29th International Symposium on Ballistics, 9-13 May 2016, Edinburgh, Scotland, UK.
https://www.ballistics.org/29th_isb.php

The Study of a Human Head Simulant's Dynamic Response to a Blast Wave

Benjamin Harington, Alan Peare, Ian Horsfall

Impact and Armour Group, Centre for Defence Engineering, Cranfield University,
Defence Academy of the United Kingdom, Shrivenham, SN6 8LA, UK.

The prevalence of body armour and helmets in military forces combined with the availability of combat medical support and timely evacuation of injured soldiers has increased the survivability rates of those who have been exposed to blast. Despite this, the incidents of traumatic brain injury (TBI), as a result of primary blast, have been described as the 'signature injury' of modern warfare. The physical interaction between a blast wave and a human head is not well understood and there is some conjecture as to whether helmets are attenuating or amplifying the blast effects on the human head. The aim of this study was to improve the understanding of the interaction of primary blasts on the human head with different attachments such as a helmet and face shield.

INTRODUCTION

Wherever armies have developed weapons to engage in war, combatants have in turn developed personal armour to protect themselves against the prevailing threat. The relative effectiveness of the weapon and armour systems has varied over the years as advancements have been made in each. The First World War saw the threat of long range artillery drive a requirement for a ballistic protective helmet that would protect against high velocity projectiles. The use of helmets by western armies is now prolific and the ballistic protection the helmets offer is quite well understood.

The current asymmetric warfare experienced in Iraq and Afghanistan has seen a focus on blast weaponry such as Improvised Explosive Devices (IEDs) accounting for 75% of all combat casualties in the US [1]. Survival rates of combatants exposed to IEDs have increased considerably due to improvements in ballistic protection of modern body armour, deployable medical facilities and faster evacuation rates. Notwithstanding, there is a prevalence of veterans surviving such explosions then suffering from Traumatic Brain Injury (TBI) [2,3].

In particular, blast-induced TBI has been described as a ‘signature injury’ of modern combat [1,4,5,6,7]. Understanding blast and the mechanisms by which the energy from blasts is transferred to the head and brain is not well understood, particularly how it relates to TBI [4,8].

Increased understanding of TBI is critical to modern military forces as they seek to improve survival rates and minimise injury rates of veterans. The current ballistic armour and helmet systems have sound specifications relating to their ballistic properties, but there is a prevailing omission of specifications relating to blast protection. Without specifying blast protection in the armour systems, there will be a continued shift in the pattern of blast injury from the lungs to the head [9]. Understanding TBI has been one of the leading challenges for the US and other western militaries [8].

The understanding of this area of study is complex and challenging with an ongoing academic debate concerning some of the observed and measured phenomena. An example of such a debate concerns the effect ballistic helmets have on primary blast; some simulations and tests indicate that helmets may actually amplify the primary blast impact on the human head whilst other studies counter this conjecture [10]. Adding to the complexity are the ethical issues regarding testing. There are many studies that attempt to understand the problem using other methods of testing including computer simulation, testing on animals and cadavers, analysis of combat data and the use of mechanical simulants.

The primary aim of this study was to add to the understanding of primary blast effects on the human head with various layers of protection. There is also a need to understand how structures such as skin, eyes, nose, helmet and face shield effect the size and duration of pressure and strain experienced within the brain.

BACKGROUND

Combat Helmets and Face Shield

Modern combat helmets are ubiquitous within western armies and their design and the ballistic protection offered are quite similar to each other. Historically, combat helmets have been designed to increase survivability through prevention of ballistic injury [7]. Advances in traditional injury protection have seen greater survivability, however, there has been an associated increase in Mild Traumatic Brain Injury (mTBI) [4]. It has not yet been established whether or not this is a causal link. Merkle et al. observed that helmets have been shown to increase the Intra Cranial Pressure (ICP) measured within the brain [5] which is a source of ongoing debate and research.

Whilst helmets on their own can attenuate or magnify the peak pressure of the waveform, studies to date have indicated that face shields will attenuate the peak pressure in the order of 75% [5]. The study conducted by Merkle et al. suggested that the face shield attenuates the pressure because it changes the profile of the face and so reduce the drag coefficient.

The Human Head and Simulants

Describing the human head quantitatively is difficult. There is great variability in age, race and gender that can cause a large standard deviation in measured data. The cranium thickness and geometry varies throughout the skull. The human brain weighs about 1.2kg, sits within the skull, and ‘floats’ within cerebrospinal fluid primarily to protect the brain against mechanical shocks [4]. Brain tissue has been shown to be nearly incompressible and viscoelastic. There are several mechanisms by which the brain can be injured including absolute positive and negative pressures, sudden pressure changes and relative movement of the head an brain.

Designing a Human Head Simulant

Use of the Anthropomorphic Test Device (ATD) or the Hybrid III has limitations when exploring blast impacts on the human head [5]. Principally it is unable to completely match the mechanical properties of the human head when exposed to a blast wave as was observed by Goatman [11]. Merkle et al. [5] broke the human head down into five components, each with different mechanical properties as follows: skull, skin, eyes, brain and face. With these points in mind, it was decided to construct a head simulant similar to that developed by Merkle et al. The design of the neck and mounting was a point of consideration. Merkle et al. and Goatman [5,11] mounted their head simulant on a responsive neck, but this is not necessarily an accurate representation of a combatant exposed to a blast. Another consideration for a head simulant is the location of the sensors and whether these will interfere with the true results if the sensors were not present. Mediavilla Varas et al. [3] tested the effect of sensors with computer modelling, compared to physical results and concluded that the sensors such as pressure and strain gauges are ‘limited and only introduces a high frequency component’. In this same study, a skull brain surrogate was constructed with a small opening at the base to represent the spinal column where the brain joins the grey matter within the spine. Mediavilla Varas et al. tested whether the deformation in the opening was representative of the pressure of the blast as well as the orientation of the head simulant [3]. These simulations and physical tests showed that orientation is a large factor in the pressures recorded.

Blast Injury Mechanisms

Head injuries have historically been attributed to either direct impact or rapid acceleration or deceleration. Blasts expose the head to a short duration, large magnitude rise in overpressure [5]. The effects and mechanisms of blast on the human body can be divided into four (sometimes five) distinct areas [4,6]: primary, secondary, tertiary, quaternary and quinary blast injuries. Primary blast effects were the only effects that were analysed in this study. The primary blast effects are where a pressure wave or shock wave interacts with the body’s air-filled organs such as lungs and ears [6,8,12,13]. Historically, primary blast injuries have been considered an ‘injury of the dead’ as the casualty has been typically killed by other means. However, this statement is no longer true of modern combatants wearing ballistic protective equipment [12]. Primary blast effects are currently the primary cause of TBI for military personnel from western armies [2]. The impact of primary blast effects on air containing organs are reasonably well understood, but the impact on the brain is not [2,6]. The brain has displayed a greater tolerance than the lung for primary blast injury for fatalities, although for non-fatal injuries, the incidents of mTBI can occur well below the 50% fatality risk level for lung injury [2].

The primary blast can be split into two phases: the kinetic response and kinematic response. The kinetic phase lasts up to 10ms and is characterised by a large linear acceleration and rise in ICP with limited head motion [5]. The kinematic phase is longer (in the order of hundreds of milliseconds) and is characterised by global head motion and relative motion of the brain with respect to the head. Previously, most studies of combat Personal Protective Equipment (PPE) have focused on the second kinematic phase [5].

A typical blast pressure wave has an initial, rapid increase in pressure which is often cited as the cause of primary blast injury [12]. The peak pressure and the time duration have a strong correlation to the injury rates of those exposed. One set of data of primary blast lethality rates have been Bowen curves. These curves were based on data collated in 1968 from a variety of sources and were presented to be the 'estimate of man tolerance to the direct effects of air blast' [12]. Bowen curves consider the lethality of a blast as a function of the peak pressure and the duration of the positive phase [13]. An update of the curves was attempted by Bass based on the same data, however Bass did not apply any distinction between standing in a free field or in front of reflecting wall [13].

METHOD

Creating the Head Simulant

The head simulant in the present study was designed to characterise the interaction between blast and the human head when used in conjunction with blast loading from a shock tube. It had been observed that both the strain of the skull and ICP is likely to be linked to TBI so the head simulant was required to have both strain gauges and pressure sensors to measure the effect of the blast. The criteria to create such a head simulant were: -

- being mechanically similar to a human head; specifically the size, density and compressibility of the skull and brain.
- would not change over time and would not deteriorate after each blast exposure.
- being able to have sensors such as pressure gauges and strain gauges located within the simulant.

The head would comprise of the following components: - skull, brain, skin, eyes, cheeks and nose; the skull being an off-the-shelf cast of a real skull and made from PX5210 isocyanate, as shown in Figure 1. The brain was made from Perma-Gel®; a synthetic, non-toxic, medium that is a substitute for ordnance gelatine and commonly used in ballistic research (figure 2).

Figure 1. Skull components.

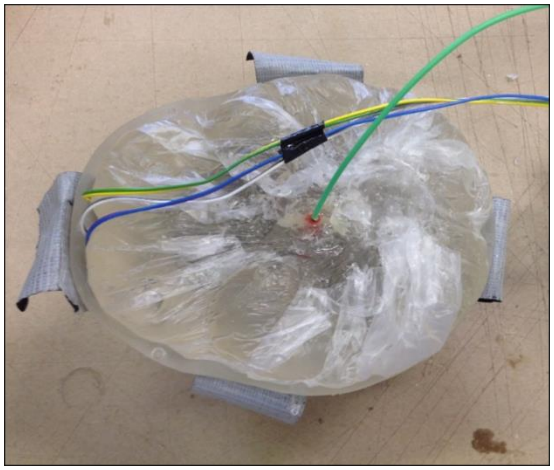


Figure 2. Perma-Gel® brain with pressure sensors in the skull top

Data Capture

A schematic for the overall location of the sensors is shown in Figure 3. Two pressure gauges were used to measure the effect of the blast. The first measured the external reference incidence pressure and was mounted on to the shock tube in line with the head simulant. The second was inserted into the centre of the brain within the skull. As was investigated by Mediavilla Varas et al., one theory to measure the severity of TBI is to measure the amount that the spinal column bulges out of the skull [3]. This is based on the fact that the brain is confined and the only space for it to deform into is the grey matter within the spine. If this were to be true, by measuring the deformation of the Perma-Gel™ at the top of the spine and comparing this with the deformation of the skull, the difference would indicate how much bulging was taking place during a blast. The bulging was measured with two short range laser range finders; one on the spinal column and one on the skull. The head simulant was suspended by four flexible cables and was able to swing freely, as shown in Figure 4.

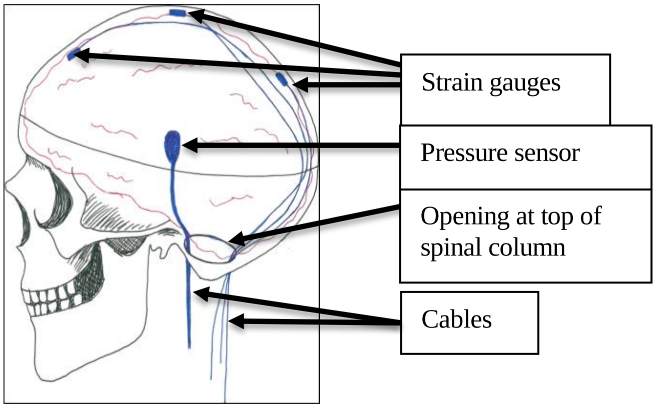


Figure 3. Summary of sensors in the head simulant

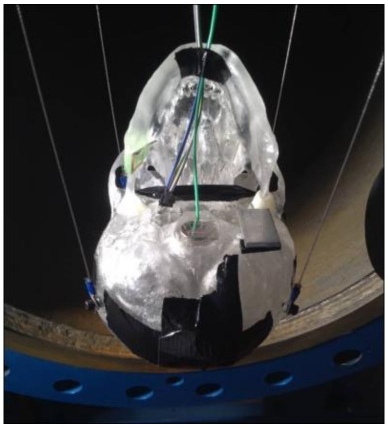


Figure 4. Head simulant suspended in the shock tube

Helmet and Face Shield

The helmet used was a common law-enforcement helmet with a face shield. The selection of the helmet was based on using a helmet with a padded harness and optional face shield. The padding within the harness system was used to hold the head simulant.

Shock Tube

The shock tube used was a bespoke design built specifically large enough to accommodate a head form, see Figure 5, and capable of generating a blast similar to an explosion in the open. The shock tube had an 563mm internal diameter with a 500mm long driving tube and a 4m long driven tube.

Figure 5. The shock tube used to provide blast loading

RESULTS AND ANALYSIS

Pressure Sensor Data

The configuration of the six tests are summarised in Table 1.

TABLE 1. SUMMARY OF TESTS

Test N^{o}s	**Head configuration**	**Attachments**
1, 2 & 3	Naked skull	Nil
4	Skin, cheeks, eyes and nose	Nil
5	Skin, cheeks, eyes and nose	Helmet and face shield
6	Skin, cheeks, eyes and nose	Helmet only

The internal brain pressures for all six blasts can be seen in figure 6 and can be compared to the external pressure (side on static overpressure in the free field). The internal pressure characteristics of each test are summarised in Table 2. It can be seen that the brain responds to the external pressure and continues to oscillate for over 200ms. The magnitude of the internal pressure oscillations vary within ≈ ±0.5bar.

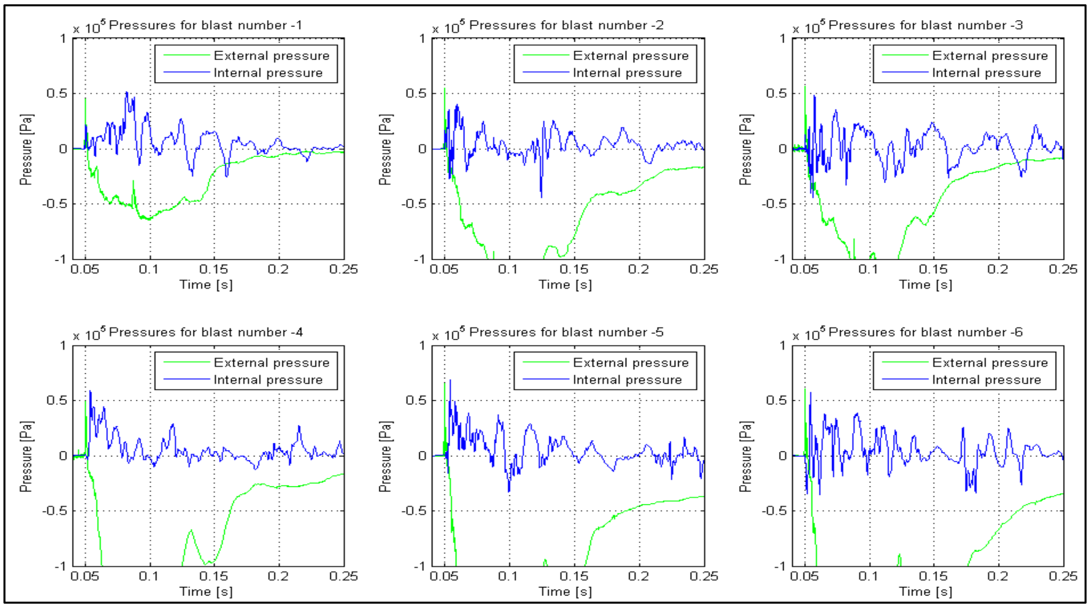


Figure 6. Ambient and brain pressure during each of the six tests.

TABLE 2. INTERNAL AND EXTERNAL (AMBIENT) PRESSURE MAXIMA AND MINIMA

Blast number	**Max positive brain pressure (kPa)**	**Max negative brain pressure (kPa)**	**Max positive ext. pressure (kPa)**	**Max negative ext. pressure (kPa)**
1	52.0	-25.6	44.8	-65.1
2	40.6	-44.4	54.5	-123.9
3	48.5	-44.5	56.1	-113.0
4	58.6	-12.6	48.8	-180.1
5	68.8	-33.1	65.4	-278.6
6	57.1	-35.6	60.7	-300.5

There appears to be no correlation between the internal pressure and the length and magnitude of the negative pressure phase. The maximum positive pressure does appear to increase from blasts 1, 2, 3 to blasts 4, 5, 6 by approximately 20%, but the maximum external pressure was also higher. Interestingly, the maximum negative pressure decreases from blasts 1, 2, 3 to blasts 4, 5, 6 even though the external negative pressure is far greater. A frequency analysis was also conducted but did not reveal any significant resonant frequencies. This result does not support the comments made by Merkle et al. [5], who asserted that the inclusion of the face shield attenuates ICP in the order of 75%.

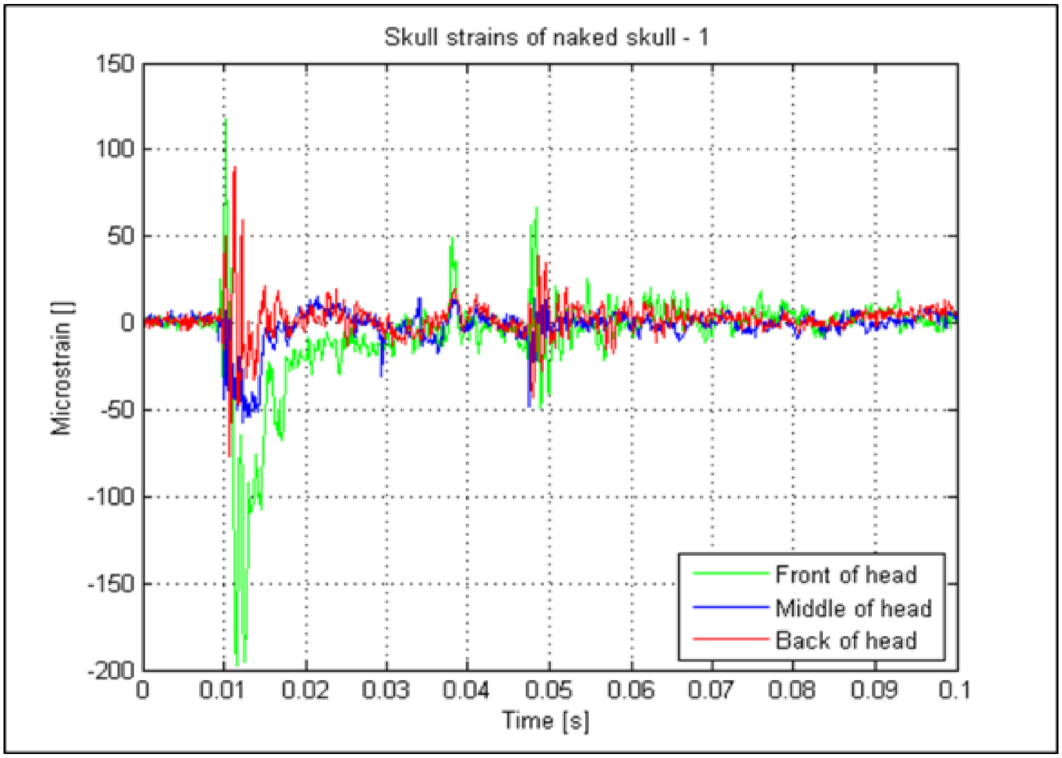
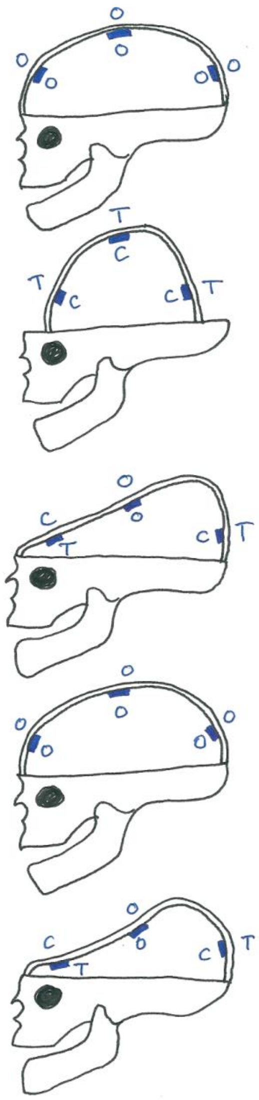


Figure 7. Strain from three strain gauges on a bare skull during blast loading

Strain Gauge Data

The strain gauge results for all six blasts were captured using three channels for each blast: front of head, middle of head and back of head. The graphs of the results of the first blast are shown in Figure 7. The strain gauges located on the inside of the skull all responded to the blast simultaneously. There was a response at the front, middle and back of the skull at the same time, however the size and direction of the response was not the same.

There are many complex physical interactions taking place as the blast wave passes over the head simulant and it is difficult to deconvolution the separate processes within this data set. But examination of the data suggests the following: When the blast hits the head, the blast pushes on the outside of the skull causing tension on the outside and compression on the inside. Because the impact is so rapid, the back side of the skull initially responds to the pressure rise at the same time as the front surface and the momentum of the skull has not allowed the skull to move yet. A combination of the higher loading on the front face and translation of the skull causes oscillations in the skull which are transmitted to the brain. Once the shock wave passes, the head is then surrounded by a field of negative pressure that is not changing as rapidly. The brain is now undergoing pressure oscillations which are observed in the pressure in the brain and as strains in the skull. This is shown diagrammatically in Figure 8.



1. Start point
The skull is at rest with no tension or compression.

2. The initial blast hits and the skull is squashed from the outside causing compression on the inside.

3. The blast has passed and the brain now has momentum and is compressing at the rear and in tensile at the front.

4. The brain returns to rest.

5. The brain continues to oscillate between the rear and rest.

Legend:
C is compression
0 is near zero strain
T is tension

Figure 8. Schematic representation of skull deformation.

CONCLUSIONS

The exposure of soldiers to blasts and the subsequent suffering from TBI will continue to be a ‘signature injury’ of modern conflicts particularly in asymmetric warfare. The tests have confirmed that the physics involved in a shock wave incident on a human head are complex and are still not yet completely understood. This study has also demonstrated a head simulant and exposed this head to the same blast six times with various attachments. The first significant deduction as a result of the analysis was that the brain does respond with an internal pressure of similar magnitude to the external pressure applied. However, once the external pressure has passed, the internal brain pressure continues to rise and fall for approximately 0.25s, far longer than the initial positive peak pressure. The strains within the skull also respond to pressure with the peak pressure being associated with the highest pressure gradients external to the head.

Another pertinent hypothesis was that the attachment of features such as the face, skin, eyes and nose caused a longer and larger response from the strain gauges indicating that the face and skin provided better coupling with the blast thus increasing the momentum of the head. The addition of the helmet and face shield did not alter the pressure results drastically and the internal pressure was of the same order when compared to that of a naked skull. The addition of the helmet and face shield altered the characteristics of the skull strains but the overall magnitudes remained similar.

REFERENCES

1. MacGregor, A. J. Dougherty A.L. Galarneau, M. R. 2011. Injury-specific Correlates of Combat-Related Traumatic Brain Injury in Operation Iraqi Freedom. J. Head Trauma Rehabil. 26(4):312-8
2. Rafaels, K. Bass, C. R. D. Panzer, M. B. Salzar, R. S. Woods, W. Feldman, S. H. et al. 2012. Brain injury Risk from Primary Blast. J. Trauma Acute Care Surg. 73(4):895-901
3. Mediavilla Varas, J. Philippens, M. Meijer, S. R. van den Berg, C. Sibma, P. C. van Bree, J. L. M. J, et al. 2011. Physics of IED Blast Shock Tube Simulations for mTBI research. Front. Neurol. 2: 58.
4. Singh, D. Cronin, D. S. Lockhart, P. A. Haladuick, T. N. Dionne, J. 2012. Evaluation of Head Response to Blast Using Sagittal and Transverse Finite Element Head Models. Defence R&D Canada - Valcartier.
5. Merkle, A. C. Wing, I. D. Carneal, C. M. Effect of Helmet Systems on the Two-phased Brain Response to Blast Loading. Proceedings Personal Armour Systems Symposium (PASS 2012), Nuremberg, Germany.
6. Haladuick, T. N. Cronin, D. S. Lockhart, P. A. Singh, D. Bouamoul, A. Dionne, J-P. Ouellet, S. 2012. Head Kinematics Resulting from Simulated Blast Loading Scenarios. Proceedings Personal Armour Systems Symposium (PASS 2012), Nuremberg, Germany.
7. Levine J, Dionne J. Head Injury Criterion (HIC) - based injury correlation for helmet-mounted sensors. Proceedings Personal Armour Systems Symposium (PASS 2012), Nuremberg, Germany.
8. Nyein, M. K. 2013. Computational Modeling of Primary Blast Effects on the Human. PhD thesis. Massachusetts Institute of Technology.
9. Bass, C. R. Panzer, M. B. Rafaels, K. Wood, G. Shridharani, J. Capehart, B. Brain Injuries from Blast. Ann. Biomed Eng. 2012. Jan;40(1):185-202.
10. Shridharani, J. K. Wood, G. W. Panzer, M. B. Mathews, K. A. Perrit, C. Masters, K. et al. Blast Effects Behind Ballistic Protective Helmets. Proceedings Personal Armour Systems Symposium (PASS 2012), Nuremberg, Germany.
11. Goatman RL. 2013. Development of a shock tube technique to invesigate blast effect on a combat helmet/head system. MSc thesis. Cranfield University.
12. Gotts PL. 2012. Blast and Ballistic Assessment of EOD suits. Proceedings Personal Armour Systems Symposium (PASS 2012), Nuremberg, Germany.
13. van Doormaal, J. C. A. M. Teland, J.A. Overview of Injury Models for Complex Blast Waves. Proceedings Personal Armour Systems Symposium (PASS 2012), Nuremberg, Germany.