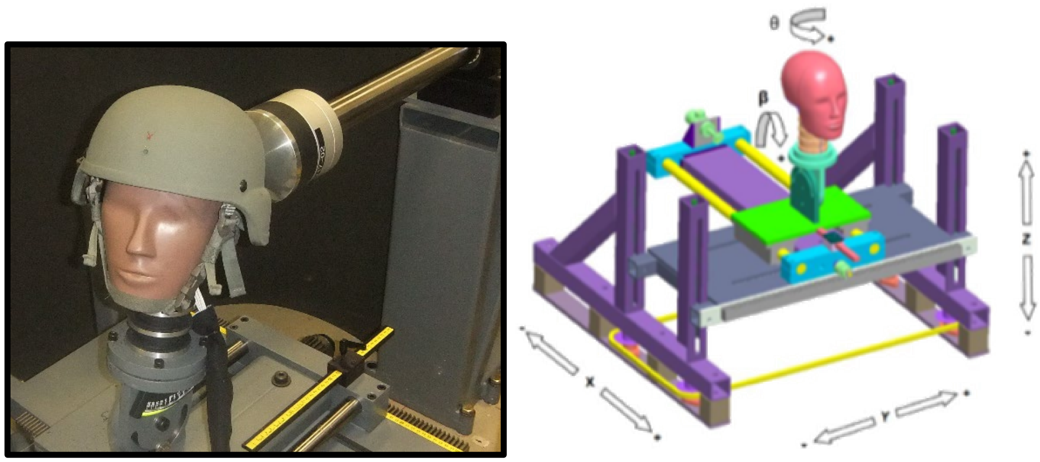

Fig. 2 **(Left) Pneumatic ram machine setup. (Right) Target assembly used in the pneumatic ram method consisting of surrogate head and neck on a translating table (Biocore 2019).**

Rotational motion is achieved at the head–neck joint as well as in the flexible neck itself in this setup. The Hybrid III contains a pre-tensioned cable that runs along the axial direction of the neck. This cable is checked intermittently to ensure that it maintains its intended torque value at 12 in-lb during impact testing. The headform contains a nine-accelerometer array package (NAP) consisting of nine linear accelerometers in a 3-2-2-2 arrangement (Padgaonkar et al. 1975). The first set of three accelerometers in this package are located at the center of gravity (CG) of the Hybrid III head, while the additional three sets of two accelerometers are positioned at a fixed distance away from the head CG. This accelerometer package allows for the calculation of linear and rotational head kinematics as detailed by Padgaonkar et al.

One major difference between the test protocols used by the NFL and NOCSAE is the use of different impactor tips. The NFL impact tip uses a nylon cap backed with vinyl nitrile foam to mimic the construction of a typical football helmet. The NOCSAE tip consists of a polyurethane elastomer disk (124 mm diameter × 38 mm thick, durometer 42 Shore A) sandwiched between two aluminum plates;

the outer plate that strikes the helmet has a radius of curvature of 127 mm. The NFL tip is intended to mimic a helmet-to-helmet impact in a football collision; the NOCSAE tip provides a combination of hard impact surface with a compliant response to impart a repeatable nonrigid impact without inducing gross structural deformation in the helmet shell and thus is judged to be more relevant to potential military impact scenarios.

The helmet impact locations for the existing monorail-based test, as required by the ACH purchase description and further detailed by the Army Test Center (ATC) test procedure, are shown in Fig. 3 (Bruggeman 2013). Figure 3 also shows similar impact locations reproduced in the pneumatic ram impactor test setup. These impact positions were determined by placing an ACH on the Hybrid-III headform and adjusting the table height and head and neck pose to achieve an identical impact location. Helmet fit was kept consistent between each impact with a helmet positioning tool that was used to ensure a repeatable nose-to-brim distance of 3 inches. Helmet retention strap lengths were also kept consistent throughout this test procedure and in agreement with the ATC procedure. Consistent helmet fit is an important factor to consider when using the pneumatic ram test method, since the head and neck undergo large excursions post impact, which shift the helmet significantly from its original position. A positioning guide is used to ensure the head is impacted at a repeatable point in space that exists just beyond the point where the impact ram is in free-flight and has reached the target velocity.

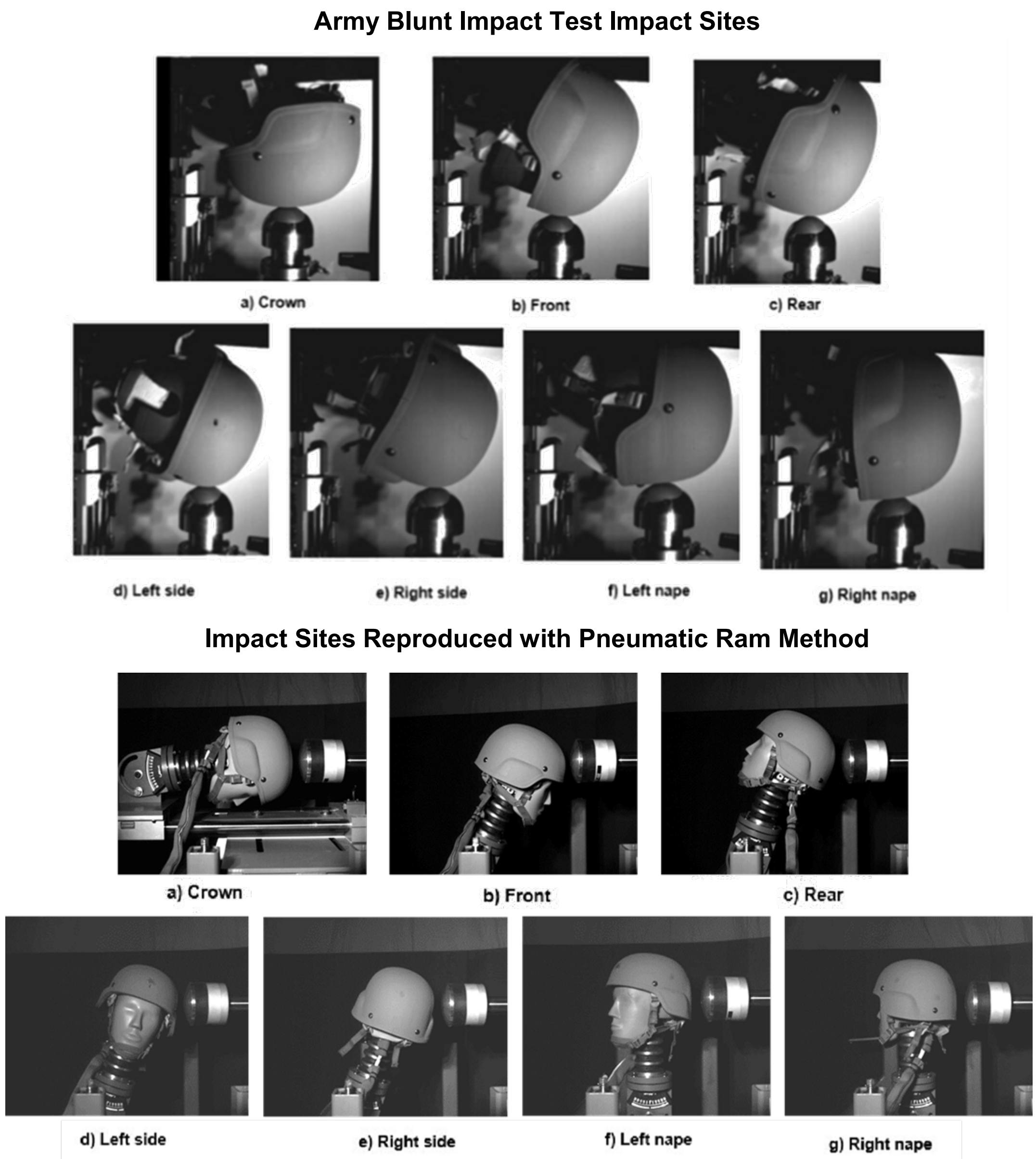

Fig. 3 (Top) Helmet impact location per the ATC blunt impact procedure on the Army standard monorail-based test format. (Bottom) Corresponding helmet impact locations for the pneumatic ram test setup.

The translations and rotations in head and neck pose to replicate the impact locations of the Army standard ATC procedure were carefully tracked. First, a general head position was defined so that any head configuration adjustments could be related to a neutral position. The *general position* was defined by the neck being vertically oriented at 90° with respect to the sliding table and the centerline of the impactor aligned to the intersection of the transverse and sagittal planes of the head. Rotations of the head to the left and right were kept symmetric to ensure identical hits were delivered to the left and right sides of the head. Despite this measure, the left and right responses may differ due to possible unintended asymmetries in the headform, neck, sensors, or helmets. Table 2 illustrates the translations and rotations of the table and neck positioning hardware to achieve the desired impact locations. Note that all theta head rotations were made around the central vertical axis of the Hybrid III neck. Pneumatic ram impact machines from other manufacturers may not necessarily give theta rotations about the same neck-centered axis.

Table 2 Pneumatic ram translation and rotation adjustments to achieve comparable helmet impact locations to those used in the ATC blunt impact procedure (Biocore 2019)

Impact location	Translation table		Rotation assembly	
	Y (mm)	**Z (mm)**	**β (°)**	**θ (°)**
General position	0	0	0	0
Crown	0	287	80	0
Front	0	10	26	0
Rear	0	39	20	180
Left side	–12	4	20	–90
Right side	12	4	20	90
Left nape	–5	2	0	–150
Right nape	5	2	0	150

3. Bare Head and ACH Pneumatic Ram Impact Testing

Impacts were performed at 10, 14, and 17 fps on a bare Hybrid-III head and then repeated when fitted with a size large ACH helmet with Team Wendy suspension padding in the standard configuration. A total of 126 impacts are included in this study, consisting of three impacts at each of the seven impact locations, repeated at the three different velocities. Approximately 2 to 3 min elapsed between impacts at the same location. Greater than 10 min elapsed between impacts when changing to other impact sites while the head and neck were repositioned. There was no noticeable degradation of the helmet shell or padding during this test series. High-speed video was recorded for each impact sequence at 2000 fps. The NAP sensor data was collected at 10 kHz and filtered with a CFC 180 filter, the same filter as called for in football helmet test protocols. A tool for evaluating the consistency of the NAP provided by Biocore was used to check the feasibility of accelerometer channel results (Gabler et al. 2018). A NAP consistency check was performed at the beginning and end of each test series. For the experiments included in this research, no abnormalities were detected by the NAP check. The helmet was repositioned on the head after each impact to keep consistent impact positioning. The Hybrid III neck was re-torqued to 12 in-lb twice during this test series, after impacts numbered 54 and 90 out of 126. No degradation to the neck, in the form of tearing or permanent deformation in the rubber components, was observed during this test series.

Figure 4 presents representative rotational data traces calculated from the resultant of the x, y, and z direction during a front impact at 14 fps along with corresponding video screenshots at peak angular acceleration and peak angular velocity. Peak linear and angular accelerations occur early in the time history, in this example, within the first 5–7 ms of the loading event. Peak angular velocity occurs later in time, in this example, around 25 ms. Between the pre-impact and peak angular acceleration frames, compression of the ACH pads can be observed, as the standoff distance between the front of the ACH shell and Hybrid III head decreases. It is during this time that the linear and angular accelerations are reaching peak value. Later in the time history, the head begins to rotate and the neck flexes. The head reaches peak angular velocity at 25 ms (in this example), though continues to move at a lower velocity after the 30-ms segment of data presented. All head kinematic data presented is according to the standard J211 head coordinate system.

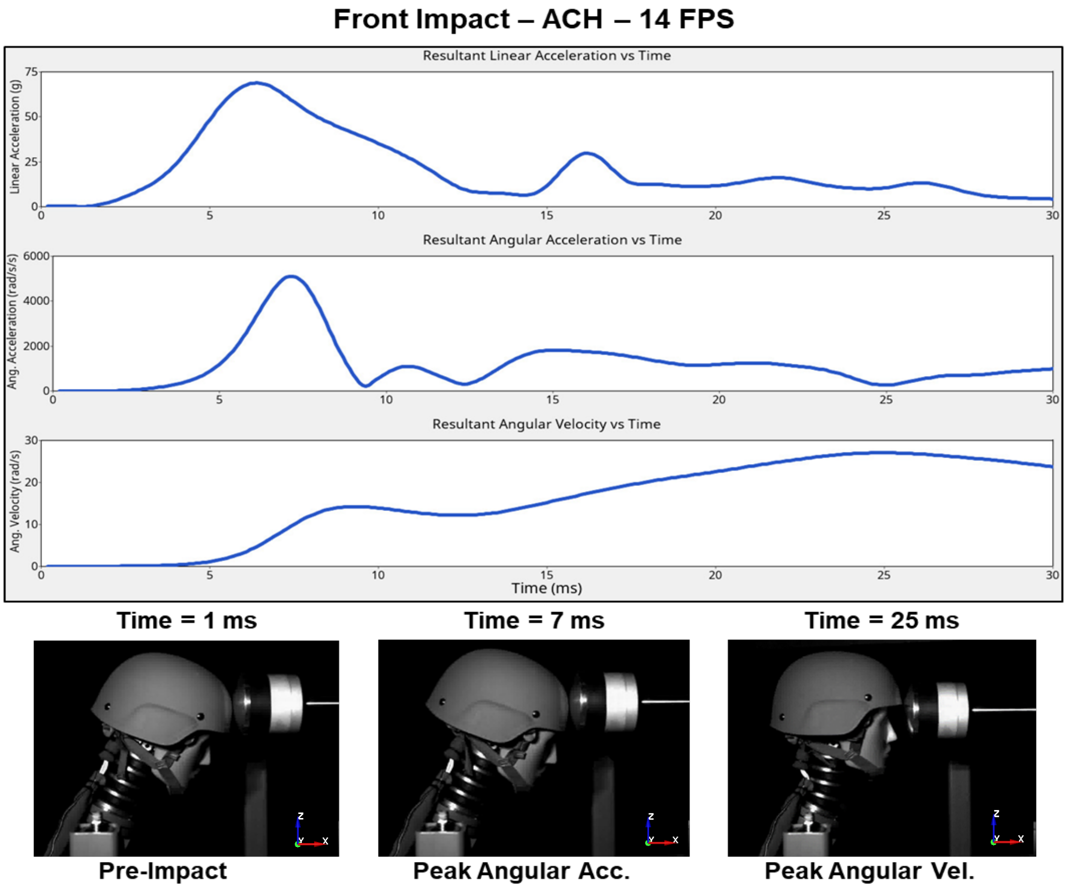

Fig. 4 Example of resultant data traces for a pneumatic ram impact to the front of a Hybrid III head with ACH helmet at 14 fps. (Bottom) Images of the head and helmet at stages between pre-impact and peak angular acceleration during pneumatic ram impact. Compression of the front pads can be observed as the space between the forehead and helmet shell decreases. At peak angular velocity, the head is rotating and neck is flexing away from the impactor.

Peak resultant linear accelerations were compared between the bareheaded and ACH helmeted impact conditions (Fig. 5). Peak linear accelerations in the helmeted case ranged from 23 g (crown: 10 fps) to 144 g (left nape: 17 fps). The helmet effectively mitigates linear headform acceleration at 10 fps, which corresponds to the velocity at which current blunt impact testing is performed per the ACH Purchase Description (PEO Soldier 2012). Impact mitigation performance decreases as impact velocity is increased. This holds especially true at certain locations in 17-fps impacts, as seen in Fig. 5, where peak linear acceleration was hardly reduced at the front and nape locations between bareheaded and helmeted conditions. This is due to an overmatch condition, where fewer pads (less pad area) are engaged during the event and the pads reach full compression (bottoming out) prior to fully arresting the headform. Across all impact locations and velocities, crown impacts were generally the least severe, due in part to the greater amount of pad area engaged. However, at all three velocities the peak linear acceleration stayed below 150 g's for both bareheaded and helmeted conditions.

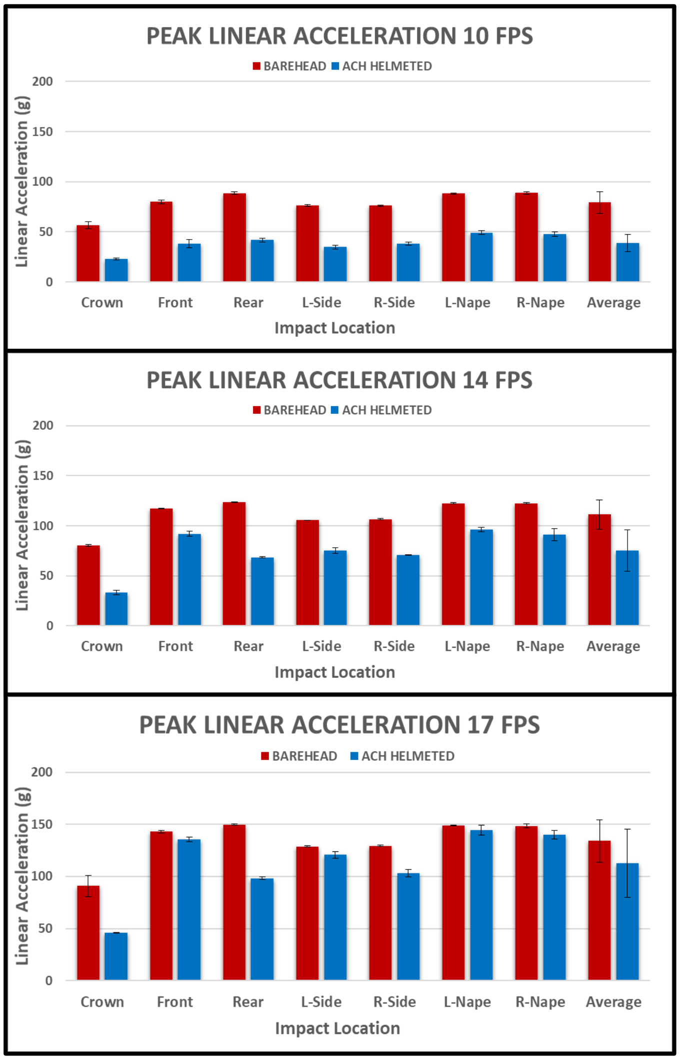

Fig. 5 Peak linear accelerations at each impact location for bareheaded and ACH helmeted Hybrid-III at 10-, 14-, and 17-fps impacts

Peak resultant angular accelerations and velocities were compared between the bareheaded and ACH helmeted conditions (Fig. 6). Peak angular accelerations with the ACH ranged from 962 rad/s/s (crown: 10 fps) to 10,747 rad/s/s (left side: 17 fps) and peak angular velocities with the ACH ranged from 3.0 rad/s (crown: 10 fps) to 40.7 rad/s (rear: 17 fps). In certain test conditions, like the front impact, the peak angular acceleration actually increased in helmeted versus unhelmeted impacts. In other conditions, such as the rear location, peak angular acceleration decreased while peak angular velocity increased. The cause of these phenomena are not fully understood, but may be due to the increased moment arm created by contacting the helmet further away from the CG of the head in a helmet versus unhelmeted impact. On average, the ACH decreases peak angular accelerations of the head, although similar to linear acceleration, there are diminishing returns at higher velocities. Looking solely at angular velocity, the ACH consistently reduces peak resultant velocity across all impact velocities. The relative importance of angular acceleration versus angular velocity as a helmet performance metric is addressed in the following section.

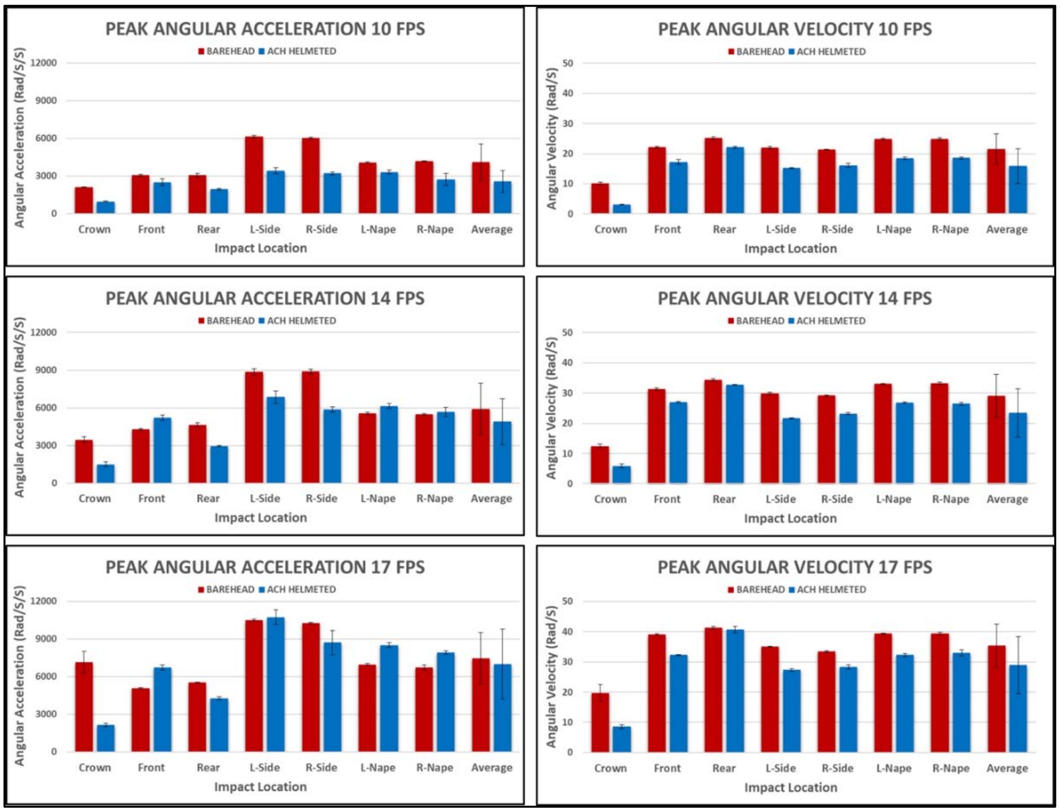

Fig. 6 Peak rotational acceleration (left) and velocity (right) by impact location for bareheaded and ACH helmeted Hybrid-III at 10-, 14-, and 17-fps impacts

4. Analysis of Results

Peak linear accelerations from the pneumatic ram impacts were compared to FMVSS 218 drop tower tests on the ACH. As shown in Fig. 7, the drop test produces higher linear accelerations than the pneumatic ram test for equivalent impact velocities and locations, which arises from the different boundary conditions between the test methods. For example, the change in velocity in the headform as a result of impact between these methods is different. In the drop tower format, the helmet approaches the impact anvil at a prescribed velocity. Upon impact, the headform is free to rebound off the anvil, thus creating a change in velocity that is higher than the drop velocity. In the case of the pneumatic ram, the helmet is originally at rest, so that the change in velocity is simply the speed of the impact head. Another difference in the setups is the impacting surface. The drop tower uses a rigid steel anvil, while the pneumatic ram impactor consists of an aluminum strike face backed by a deformable elastomer with an aluminum cap. The head and neck hardware for both formats are also different. The drop tower uses a magnesium headform with a rigid metal attachment to the guiding monorail, while the pneumatic ram uses a Hybrid III with vinyl skin and a flexible neck, which imparts a significant amount of impact attenuation to the test apparatus. Due to its higher relative rigidity, the current drop tower method may be a better test to evaluate the helmet's ability to resist denting and prevent skull fracture.

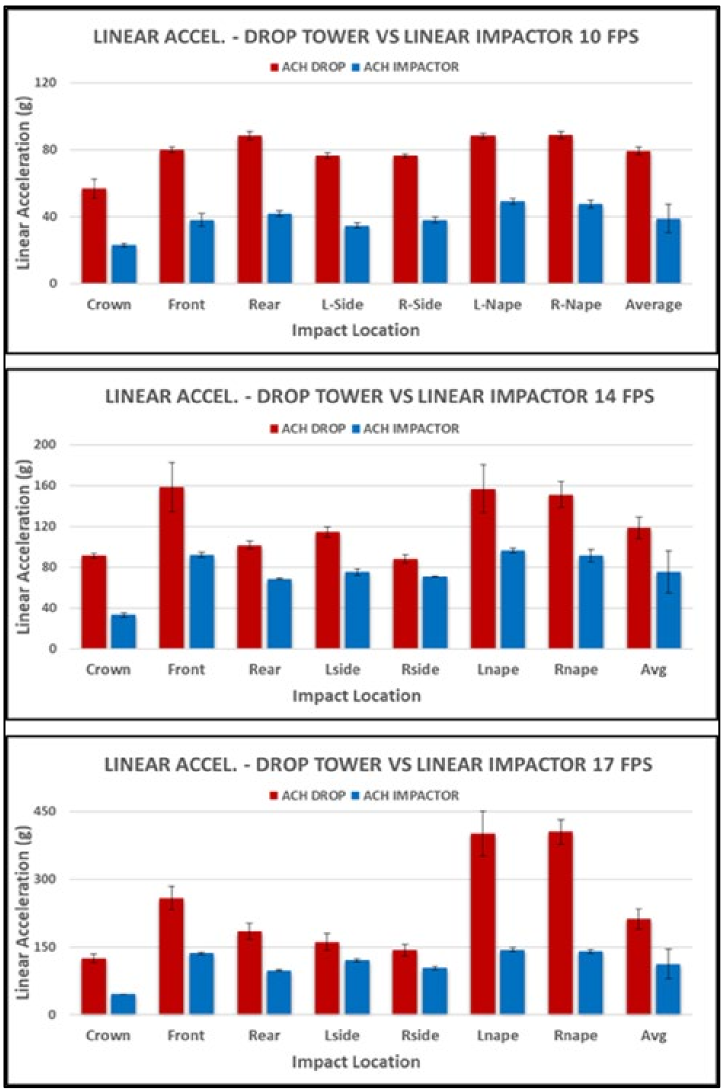

Fig. 7 Comparison of ACH peak linear accelerations between the Army standard drop tower method and nonstandard pneumatic impactor test methods

A metric was applied to assess the severity of the rotational loading and derive an overall performance score for the simulated head impacts in this test series. Debate continues within the helmet certification community as to the best kinematic predictors of diffuse brain injury to use in laboratory helmet evaluation. Maximum principal brain strain (MPS) quantifies the extent of brain deformation, leading to TBI, and is most commonly estimated through computational modeling of the brain tissue using finite element analysis (Takhounts et al. 2013; Giudice et al. 2018). To alleviate the computational expense of performing lengthy, tissue-level finite element analysis for every impact scenario, the Diffuse Axonal Multi-Axis General Evaluation (DAMAGE) model has been developed to offer a quick assessment of MPS due to rotational kinematics (Gabler et al. 2019). DAMAGE is a multibody analytical code that incorporates directional-specific stiffness and damping coefficients to predict brain tissue response within seconds. Of the available kinematic-based assessments, DAMAGE has shown the highest correlation to finite element brain deformation models (Gabler et al. 2019). Currently, DAMAGE is best used to compare relative helmet performance, as the link between brain deformation and TBI is not firmly established. As the understanding of the injury tolerances of the human brain are expanded, MPS predictions from DAMAGE and computational simulations can be used in conjunction with injury risk functions to determine the probability of a TBI occurring.

DAMAGE scores were generated for each of the impacts and averaged at each location and velocity (Fig. 8). With this analysis, it is possible to estimate the level of strain imparted to the brain during an impact event as well as determine the relative severity of different impact locations around the helmet. DAMAGE analysis suggests that the most severe hit location in this test series is the rear impact site. This is counterintuitive, if one were to solely evaluate the severity of impact by peak angular acceleration. For instance, in the bareheaded condition, the peak angular accelerations of side impact (8800 rad/s/s at 14 fps) were greater than those of the rear impact (4600 rad/s/s at 14 fps). Rear impacts did however experience higher angular velocities: 34 rad/s at 14 fps compared to 30 rad/s at 14 fps for side impacts. This result suggests that rotational velocity is a better correlate to predicted brain strain for the pneumatic ram test setup.

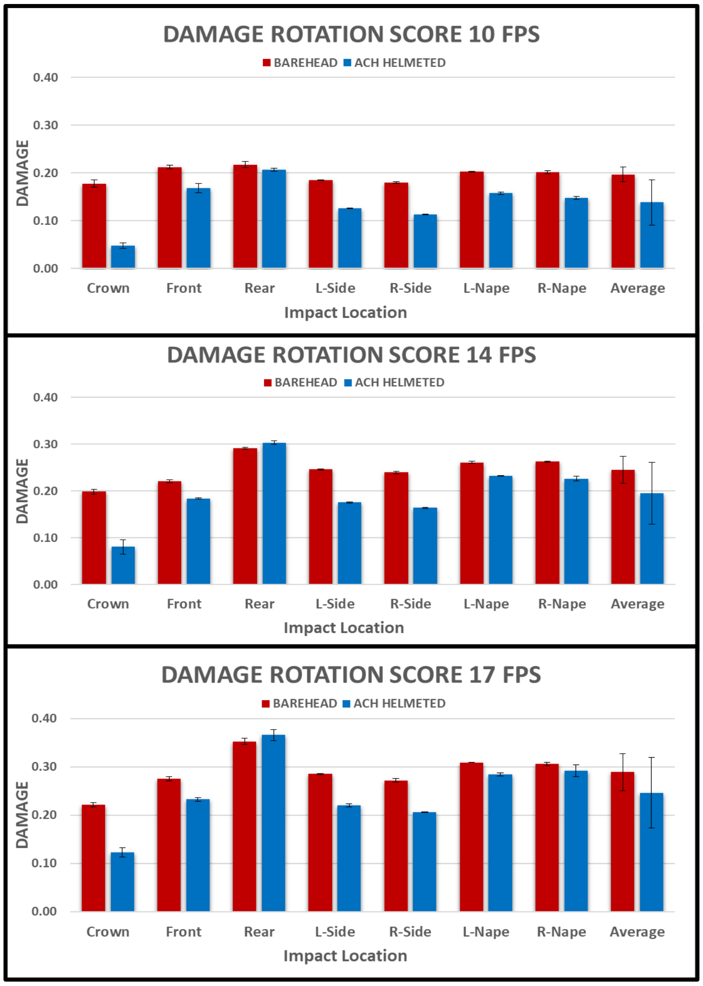

Fig. 8 Comparison of calculated DAMAGE scores for each impact location and velocity. DAMAGE is a metric that predicts maximum brain strain based upon translational and rotational kinematics of the head.

To confirm this assertion, linear regression was used to calculate coefficient of determination (R^2) values for each peak kinematic value (Fig. 9). Peak angular velocity showed the highest correlation by a significant margin compared to peak angular acceleration or peak linear acceleration. Peak angular velocity is a metric that results from the area under the curve of angular acceleration–time traces, giving a better characterization of the entire loading event. This finding may not hold true for other head loading conditions, such as elevated acceleration over longer time durations, which may be observed during automotive collisions. Additionally, the impact vectors delivered in this study were directed in large part toward the head CG. As impacts are delivered further away from the CG, this correlation may change.

It should be noted that these findings run counter to the existing NOCSAE rotational certification standard for football helmets, which evaluates rotational performance in terms of a peak angular acceleration limit, currently set at 6000/rad/s/s (NOCSAE DOC (ND) 081 [NOCSAE 2019]). This research suggests that the NOCSAE standard may consider targeting kinematic performance metrics based upon rotational velocity in addition to rotational acceleration. Furthermore, although analysis suggests rear impacts are the most severe in terms of resulting head kinematics, it does not necessarily make them the most important to the military since it does not consider the frequency of a particular impact location in the Soldier population. Field data that documents the blunt impact conditions leading to head injury can guide the development of future helmet test protocols and standards.

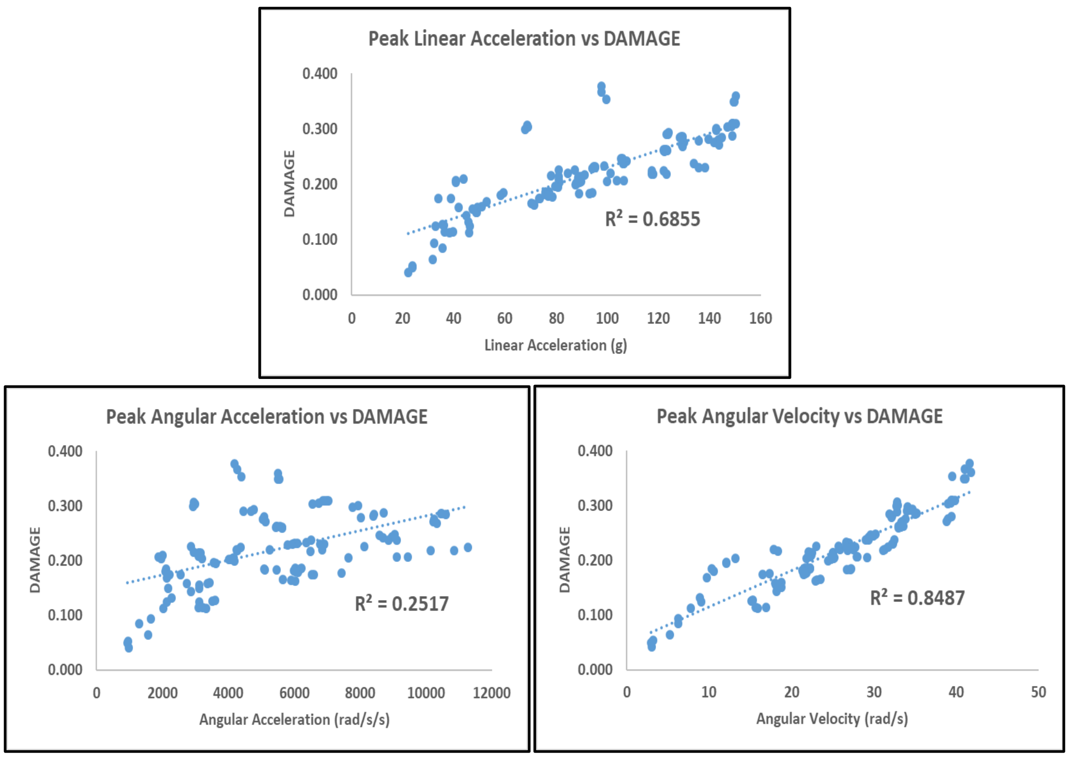

Fig. 9 Linear regression of bareheaded and helmeted headform kinematic response to calculated DAMAGE score

5. Conclusion

Military helmets are evaluated for blunt impact protection based on test methods that seek to limit peak linear acceleration of the head. Translational acceleration is a useful metric to determine the likelihood of head injury such as a skull fracture, and for this reason, combat helmets have proven highly effective at mitigating focal injuries such as these over the years. Consequently, the vast majority of head injuries in the military are nonpenetrating mTBI, without any associated skull fracture. Increasingly medical experts are attributing excessive rotational head motion as a major contributor to mTBI. In reality, head impact events involve components of both linear and rotational loading, which can be simulated in a laboratory through various techniques, including the pneumatic ram method as demonstrated within this study on a representative combat helmet.

The pneumatic ram impact method produces linear and rotational kinematics utilizing a commonly used anthropomorphic test device, the Hybrid III. Impact conditions from the existing Army blunt impact test protocol were transferred to the pneumatic ram test setup. A specific positioning and test matrix was created to serve as a draft method of a rotational blunt impact protocol for combat helmets. A total of 126 bareheaded and helmeted impacts were performed in this test series. Data traces and high-speed video of each impact were analyzed. Peak angular and rotational kinematics were assessed, determining that, on average, the combat helmet reduced peak linear acceleration, peak angular acceleration, and peak angular velocity compared to the bareheaded condition. Diminishing protection was observed in terms of mitigating linear and angular acceleration as impact velocity increased. Peak linear accelerations from the pneumatic ram method were compared to similar impacts in the existing Army blunt impact evaluation test. Peak accelerations from the existing method were significantly higher than the pneumatic ram tests. The higher rigidity of the headform and anvil of the drop tower method make it a more severe test that is well suited for evaluating the integrity of helmet shells to resist denting and prevent skull fracture. However, the head is constrained to move only in the vertical direction and thus does not permit the rotational motion that is characteristic of real-world impact events and believed to contribute to brain injury. DAMAGE, a tool that evaluates rotational kinematic data traces and predicts a corresponding maximum brain strain, was used to evaluate the relative severity of each impact. The DAMAGE calculation suggested that, on average, the combat helmet decreases predicted brain strain compared to bareheaded impacts. This analysis also suggested that the most severe impact location was at the rear of the head–helmet. Linear regression analysis was used to assess the correlation of measured peak kinematic values to predicted brain strain. This analysis found that rotational acceleration was a poor predictor of brain strain, while rotational velocity had a much higher correlation. This finding runs counter to the current NOCSAE rotational helmet certification standard, where the pass/fail metric is based on a peak rotational acceleration threshold (NOCSAE Doc081 [NOCSAE 2019]). Future rotational impact certification standards should consider including performance metrics to limiting peak angular velocity in addition to angular acceleration.

We note the limitations in the biofidelity of the Hybrid III neck used in the pneumatic ram impact method. Though often used in direct impact test scenarios, the Hybrid III neck has only been validated for inertial loading scenarios related to automotive collisions. Additionally, the Hybrid III neck does not replicate the biofidelic response of the human neck in terms of muscle conditioning and activation, which can have a significant influence on head kinematics during impact events (Reynier et al. 2020). Additionally, the analysis contained in this technical report only considers blunt impact-induced brain–head loading and does not consider brain injury resulting from blast exposure and primary effects related to overpressure (Institute of Medicine 2014; Azar et al. 2019).

6. References

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List of Symbols, Abbreviations, and Acronyms

ACH	Advanced Combat Helmet
ATC	Army Test Center
Biocore	Biomechanics Consulting & Research
CG	center of gravity
DAMAGE	Diffuse Axonal Multi-Axis General Evaluation
DOD	US Department of Defense
DOT	US Department of Transportation
DVBIC	Defense and Veterans Brain Injury Center
FMVSS	Federal Motor Vehicle Safety Standard
MPS	maximum principal strain
mTBI	mild traumatic brain injury
NAP	nine-accelerometer array package
NFL	National Football League
NOCSAE	National Operating Comm. on Standards for Athletic Equipment
PEO	Program Executive Office
TBI	traumatic brain injury
ZAP	Zorbium Action Pad

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