Hybrid III and THUMS Headforms Comparison for EOD Helmet Blast Mitigation Performance

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Abstract. Historically, blast overpressure protection of Explosive Ordnance Disposal (EOD) Personal Protective Equipment (PPE) was assessed through full-scale blast experiments, with and without PPE. Existing injury criteria, sometimes from adjacent fields, were applied often without proper validation for blast scenarios. Over the last decade, advanced physical surrogates have been developed, focusing on the blast response of the human body, allowing for measurements that are more representative of an actual human head. Unfortunately, these advanced physical surrogates are complex and rarely available beyond the laboratories where they have been devised. Irrespective of the biofidelity of physical surrogates, blast testing of EOD PPE is challenging due to the severe threat and the inherent variability of blast. Thankfully, numerical models of the human body provide more insight on the body response and broaden the types of measurements reported. Importantly, the potential benefits of PPE can also be investigated with such human models, so long as the PPE itself is properly modelled (rate-dependent material properties, interaction with physical surrogates, sufficient resolution). The current study quantifies the performance of EOD helmets at mitigating blast overpressure, using data obtained from 1) blast experiments with physical Hybrid III mannequins, 2) computational simulation with a numerical Hybrid III model, and 3) computational simulation with a biofidelic human model. Numerical simulations of both the protected (EOD helmet) and unprotected cases revealed important differences between the two simulated scenarios as well as differences between experimental and numerical results for the Hybrid III case, when comparing common parameters. The computational biofidelic head models used also highlighted challenges in applying existing injury criteria since the exact locations within the head where parameters must be measured are not well-defined. This study is the most advanced numerical investigation to date of the performance of EOD PPE under representative blast loading, involving a human surrogate head.

1. INTRODUCTION

Explosive Ordnance Disposal (EOD) operators expect their EOD Personal Protective Equipment (PPE), also referred to as "bomb suits" (Figure 1) to provide protection from the blast overpressure threat. Unfortunately, there is currently no universally accepted quantitative test methodology for blast overpressure performance testing of bomb suits. Indeed, the US National Institute of Justice NIJ 0117.01 standard for public safety bomb suits [1], released in 2016, only addresses "blast integrity". This consists of observing the capacity of the bomb suit ensemble to resist a blast, from a purely qualitative standpoint. The NIJ rationale for the omission of quantitative requirements is that present research and data related to the effects of blast overpressure (e.g., blast head trauma, blast thoracic injury, blunt thoracic injury, blunt lower neck trauma, other neck injury, and blast ear injury) are limited. However, given that one of the main roles of bomb suits is to protect against blast overpressure, it remains highly relevant to quantify their blast overpressure mitigation performance, to ensure that end-users do not end up donning a poorly designed bomb suit, not providing sufficient blast overpressure protection.

Figure 1. Bomb suit for Explosive Ordnance Disposal

In the absence of a widely accepted quantitative blast overpressure standard, bomb suit manufacturers typically quantify PPE protection through percentage reductions in engineering variables measured on anthropomorphic mannequins, with and without bomb suit protection. The variables include head acceleration, as well as ear and chest overpressure, without any direct link to injury potential being provided. Dionne et al. [2] conducted a statistical analysis of the experimental blast overpressure test results related to these three variables. It was assumed that a reduction in engineering parameters measured on mannequins, must correlate with a reduction in blast injuries.

The Hybrid III mannequin mentioned in the NIJ standard [1] and by Dionne et al. [2] has only been validated for automotive crash tests. As such, its applicability for blast overpressure testing is of much debate. On the other hand, numerous human surrogates developed specifically for blast applications have been developed and tested over the last decade. The Warrior Injury Assessment Manikin (WIA Man)

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was developed for military vehicle under-belly blast testing [3]. This surrogate is aimed at quantifying vertical loading and human extremity response and is thus not suitable for bomb suit blast overpressure testing. Other suitable surrogates have also been developed by other groups, such as the Human Surrogate Head Model (HSHM) [4] and the Brain Injury Protection Evaluation Device (BIPED) [5]. However, these advanced blast surrogates tend to be expensive, possibly frangible, and are not standardized (at least not yet). Moreover, they are not readily available for purchase by industry, making their suitability for the severe EOD tests questionable. As a result, bomb suit manufacturers still rely on the Hybrid III mannequin (or equivalent) to characterize the protection performance of their products.

An alternative to blast overpressure experimental testing is to conduct numerical simulations. Indeed, computational modelling and simulation techniques have been used to study blast induced traumatic brain injury (TBI) and investigate the complex biomechanical and physiological factors leading to injury. For instance, Lockhart [6] implemented a rigid-body model (GEBOD) in LS-Dyna to compute the head response in blast scenarios. The head acceleration based HIC15 criterion was applied to investigate the effects of a PASGT helmet. Furthermore, a 2D sagittal biofidelic head model was used to explore overpressure distribution around the head. Unfortunately, no injury parameters at the brain tissue level were studied. Addressing this gap, Nyein [7] developed a 3D biofidelic human head model to investigate the effects of a military ACH helmet on the propagation of stress waves within the brain. Specifically, changes in intracranial pressure due to the ACH helmet were studied. More recently, Yu and Ghajari [8] implemented a high-fidelity human head model to study the effects of an ACH helmet worn with goggles on the head response to blast. It is reported that this protective gear led to increases in intracranial pressure (ICP), cerebrospinal fluid (CSF) cavitation, as well as brain strain and strain rate, compared to the unprotected case. The above studies all focused on military helmets aimed at protecting from blunt impacts and ballistic penetrations, not blast overpressure ingress.

Valverde-Marcos et al. [9] conducted an extensive study of the protective capability of an EOD helmet for small blasts, also using computational models. They used the HHFEM (Human Head Finite Element Model) model developed by J. Antona-Makoshi [10] and modelled an existing EOD helmet used by the Spanish police. In comparison with the unprotected case, the EOD helmet was found to delay the impact of the shockwave on the wearer's head and reduced the maximum head acceleration by 80% in all three cases simulated. Comparing to relevant published injury thresholds, they concluded that wearing an EOD helmet reduced the severity of injuries from a highly probable death (when unprotected) to a low probability of injury, of a mild and localized nature. It must be emphasized however that these findings were obtained through simulating relatively low explosive charges. In addition, the simulation model developed by Valverde-Marcos et al. [9] looked at the EOD helmet in isolation. Indeed, no interaction with an EOD suit was modelled.

In the present study, numerical simulations of the EOD helmet mitigation performance were conducted against the representative explosive charge described in the NIJ 0117.01 bomb suit standard. Even though far from biofidelic, a numerical Hybrid III head and neck model (Figure 2) [11] was used for a first set of numerical simulations, with the purpose of directly comparing with experimental results obtained with that same surrogate. For these simulations, an EOD helmet and an EOD suit (including blast-protecting collar) were modelled, both in terms of geometry and material properties.

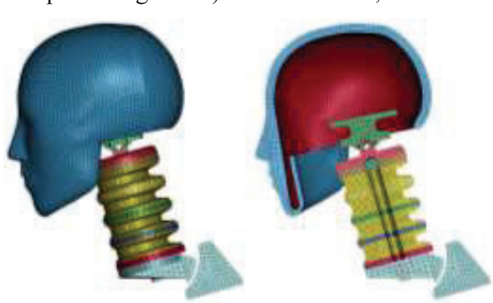

Figure 2. Hybrid III head and neck model [11]

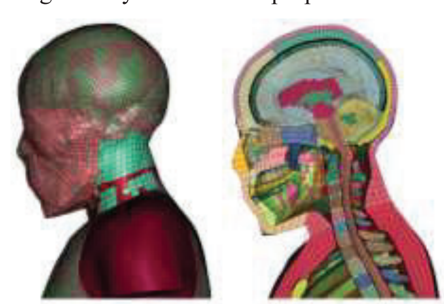

Figure 3. THUMS model [12], focusing on the head/neck portion

Simulations were conducted both in the protected (EOD helmet and suit) and unprotected scenarios. The work on the numerical Hybrid III model was funded in part by the US Army (2017-19) with an objective to get insight into the protection capabilities of EOD helmet protection concepts and validate the numerical EOD Helmet models. Similar simulations were then performed using a much more advanced head model (THUMS, Figure 3, [12]), also developed for the automotive industry, but featuring morphologically accurate details of the human head and brain. The rationale for using the THUMS head model was to quantify the response of actual brain tissues when subjected to blast, through

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parameters having the potential to be linked to injury mechanisms. It must be emphasized though, that to our knowledge, the THUMS model has not been validated for blast. Comparisons were made between predictions from the Hybrid III and THUMS numerical models, in the NIJ standard explosive scenario, but in three different orientations of the head surrogate with respect to the blast: 0° (directly facing), 45° (oblique) and 90° (sideways). To our knowledge, this is the first time EOD helmet response is simulated in orientations other than directly facing the blast.

2. METHODOLOGY

Simulations were generated using LS-Dyna, an advanced general-purpose multiphysics simulation software package developed by the Livermore Software Technology Corporation (LSTC) and owned by ANSYS. A 3D Finite Element Analysis (FEA) model was created for this study, which includes an air domain of 1200 mm by 840 mm by 1200 mm modelled using the Multi-Material Arbitrary Lagrangian Eulerian (MMALE) method.

To reduce computational time and increase the accuracy of the results, a technique that allows the mapping of results from 1D to 3D Eulerian domains has been employed [13]. A 1D spherical symmetry model with an element size of 1 mm was used to model the C4 explosive and its detonation. After the blast wave propagated to the boundary of the 3D air domain, the pressure distribution and particle velocity distribution were then exported to a binary map file. This map file was then used to initiate the 3D domain in the subsequent 3D simulation. The timestep was controlled by the LS-Dyna solver to achieve numerical stability. In Arbitrary Lagrangian Eulerian (ALE) simulations, the timestep is typically less than $1.0e^{-7}$ second, which might not be small enough to capture the real peak in blast waves, but still acceptable in the current context. The simulation data was generated at a rate of 1 MHz. Figure 4 demonstrates the 3D model schematic, where the blast wave comes from the left side.

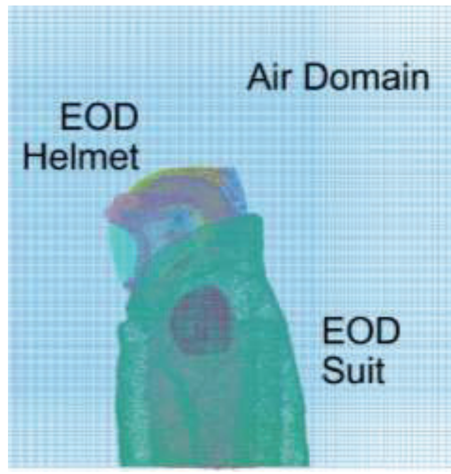

Figure 4. 3D model schematic of the simulation domain

In the 3D model, the Hybrid III mannequin head and neck model and the THUMS model were placed in the middle of the air domain, either with or without EOD protection. The Fluid-Structure Interaction (FSI) method was then applied to transfer the pressure from the air and explosive to the mannequin or the EOD helmet and suit in the protected case.

The EOD helmet model includes the helmet shell, the impact/comfort liner, the retention system, the face shield, and the housing for the electronics. These components were modelled with hexahedron elements using a Lagrange formation. Meshes were generated from CAD models of a Med-Eng developed EOD helmet version. The corresponding material models for these components were deformable. While the model constants were determined experimentally, the geometry of the EOD suit was scanned from a Med-Eng developed EOD suit, with its material properties estimated based on aramid fabric textiles. The purpose of this EOD suit in the simulation was to provide realistic surfaces to generate reflected waves that eventually influenced the head response. Based on the experimental studies conducted in the past, the reflected waves from the EOD suit, especially the collar, significantly modify the loading on the EOD helmet, and subsequently the head response [14].

All simulations were conducted to generate over 8 ms of data. This duration is sufficient to capture the original motion of the mannequin and PPE, given the absence of reflecting surfaces. Ground reflections would occur later and would induce response levels lower than for the original blast wave impact. Table 1 summarizes all simulations conducted (test matrix).

For both Hybrid III and THUMS cases, the global head acceleration was tracked. For the Hybrid III, an accelerometer positioned at the centre of gravity of the head is always used for the purpose of this measurement. On the other hand, the THUMS mannequin is deformable and no set location within the head model is dedicated for global head acceleration tracking. Ideally, the global THUMS head acceleration would have been obtained by computing the location of the centre of gravity at each time interval. But to simplify the calculations, a specific unique location, similar to the position of the accelerometer in the Hybrid III head, was determined as its centre of gravity. The head acceleration was tracked at that location throughout the event duration. With the head acceleration data available for both

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head models, the Head Injury Criterion (HIC) was then calculated. In addition to the standard definition of the HIC15 (calculated over a maximum 15 ms duration), a version referred to as "HIC15d" was also calculated. The HIC15d is better suited when using head surrogates attached to mannequin bodies [15]. The equations for the HIC15 (free floating headform) and HIC15d (attached headform) are:

$$HIC15 = \frac{1}{(t_1 - t_2)} \int_{t_1}^{t_2} a_{res}^{2.5} \, (t_2 - t_1) dt \qquad HIC15d = 0.75446(HIC15) + 166.4$$

As the THUMS head model includes realistic human features, additional parameters such as the intracranial pressure (ICP), the cerebrospinal fluid pressure (P-CSF), as well as the cerebellum effective strain (ε_{eff}), were extracted. To determine the optimal location within the head to extract these measurements, a first simulation was first conducted, from which the approximate maximum location (for a given parameter) could be visually determined. A follow-on simulation then tracked the parameter at this selected location. The measurement locations thus varied according to the parameter being measured, and for each combination of orientation and protection configuration. Detailed results are presented below, for the three orientations with respect to the blast: front facing (0°), oblique (45°) and sideways (90°), for all variables of interest (head acceleration, intracranial pressure, cerebrospinal fluid pressure and cerebellum effective strain.

Table 1. Test matrix for all numerical simulations conducted

Headform	Protection	Orientation
Hybrid III	Unprotected	0°
		45°
		90°
	EOD Helmet	0°
		45°
		90°
THUMS	Unprotected	0°
		45°
		90°
	EOD Helmet	0°
		45°
		90°

3. RESULTS

3.1 Front facing (0°)

Figure 5 illustrates the numerical simulations conducted for all scenarios in the 0° orientation. Peak values for all parameters are listed in Table 2. Finally, detailed traces are provided in Figures 6 to 10.

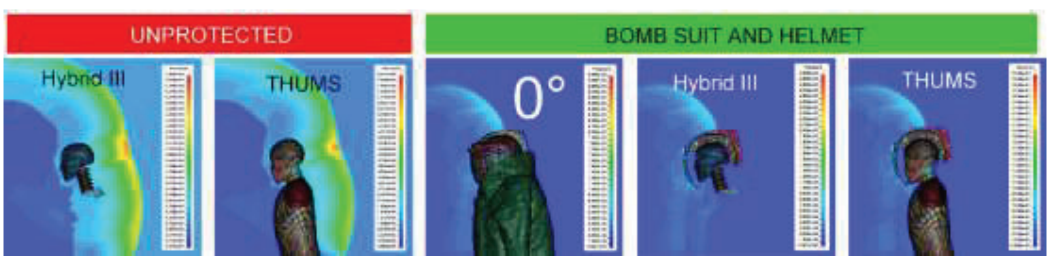
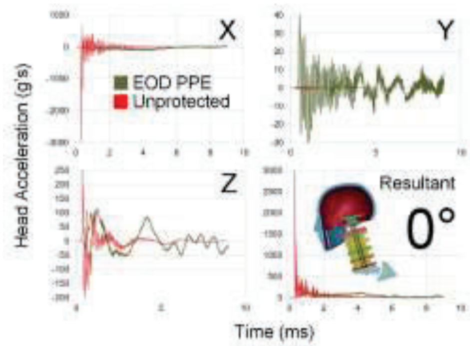


Figure 5. Images from numerical simulations at 0° (Hybrid III and THUMS, unprotected and EOD)

Table 2. Results (peak values and percentage reductions) obtained at 0° orientation

0°	Hybrid III			THUMS							
	a_{max} (g's)	HIC15	HIC15d	a_{max} (g's)	HIC15	HIC15d	ICP_{max} (kPa)	ICP_{min} (kPa)	P-CSF$_{max}$ (kPa)	P-CSF$_{min}$ (kPa)	ε_{eff}
Unprotected	2985	8648	6691	1967	18470	14100	2692	-2432	2894	-2627	0.15
EOD Helmet	130	258	361	305	266	367	210	-89	193	-123	0.20
% Reduction	95.6%	97.0%	94.6%	84.5%	98.6%	97.4%	92.2%	96.3%	93.3%	95.3%	-36.0%



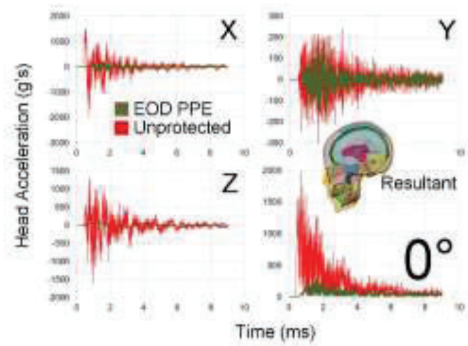


Figure 6. Hybrid III X, Y, Z, & Resultant head acceleration traces in the 0° orientation

Figure 7. THUMS X, Y, Z, & Resultant head acceleration traces in the 0° orientation

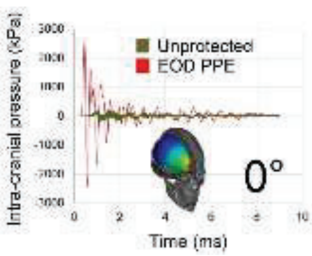


Figure 8. THUMS IC pressure in the 0° orientation

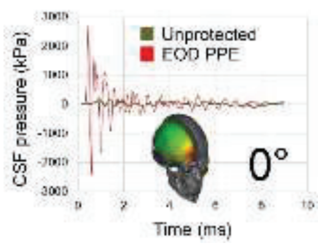


Figure 9. THUMS CSF pressure in the 0° orientation

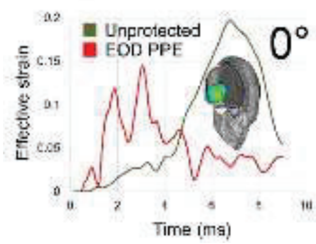


Figure 10. THUMS effective strain in the 0° orientation

3.2 Oblique (45°)

Figure 11 illustrates the numerical simulations conducted for all scenarios in the 45° orientation. Peak values for all parameters are listed in Table 3. Finally, detailed traces are provided in Figures 12 to 16.

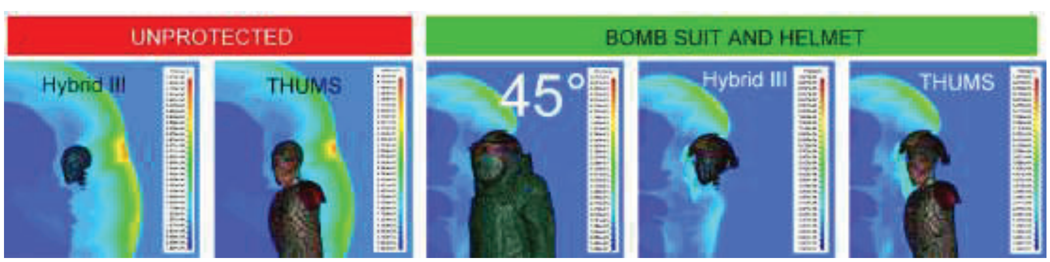


Figure 11. Images from numerical simulations at 45° (Hybrid III and THUMS, unprotected and EOD)

Table 3: Results (peak values and percentage reductions) obtained at 45° orientation

45°	Hybrid III			THUMS							
	a_{max} (g's)	HIC15	HIC15d	a_{max} (g's)	HIC15	HIC15d	ICP_{max} (kPa)	ICP_{min} (kPa)	$P\text{-}CSF_{max}$ (kPa)	$P\text{-}CSF_{min}$ (kPa)	ε_{eff}
Unprotected	2670	7722	5992	1846	18950	14470	3092	-2053	3595	-2789	0.23
EOD Helmet	121	176	299	293	234	342.9	164	-99.5	154	-101	0.05
% Reduction	95.5%	97.7%	95.0%	84.1%	98.8%	97.6%	94.7%	95.2%	95.7%	96.4%	79.6%

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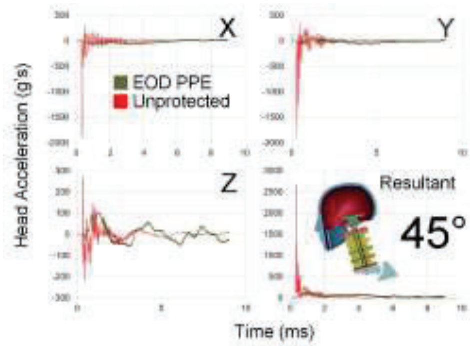

Figure 12. Hybrid III X, Y, Z, & Resultant head acceleration traces in the 45° orientation

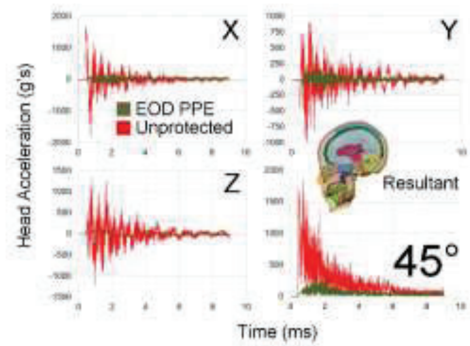

Figure 13. THUMS X, Y, Z, & Resultant head acceleration traces in the 45° orientation

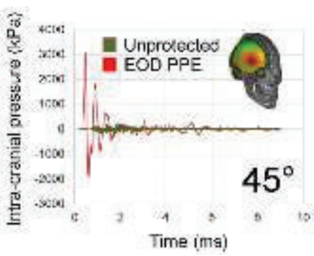

Figure 14. THUMS IC pressure in the 45° orientation

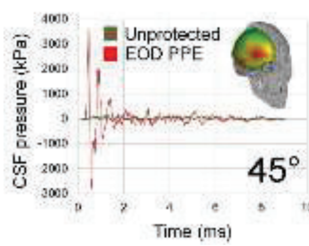

Figure 15. THUMS CSF pressure in the 45° orientation

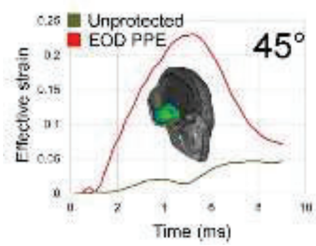

Figure 16. THUMS effective strain in the 45° orientation

3.3 Sideways (90°)

Figure 17 illustrates the numerical simulations conducted for all scenarios in the 90° orientation. Peak values for all parameters are listed in Table 4. Finally, detailed traces are provided in Figures 18 to 22.

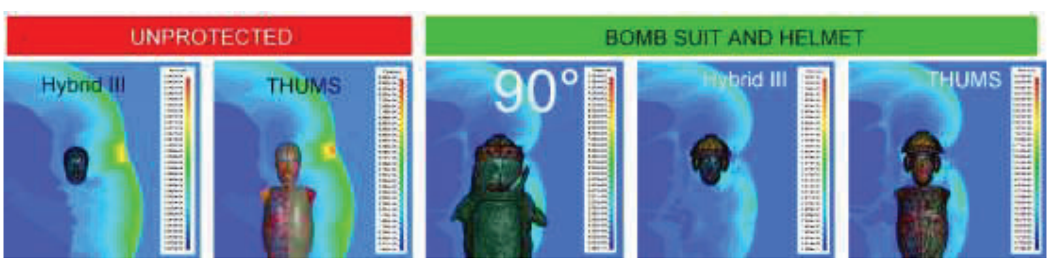

Figure 17: Images from numerical simulations at 90° (Hybrid III and THUMS, unprotected and EOD)

Table 4: Results (peak values and percentage reductions) obtained at 90° orientation

90°	Hybrid III			THUMS							
	a_{max} (g's)	HIC15	HIC15d	a_{max} (g's)	HIC15	HIC15d	ICP_{max} (kPa)	ICP_{min} (kPa)	$P\text{-}CSF_{max}$ (kPa)	$P\text{-}CSF_{min}$ (kPa)	ε_{eff}
Unprotected	3361	12400	9521	3233	43500	32980	4937	-2787	5500	-3391	0.11
EOD Helmet	173	171	296	533	944	803	205	-137	268	-190	0.10
% Reduction	94.9%	98.6%	96.9%	83.5%	98.1%	97.6%	95.9%	95.1%	95.1%	94.4%	10.1%

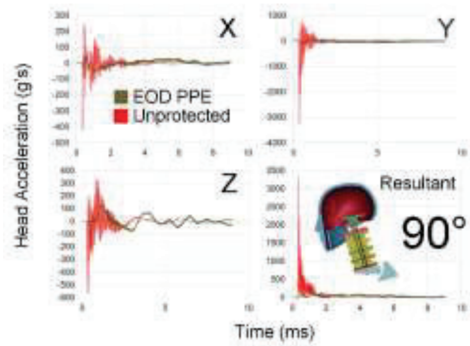

Figure 18. Hybrid III X, Y, Z, & Resultant head acceleration traces in the 90° orientation

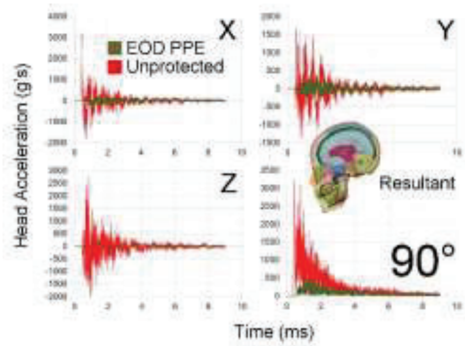

Figure 19. THUMS X, Y, Z, & Resultant head acceleration traces in the 90° orientation

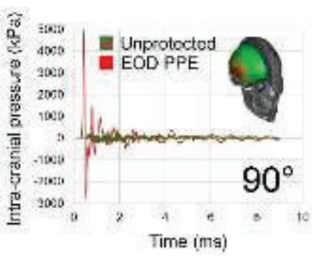

Figure 20. THUMS IC pressure in the 90° orientation

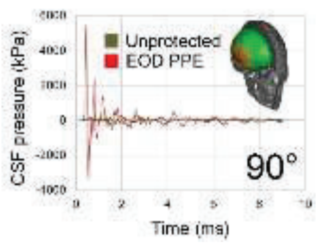

Figure 21. THUMS CSF pressure in the 90° orientation

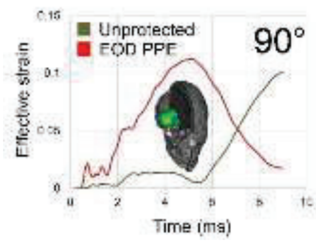

Figure 22. THUMS effective strain in the 90° orientation

4. DISCUSSION

Figure 23 tabulates the peak resultant head accelerations and HIC15d from the Hybrid III and THUMS models at all three mannequin orientations, with and without an EOD helmet. The results highlight that the PPE dramatically reduces the head acceleration and its derived HIC15d values, thus suggesting that the PPE provides effective protection to the EOD technicians. Moreover, this reduction is consistent across all three orientations for both Hybrid III and THUMS models. This insensitivity of results with respect to the orientation of the mannequin implies that the overall mass (inertia) of the head and helmet dominates over the kinematics of the head itself. Nevertheless, there are some notable differences in aerodynamic loading, with higher peak values for the 90° orientation, compared to the other two cases. This is due to the larger projected area inducing increased drag for the side (90°) exposure.

In terms of intracranial pressure (ICP), Figure 24 indicates that the *unprotected* peak pressures vary with orientation. While in the 0° and 45° cases, the maximum pressures are similar (approximately 2800 and 3000 kPa, respectively), a value nearing 5000 kPa was obtained in the 90° orientation. This large difference in the 90° case is due to the grey matter cerebrum not being spherically symmetric, implying that locations and values for the maximum ICP vary with the orientation. It must be kept in mind that peak acceleration values are very sensitive to the high frequency responses since the blast wave has a very short (almost zero) duration in the initial pulse around the peaking time. In contrast, the EOD helmet induced a significant reduction down to approximately 200 kPa for all cases, with differences in absolute pressures being modest across all orientations. The hard helmet shell and soft impact liner prevent a direct exposure of the head to the blast wave. The low ICP variations in the EOD case are thus a direct result of the interaction between the helmet and the head. Since the shock wave reflecting from the helmet outer surface remains at a similar level for all orientations (single blast charge and standoff), the orientation only exerts a small relative influence on the helmet kinematics. Consequently, the peak ICP values for the EOD case remained within a relatively limited range, likely below any injury threshold.

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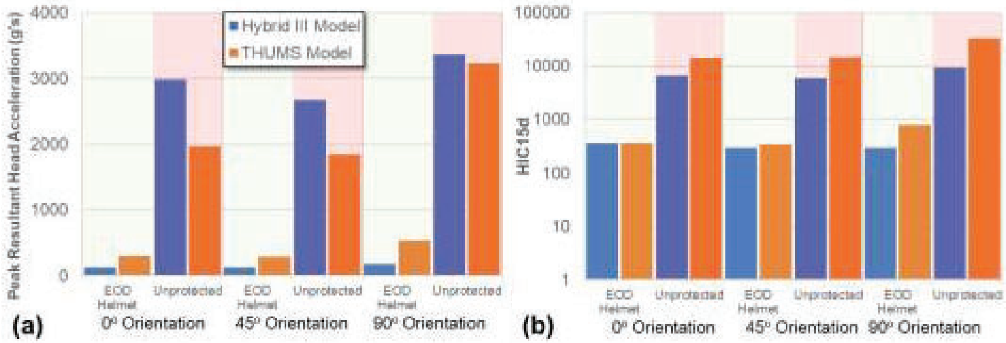

Figure 23. Comparison of the Hybrid III and THUMS models in terms of (a) peak head acceleration, and (b) HIC15d, in all three mannequin orientations (0°, 45°, and 90°) for both unprotected and protected (EOD) cases

Similar to the ICP, the cerebrospinal fluid pressure (P-CSF) also showed divergent peak pressures and locations based on the mannequin's orientation to the blast. Indeed, maximum CSF pressures of approximately 2900 kPa, 3800 kPa and 4900 kPa were noted in the 0°, 45°, and 90° orientations, respectively. Again, the EOD helmet reduced the pressure considerably down to approximately 200 kPa, in all three orientations. The reason for this phenomenon is the same as discussed above for the ICP.

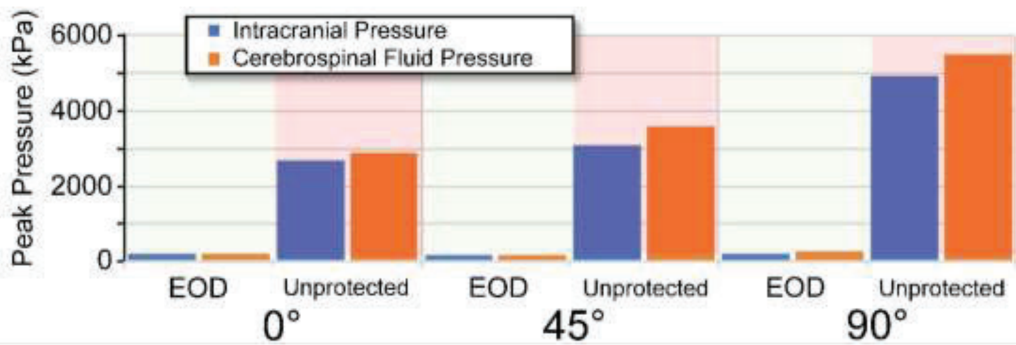

Figure 24. Comparison of intracranial and cerebrospinal fluid pressures with the THUMS model in all three mannequin orientations (0°, 45°, and 90°) for both unprotected and protected (EOD) cases

As for the cerebellum effective strain, the ability of the EOD helmet to reduce peak values is not consistent (Figure 25). Only in the 45° orientation was the strain reduction (80%) on par with those observed for the intracranial and cerebrospinal pressures. In the 90° orientation, the EOD helmet yielded a mere 11% reduction, compared to the unprotected. Finally, in the 0° orientation, the cerebellum effective strain was noted to *increase* by 36% with the EOD helmet. However, it should be noted that the time to reach maximum strain is much longer when wearing a helmet. Moreover, the rising rate of the cerebellum effective strain is dramatically reduced when introducing protection. Focusing on just peak strain might therefore not draw the complete picture. Unlike stress or pressure, the effective strain is a cumulative parameter and thus depends on the loading duration causing plastic deformation. As the cerebellum tissue has a low yield stress, accumulated plastic strain behaves completely different as a parameter, compared to the ICP and CSF pressure.

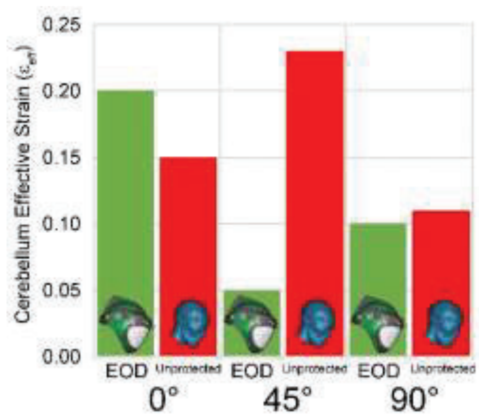

Figure 25. Comparison of cerebellum effective strain with the THUMS model for all three mannequin orientations (0°, 45°, and 90°) for both unprotected and protected (EOD) cases

4.1 Comparison with Experimental Results

As stated earlier, full-scale experimental blasts tests were also conducted with a physical 50th percentile Hybrid III mannequin for comparison with the modelling results. The experimental trials saw the mannequin placed in the same NIJ 0117.01 standard configuration [1], i.e., 60 cm away horizontally from a 0.567 kg C4 explosive at a 77 cm vertical height of burst. The experimental head accelerations were acquired at a sampling rate of 200 kHz with CFC1000 anti-aliasing filtering (1650 Hz cut-off frequency). Experimental trials were only conducted at a 0° orientation, in accordance with the NIJ standard requirements. Figure 26 compares the average and range of peak head acceleration and HIC15d for both the unprotected and protected (EOD) cases. Both numerical models (Hybrid III and THUMS) were found to overpredict the peak head acceleration and HIC15d values. Most simulated peak values nevertheless fall within the range of experimental data, with the exception of the head acceleration results in the EOD/THUMS case, and for both head models for the unprotected case.

The lower peak accelerations observed in the experimental data is likely due to filtering effects. Indeed, filtering at 1650 Hz leads to smoothing of the sharpest peaks in head acceleration signals. The HIC algorithm on the other hand, which incorporates the integration of the acceleration signal, is not as sensitive to filtering effects. As such, differences between experimental and simulated HIC15d values are not as significant, especially considering the scatter in experimental data. The high level of scatter in the experimental data is common to blast testing using Hybrid III mannequins, as previously reported by Dionne et al. [2].

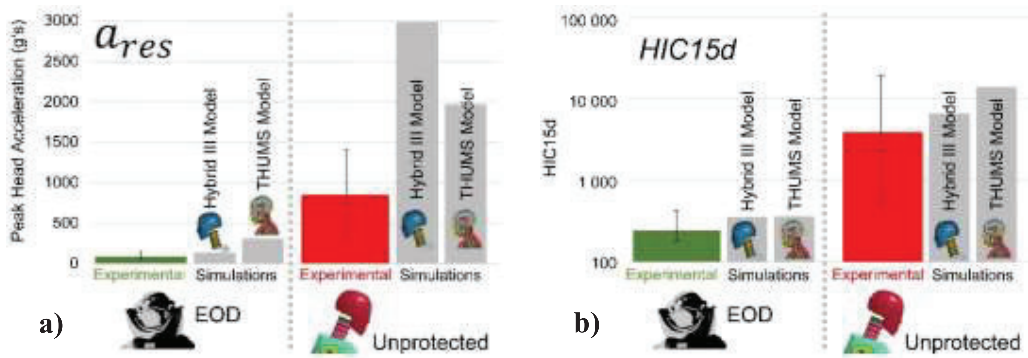

Figure 26. Average and range of the experimental (a) peak head accelerations, and (b) HIC15d, for both EOD helmet (green) and unprotected (red) experimental trials. Also denoted are the extracted values from the Hybrid III and THUMS simulations (grey)

5. CONCLUSION

The results obtained with the Hybrid III and THUMS numerical models do not show a very strong match with the overall experimental results (Figure 26), but among those, the protected (EOD) scenarios show a more favourable match, with the numerical results being more conservative. The use of numerical simulations to conservatively evaluate the effectiveness of EOD PPE (bomb suits) in configurations other than those tested experimentally is therefore promising, based on these results. In particular, the current study explored the effect of orientation, which is of interest to the EOD community, given that despite all efforts to follow standard operating procedures, EOD technicians do not always directly face an explosion. In particular, the 90° orientation for the EOD helmet differed from the other two cases, with higher acceleration and cerebrospinal fluid pressure values, thought to be due to a larger exposed surface area. For the Head Injury Criterion (HIC) though, much higher values were predicted for the more anatomically accurate THUMS model, compared to the Hybrid III, especially at 90° for the EOD helmet case, and in all directions for the other two orientations. The THUMS therefore yielded more conservative predictions for head injury, which is deemed preferable when designing protective equipment. Moreover, the THUMS advanced anatomical model allowed for the measurement of additional parameters (e.g., intracranial pressure, cerebrospinal fluid pressure, cerebellum effective strain) at various locations within the brain, helping to draw a more complete picture of the brain response under blast loading, and the role of blast protective helmets. Indeed, for all parameters measured, with the exception of the effective strain, substantial reductions (above 80% and often exceeding 95%) resulted from the presence of an EOD helmet.

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Numerical simulations could also be conducted to investigate a wider range of explosive configurations (charge size, standoff, technician posture), at a much-reduced cost, compared to experimental trials. It is thus hoped that such numerical simulations could eventually guide and optimize EOD helmet design, when conducted in parallel with physical helmet development. However, further efforts will be required towards validating the numerical human body models and PPE models for blast, before numerical simulations can play a substantial role in the design of blast protective PPE.

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