Development of a biofidelic model of the human head under simulated blast loading

Jack B. Walters

University of New South Wales at the Australian Defence Force Academy

This report develops part of the framework for a paired numerical and experimental model consisting of synthetic alternatives to study the effects of blast loading on Traumatic Brain Injury (TBI). A method using aluminium and polyurethane foam projectiles fired from a gas gun shows promise to produce simulated blast loading for the study of such injuries. To understand the suitability of synthetic alternatives, compression testing on three bone candidates was completed to compare their relative biofidelity. Bonesim proved to be the most viable bone candidate and was therefore used along with a skin simulant and Sylgard 527 gel moulded into a clear acrylic tube to successfully view pressure wave propagation through the resulting synthetic ‘head section’. This configuration may provide the capacity for new insights into the mechanisms of TBI and a robust test system for novel protection systems.

Contents

I. Introduction 1
   A. Motivation 1
   B. Aims 2
II. Background 2
   A. Synthetic System Modelling 2
   B. Simulated Blast Wave Generation 3
III. Material Characterisation 4
   A. Synthetic Bone Compression Testing 4
   B. Sylgard 527 Index of Refraction 5
IV. Numerical Modelling 5
   A. Foam Impact Development 5
   B. Geometric Modelling 6
   C. Constitutive Material Models 6
V. Experiment 8
   A. Foam Impact Validation 9
   B. Head Section Impact 10
VI. Conclusions and Recommendations 11

Acknowledgments 11
References 11

I. Introduction

A. Motivation
   During training and deployment military personnel are exposed to blast waves of varying duration, magnitude (overpressure) and source. In 2013, blast gauges were issued to Australian troops deployed in Afghanistan and within the year a quarter had recorded one or more blast events [1]. The primary phase of blast injury occurs when the body is subjected to a high velocity pressure discontinuity followed by decay to negative or atmospheric pressure. While blast injury isn’t completely understood, exposure has been shown to cause arterial air embolism (blockage) and haemorrhage (rupture), with air emboli being linked to high rates of fatality in experiments involving animals [2]. Arterial rupture is easily identified due to bleed spots in the brain during post-mortem
analysis, however the most widely characterised brain injury mechanism is axonal shear, the incidence of which can be related to TBI severity [3].

Axons are the nerve fibers providing the connections and information passage between the cell bodies of neurons. Under typical loading conditions axons display elastic behaviour, however when subjected to high strain rates they tend to become brittle in nature. High strain rate loading such as a blast, can lead to direct shear or to damage from ‘secondary axotomy’ when the brain incurs strain due to swelling. The damage of axons can lead to individual neuron death and possibly chronic neurodegeneration, even after a single TBI [3]. These injuries primarily effect the areas of the brain that regulate emotion and memory, with direct correlation between mild Traumatic Brain Injury (mTBI) and Post Traumatic Stress Disorder (PTSD) [4].

The magnitude of TBI in the military is significant. It is estimated that between 15.2% and 22.8% of US troops returning from Iraq and Afghanistan sustained mTBI [5], with the high occurrence being linked to extensive use of Improvised Explosive Devices (IED’s) in these wars [6]. However, there is also evidence that high magnitude blasts such as those generated by IEDs are not the only possible cause of TBI. Säliö et al. exposed anesthetized pigs to the blast conditions generated by semi-automatic rifles, a howitzer and other military occupational training exercises. The overpressures transmitted to the porcine brain (4.4 – 28 kPa) caused significant incidence of small haemorrhages of similar profile to those found in soldiers suspected to have been killed due to blast exposure [7]. Currently, there isn’t a helmet system that provides full protection to the brain over the wide range of loading conditions in a military environment [8]. Complete protection against all situations may not be feasible, however convergence toward an optimised protection system against the most common threats is likely if the brains response can be understood in greater detail.

B. Aims

The aim of this project is to generate mechanical data from synthetic head constituents to develop a biofidelic numerical model of the human head under blast loading. Such a model would enable trial experiments to be carried out without the cost of materials and time in setup. To achieve this, several project goals will need to be met and collated in parallel with Finite Element (FE) model design. These involve:

i. Perform quasi-static and dynamic tests on synthetic bone candidates and determine their relative biofidelity.

Brain simulants in the form of silicone gel are well documented with Sylgard 527 being the most widely used [9,10]. Synthetic bone simulants are not proven to the same degree under high strain rate loading, with no standard material being employed in test systems. Several commercial orthopaedic bone simulants will be tested over a range of strain rates and compared with literature to characterise their relative biofidelity and fit for use in the testing models being developed.

ii. Developing an experimental system capable of producing blast wave loading conditions within the realistic range of human exposure and viewing the response of synthetic ‘head section’.

To validate and refine the FE model, an experimental apparatus with the ability to replicate blast conditions in a synthetic head cross-section will be developed. The Impact Dynamics Laboratory (IDL) at UNSW Canberra has used a proof of concept ‘head section’ consisting of skin-bone-brain synthetics in low strain rate experiments. The brain modelled as silicone gel contains a spatter pattern and is held within an optically clear casing that provides the ability to record strain fields with Digital Image Correlation (DIC) software from high speed camera footage. The shock generating apparatus is to incorporate a modified version of this model to characterise brain response to loading from the laboratory’s light gas gun. By designing a robust apparatus, the FE and head cross-section models can be developed in parallel to produce a set of solution options which could provide both qualitative and quantitative insights into brain injury characterisation. If biofidelic models are produced, an efficient, robust and ethical way of understanding the systems that could protect the brain will have been created, enabling novel or promising concepts developed in the FE space to be accurately converted into physical test models for further investigation.

II. Background

A. Synthetic System Modelling

There is a large amount of existing research in modelling human head based systems under blast loading; either experimentally, numerically or both in parallel. Some experimental models such as the one presented by Varas et al. are anatomically shaped and utilise synthetic bone material filled with a ballistic gel [11]. These ‘headforms’ are usually impacted by air from a shock tube and allow for internal dynamic pressures to be measured. There have been other studies such as the one conducted by Goeller et al. which involve visual...
inspection of blast wave interaction with a clear synthetic gel [12]. This study was successful in identifying pressure spikes caused by cavitation bubble collapse within the gel, leading to the observation that cavitation may be a possible contributor to TBI. However, there are no systems in place which utilise all layers of the human head to account for bone and skin layer interactions while still allowing visual inspection of the brain simulant under loading.

The majority of FEA based testing on headform systems utilise viscoelastic models for the brain and elastic models for the skull, skin and other parts of the headform, such as the study completed by Chafi et al. [13]. It is also common for experimental studies that are paired with FEA to use material models based off the literature of real brain tissue to verify biofidelity of the experimental results. However, if modelling the synthetic substitute isn’t considered, the two analysis methods de-couple and the ability to apply the research to application based solutions reduces. Petr et al. identified experimental variability as well as a knowledge gap on the mechanical response of brain tissue under blast loading as the largest sources of error in blast based FEA models [14]. It is for this reason that a simplified model such as that presented by this report could aid in progressing the fundamental understanding of brain tissue response to high strain rate loading and reduce these errors.

B. Simulated Blast Wave Generation

The 70mm diameter gas gun used in the IDL can accelerate projectiles to velocities between 200ms⁻¹ and 4.5kms⁻¹. The most common way of generating planar shock waves is by accelerating a flyer plate into a target and ensuring both surfaces make complete contact by employing high tolerance surface finishing and alignment [15]. A shock wave is formed within both materials and a compressive stress wave of thickness equivalent to twice the flyer plates thickness is transmitted into the target. The wave will eventually attenuate to zero magnitude within a semi-infinite target or be reflected from a rear surface. Pressures generated within the materials are usually in the order of GPa’s over a timeframe of microseconds.

![Figure 1. The idealised Friedlander waveform (a). A more realistic waveform (b).](image)

Blast waves have a different profile. The wave can still be viewed as a travelling discontinuity of pressure, however the profile trailing the peak overpressure is one that decays exponentially to negative or atmospheric pressure. An idealised free-field (Friedlander) waveform is shown in figure 1 along with a more realistic waveform found in military context. The highly reverberated profile is caused by surface reflection and internal weapon design [2]. Blast waves are typically generated in the order of MPa and over a timeframe of milliseconds, an effect that cannot be replicated through a typical plate impact experiment. For this reason, alternative testing methods utilised with the gas gun must be investigated to mimic a blast wave profile within the realistic range of human exposure to free-field blast waves. Wood et al. specify M107 artillery rounds as one of the most common explosives modified for IED use. They developed a realistic range of human exposure with peak overpressures ranging from 20kPa to 2MPa and blast duration from 0.5 – 50ms [16].

Several options have considered to produce a simulated blast wave pressure profile using the IDL’s gas gun. A divergent piston actuated water tube like that used by Espinosa et al. [17] was identified as showing promising capability in attenuating a wave to the correct peak pressure, however due to the magnitude of attenuation most of the trailing profile was lost, resulting in a positive pressure phase that was too short. The other possible configuration for generating shock profiles is by impacting the target directly by metallic or polymeric, closed or open cell foams. Wang et al. considered the behaviour of closed cell foams, concluding that collapse occurs layer by layer along the axis of shock. The cell walls were found to display an increasing collapse resistance in the form of strain and strain rate dependant hardening [18]. Radford et al. demonstrated that these dynamic properties of foams can be used to simulate shock loading [19]. The foam is seen to disintegrate on impact followed by layer-wise collapse of cells as per the previously referenced studies. Using aluminium foams with 11-17% relative densities, decaying pressure profiles were generated with peak pressures between 50 and 80MPa at respective velocities of 360 and 460ms⁻¹. It was shown that the pressure magnitude and duration can be manipulated effectively by changing the impact velocity, foam impactor length and relative density. Similar results were found by Chen et al. but instead using a polyurethane based foam [20]. The small pore size of the foam meant that fracture trigger points had to be drilled into the rigid foam to initiate the pressure release expected from a blast wave.
III. Material Characterisation

A. Synthetic Bone Compression Testing

To ethically and affordably test biological based systems, biofidelic synthetic substitutes are ideal. While developing bone substitutes independently is possible, such as shown by Plaisted et al. [21], a variety of commercially available options developed for orthopaedic testing present a class of options with morphology difficult to obtain through independent manufacture and mechanical repeatability that can’t be replicated by real samples. Five samples were sourced and tested in compression loading transverse and longitudinal to the layer interface planes at strain rates of 0.001/s and 0.1/s. The samples shown in figure 2 have been developed by Synbone™ from a polyurethane substrate with an average density of 0.78g/cm$^3$, Sawbone™ from a polyurethane foam core sandwiched by epoxy sheets with an average density of 1.02g/cm$^3$ (A) and 1.05g/cm$^3$ (B), and Bonesim™ from ground bovine bone with an average density of 1.34g/cm$^3$ (X) and 1.41g/cm$^3$ (Y).

![Figure 2: Transverse (left) and longitudinal samples: Synbone, Sawbone A, Sawbone B, Bonesim X, Bonesim Y.](image)

![Figure 3: Synthetic bone compression testing in different orientations and strain rates.](image)

Table 1 shows the average elastic properties of the bone analogues tested. McElhaney et al. presents an average elastic modulus of human cranial bone in transverse compression to be 2413 MPa with a standard deviation of 1450 MPa [22]. Patel et al. also summarise the large range in which human bone mechanical properties can be

<table>
<thead>
<tr>
<th>Property</th>
<th>Synbone</th>
<th>Sawbone A</th>
<th>Sawbone B</th>
<th>Bonesim X</th>
<th>Bonesim Y</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elastic Modulus (MPa)</td>
<td></td>
<td>Transverse (Impact orientation)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.001/s</td>
<td>1456</td>
<td>395</td>
<td>476</td>
<td>830</td>
<td>1513</td>
</tr>
<tr>
<td>0.1/s</td>
<td>1045</td>
<td>469</td>
<td>545</td>
<td>898</td>
<td>1249</td>
</tr>
<tr>
<td>$\sigma_{y(0.2%)}$ (MPa)</td>
<td></td>
<td>Longitudinal</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.001/s</td>
<td>15</td>
<td>4.4</td>
<td>4.9</td>
<td>7.9</td>
<td>19</td>
</tr>
<tr>
<td>0.1/s</td>
<td>23</td>
<td>5.4</td>
<td>6.6</td>
<td>12</td>
<td>18</td>
</tr>
<tr>
<td>Elastic Modulus (MPa)</td>
<td></td>
<td>Longitudinal</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.001/s</td>
<td>1183</td>
<td>4673</td>
<td>5098</td>
<td>2503</td>
<td>2644</td>
</tr>
<tr>
<td>0.1/s</td>
<td>1175</td>
<td>5938</td>
<td>6616</td>
<td>3034</td>
<td>2452</td>
</tr>
<tr>
<td>$\sigma_{y(0.2%)}$ (MPa)</td>
<td></td>
<td>Longitudinal</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.001/s</td>
<td>23</td>
<td>63</td>
<td>61</td>
<td>28</td>
<td>28</td>
</tr>
<tr>
<td>0.1/s</td>
<td>32</td>
<td>82</td>
<td>79</td>
<td>29</td>
<td>34</td>
</tr>
</tbody>
</table>

Table 1: Elastic properties of selected synthetic bone surrogates.
classified, with the elastic modulus of cancellous bone ranging from 1.1-9800 MPa [23]. All samples fall within this range and could reasonably represent human cranial bone elastically. All samples also show strain rate sensitivity, almost always increasing in stiffness and yield strength with strain rate. The largest difference in loading direction comes from the samples which have greater density gradients. When loaded longitudinally the high-density layers are providing feedback instantly rather than being delayed by low density crushing as shown in transverse loading. The Synbone samples are the most repeatable and show near isotropic behaviour, while the larger pore size and density variation in the Bonesim samples present an anisotropic behaviour with a plastic deformation pattern far closer to that of real bone [24]. While the Synbone has a similar elastic region to the Bonesim, it’s region portrays continuous densification. However, real bone will develop a possible range of fracture patterns resulting in stress plateau or release [25]. For strain values up to 30% the Sawbone is a reasonable representation of low density trabecular bone, however the density variation between layers is too great to accurately represent a cranial bone composite system at low strain rates.

B. Sylgard 527 Index of Refraction

The refractive index of a transparent medium is particularly important when analysing its response through visual methods such as shadowgraph or digital image correlation. The index of refraction of Sylgard 527 has been measured to be 1.40189 at the Fraunhofer C line (\(\lambda = 656.281\) nm) [26]. It is important to verify the refractive index of the Sylgard 527 used in this report and determine the effect of any moulding methods on its optical properties. An experimental setup (as shown in figure 4) utilising a laser diode, diffraction grating (100, 300 and 600 lines/mm) and camera enables the refractive index of a medium to be measured by analysing the diffraction pattern maxima’s. The refractive index can be found through a simple modification to Snell’s law of refraction by including a diffraction order shift to account for the grating slits. Aligning the incident beam perpendicular to the grating and using the trigonometric relationship between the refraction angle, depth of thin layer and depth of medium, the refractive index can be determined through equation 1.

\[
\eta_{\text{mat}} = l \cdot \frac{\lambda}{A} \cdot \frac{1}{\sin\left(\tan^{-1}\left(\frac{d}{A}\right)\right)}
\]

As an initial test and control a sheet of wax paper was mounted adjacent to the diffraction grating, providing light diffusion to capture the diffraction pattern through air. The refractive index of air is documented to be 1.0003 [27]. Measurement of this pattern resulted in a laser diode wavelength of 643.8nm ± 8.3nm using the documented refractive index and averaging the result from the 1st and 2nd order maxima’s. This is within the diodes rated wavelength of 630-660nm.

To form a block of Sylgard 527 gel as per figure 4, the two-part mixture was combined at 1000rpm for 10 minutes, placed under vacuum until all bubbles were evacuated, and let to cure for a week. The diffraction grating was placed against the front face of the Sylgard block and wax paper applied to the back face. Measuring the resulting diffraction patterns provided an average index of refraction of 1.40 ± 0.01. This result proves that the mixing and curing methods used do not cause the gels optical properties to deviate from those expected from the manufacturer.

![Figure 4: Refraction experimental configuration.](image)

IV. Numerical Modelling

A. Foam Impact Development

As described in the background section aluminium and polyurethane based foam impacts have been proven to produce reasonable blast-like profiles and is the best method considering cost, time of manufacture and robustness. As the gas gun has a lower limit shot velocity of around 200 m/s under standard operating conditions a basic shock model was used to estimate the resultant peak pressures in the head section to ensure the foam impact generated pressures in the realistic range of exposure. Using shock hugoniots data the model derived from pressure equilibrium at the impact faces allows estimation of shock wave speed and pressure within two successive impacted materials.
To get initial estimates of generated shock pressures shock data was collated for different foams. Cheng et al. developed a shock hugoniot for aluminium foam with a relative density of 39.6%, recording an EOS equal to $u_s=0.516+1.27u_p^{2.8}$ [28]. Using this data along with shock properties for homogeneous aluminium, a power function for $C_0 (\text{km/s})$ was fitted to extrapolate the values it would take on for different relative densities ($\gamma$). A linear function was also fitted to the EOS constant ‘S’, with both functions presented as equation 2 and 3 respectively.

$$C_0 = 5.33\gamma^{2.5186}$$

$$S = 0.1159\gamma + 1.2241$$

Using these equations to generate data for 11 and 17% aluminium foam properties, the peak pressure results found by Radford et al. were replicated with reasonable accuracy. Direct impact pressures on Synbone were estimated by the shock model using the shock hugoniot presented by Appleby-Thomas et al. as; $p=800\text{kgm}^{-3}$ and $u_s=0.97+0.33u_p^{2.9}$ [29]. By using foams in the density range 150 - 400 $\text{kgm}^{-3}$ at an impact velocity of 200m/s the pressures generated by direct impact were identified as being too high and a steel witness plate was selected to reduce the pressure in the bone to within the realistic range of exposure.

B. Geometric Modelling

To further develop the experimental setup a numerical model was developed using FEA software ANSYS Autodyn. Parts were modelled to be used with the Lagrange solver in a 2D axially symmetric model space to reduce model complexity and solver time. The standard mesh of the experimental configuration is shown in figure 5 along with all relevant gauge points. Using a set of the material models identified in section C, a mesh convergence was completed and shown in figure 6. While there is no clear converging trend no major deviation between results can be seen, allowing an efficient mesh size of 0.5mm to be utilised in all future simulations. Several tube lengths were trialled, altering the length of the observable ‘brain’ section. It was determined that a length of 40mm provides enough of an observable range without compromising tube stiffness and wasting material on an area with reduced wave planarity.

C. Constitutive Material Models

As highlighted in the background section, some numerical models utilise constitutive models and parameters to represent real biological materials. However, as this numerical model is to represent the synthetic experimental model it is important for the constitutive models and parameters to represent the synthetic analogues. Utilising both sets of models can provide another avenue of verification to ensure the synthetic materials chosen will perform as a real system would under the same loading conditions. Table 2 contains a set of constitutive models for the biological materials, collated from literature and from the independent compressive tests documented in this report.

The foam shock EOS based from equations 2 and 3 was used along with an instantaneous geometric erosion strain of 2 to develop the simulated blast loading on the steel witness plate. The model also uses steel, aluminium, polycarbonate and acrylic; these materials are pre-loaded into Autodyn and no modifications were made to their parameters in the simulations run for this project.
Constitutive Model | Parameters | Reference
---|---|---
**Brain Elastic** | \( \rho \) (g/cm³) | 1.04 |
| \( K \) (kPa) | 9.3E+4 |
| \( G \) (kPa) | 1.861E+2 |
**Brain Viscoelastic** | \( \rho \) (g/cm³) | 1.04 |
| \( K \) (kPa) | 2.19E+6 |
| \( G \) (kPa) | 7.8 |
| \( G_0 \) (kPa) | 4.1E+1 |
| \( \beta \) (1/ms) | 7E+5 |
**Sylgard 527 Viscoelastic** | \( \rho \) (g/cm³) | 9.7E-1 |
| \( K \) (kPa) | 1.096E+6 |
| \( G \) Static (kPa) | 2.16E-1 |
| \( G_0 \) (kPa) | 2.53E+1 |
| \( \beta \) (1/ms) | 4.5E-1 |
**Sylgard 527 Shock EOS** | \( \rho \) (g/cm³) | 1.01 |
| \( C_1 \) (m/s) | 1.03E+3 |
| \( S_1 \) | 2.45 |
**Sylgard 527 Mooney Rivlin** | \( \rho \) (g/cm³) | 1.04 |
| \( C_{10} \) (kPa) | 8.0018E-1 |
| \( C_{01} \) (kPa) | 1.391E-2 |
| \( d \) (1/kPa) | 3E-6 |
**Synbone Shock EOS** | \( \rho \) (g/cm³) | 8E-1 |
| \( C_1 \) (m/s) | 9.7E+2 |
| \( S_1 \) | 3.3E-1 |
**Bonesim Elastic** | \( \rho \) (g/cm³) | 1.34 |
| \( K \) (kPa) | 1.008E+6 |
| \( G \) (kPa) | 4.651E+5 |
**Skin Elastic** | \( \rho \) (g/cm³) | 1.04 |
| \( K \) (kPa) | 3.47E+4 |
| \( G \) (kPa) | 5.88E+3 |

Table 2: Biofidelic Constitutive Models

![Brain Constitutive Models](image1)

![Bone Constitutive Models](image2)

**Figure 7: Brain Constitutive Models - Gauge 4.**

**Figure 8: Bone Material Models - Gauge 3.**

Figure 7 and 8 show the comparison between models presented in table 2 as pressure profiles at gauge points 4 and 3 respectively. All simulations which make up these figures were using a 13% relative density aluminium foam impactor at 200m/s. All the brain models are in good agreement except for the purely elastic model. The difference in the elastic model is due to its poisons ratio being slightly lower and therefore more compressible. The fact that the model is purely elastic means there is no parameter to account for the high strain rate behaviour as generated in the simulation. This can also be seen in the Bonesim elastic model when compared to the Synbone shock EOS. A reduction in peak stress is expected in this case due to the Bonesim having a higher density relative to the skin layer when compared to the Synbone, causing more of the pressure wave to be reflected and less transmitted through the bone.
V. Experiment

By testing the properties of synthetic material candidates independently and utilising the numerical tools available, a proof of concept experimental setup to visualise the propagation of a simulated blast wave on a synthetic ‘head section’ is shown in figure 9. The design is dimensionally equivalent to the numerical model of figure 5, however the steel witness plate is physically fixed to a frame within the gas gun chamber, allowing the complete ‘head section’ system to be compressed into the witness plate by tightening the back plate into the frame. Providing light compression ensures all components are in contact over their entire surface areas to allow for clean wave propagation. To hold the foam projectile a 650kg/m\(^3\) polymeric foam sabot is fixed to the projectile using epoxy, providing enough mass to lower the impact velocity to desirable magnitudes lower than 300m/s and clearance from the gun barrel.

![Figure 9: Experimental configuration.](image)

Figure 9: Experimental configuration.

Figure 10 shows the setup configured in the gas gun chamber, with a laser system in between the barrel and sample to allow triggering of electronics. The centre images are of the projectiles used, epoxied into their respective sabot’s. A 5% relative density aluminium foam was chosen as a low-density projectile (135kg/m\(^3\)) and Foam-iT!™ 15 (240kg/m\(^3\)) chosen as a higher density polyurethane alternative. The polyurethane foam is a two-part liquid mixture and was moulded in PVC piping before being machined to the dimensions specified. As recommended by Chen et al, six 6mm holes of 10mm depth were drilled radially 25mm from the front face of the polyurethane foam to aid collapse and simulated shock wave generation. The final image in figure 10 shows a complete ‘head section’ assembly, consisting of a skin simulant developed by Synbone (PR1043.20), a 58mm x 7mm disk of Bonesim (as recommended from the compression test results) and a 5mm thick acrylic tube filled with Sylgard 527. The gel was moulded in accordance with the methods used in the refraction experiment.

![Figure 10: From left: Gas gun chamber, Aluminium foam projectile in sabot, Polyurethane projectile during epoxy curing, complete 'head section'.](image)
A. Foam Impact Validation

Before impacting the ‘head section’ assembly, characterisation tests were carried out on both selected foams. These tests consisted of impacting the steel witness plate alone, with a Photonic Doppler Velocimetry (PDV) probe carried on a nylon mounting plate in contact with the rear surface. By using the laser trigger system mounted in the gun chamber, frequency-time signals were captured by the PDV probe to construct free surface velocity diagrams. Each impact was recorded by a Phantom™ v710 to provide visual aid in the foam crushing mechanisms.

![Aluminium foam impact through 10mm steel plate](image1)

**Figure 11:** Steel free surface particle velocity under 5% relative density aluminium foam impact at 250m/s.

Figure 11 shows the free surface velocity diagram of the steel plate after impact by the aluminium foam at 250m/s. Specific images relating to important parts of the diagram are displayed to the right with the respective time signatures noted. The first frame is at the moment of impact where no signal has been placed into the steel plate. For the first 100 μs the foam collapses in a manner that displays characteristics of a blast wave profile, before increasing in magnitude to the peak detailed in image 2. The cell size of the low-density aluminium foam is large enough to make the foam behave in a non-homogeneous manner and this harmonic behaviour could be due to fracture of thin cell walls leading to compaction and then further fracture of the next cells. However, using the constructed Autodyn model a similar velocity profile was generated at the rear of the steel plate and shown in figure 12. While the magnitude of the profile indicates that the aluminium foam model used is not of high accuracy, it does display similarly spaced pressure peaks. Because the aluminium foam model is homogeneous and utilises instantaneous cell erosion, it is unlikely that the peaks are due to large cell collapse, but rather due to the harmonic relationship between the aluminium foam impactor and the steel plate. Frame 3 shows the beginning of the sabot’s impact on the steel plate, where the pressure significantly spikes due the increased density of the sabot, before a drop due to fracture and then a final rise as the rest of the sabot’s momentum is imparted on the plate.

The second test conducted utilised the polyurethane foam, impacting a fresh steel plate at 250m/s and generating the free surface velocity diagram and images shown in figure 13. The first frame is again at the moment of impact. The polyurethane foam peaks to a higher initial velocity (and therefore pressure) than the aluminium foam, this is due to its higher density. The signal drops to frame 2, where collapse is occurring at the radially drilled holes. The reduction in velocity means that the holes aided in the collapse of the foam. However, as the velocities used in this test were much higher than that by Chen et al, the holes didn’t cause fracture to propagate through the entire length of the foam, instead only provided short term accelerated collapse. As shown at point 3, the effect of the sabot isn’t as overpowering as in the aluminium foam case because the polyurethane provided greater deceleration and lower density disparity.

![Gauge 1 free surface velocity under simulated equal impact conditions](image2)
From the results shown, the polyurethane foam shows the greatest potential in generating a blast like wave and is a far easier platform to modify. Alterations by targeted dimensional features such as holes or slots could help tailor the collapse of a projectile. The aluminium foam has demonstrated the ability to produce a profile similar to that generated in the simulations and for this reason was the foam used in the final proof of concept test.

B. Head Section Impact

The final test conducted was an impact of the ‘head section’ assembly by another aluminium foam projectile of the density previously used. After the witness plate was aligned to the barrel, the assembly was compressed lightly to ensure complete contact of all layers. The projectile was fired at 250m/s and as shown in the left side of figure 14, a set of shadowgraph frames captured the propagation of pressure waves in the Sylgard gel. The high-pressure impact of the sabot caused the steel plate to bend, cracking the acrylic tubing and creating cavitation in the gel. This proves that cavitation can occur in the brain simulant if the pressures are great enough. Next to the shadowgraphs are the results of an Autodyn simulation of the test. The pressure profile shows distinct ‘pulses’ as can be seen in the shadowgraphs. These results show that pressure waves from foam projectiles can be recorded propagating through a set of biological material simulants and that the numerical model can predict the profile of those waves.
VI. Conclusions and Recommendations

This report provides the framework for a paired numerical and experimental model consisting of synthetic alternatives, proving the capacity for new insights into the mechanisms of TBI and a robust test system for novel protection systems. All developments in the field of blast injury research are important to further understand how to effectively protect individuals from these loading conditions. While the military environment has a far higher occurrence of blast events, key findings are directly transferred to civilian practices where blast may be a threat. Valuable insights could help the development of new protection equipment, training practices and injury treatment methods in the hope to preserve life and improve quality of living post injury. Synthetic materials provide the opportunity for large scale ethical testing of injuries to the human head.

A set of synthetic bone alternatives were tested in compression, with Bonesim displaying mechanical properties in line with real bone. Along with Sylgard 527, a proven brain simulant, and a skin replicant, a ‘head section’ assembly was created to allow the display of pressure waves inside the ‘brain’. The foam projectiles used to generate a simulated blast wave to load the assembly show promise of generating a blast like pressure profile after modifications are made for use in the gas gun. To produce a better pressure profile, testing the effects of different foam geometry feature configurations along with stacked varying densities could be considered.

To allow the experiment can be conducted multiple times using the same components, a new way of holding the foam projectiles should be utilised or a sabot stripper employed to stop the large momentum transfer from the relatively high-density sabot. Also, to improve the clarity of the shadowgraph images of the ‘brain’ layer, a square tube profile could be used to remove the lens effect of the circular profile. Along with these recommendations, further data collection methods such as imbedded pressure sensors, PDV integration and digital image correlation can be used to broaden the amount of information available to provide valuable insights.

To further understand the materials used in the experiment and broaden the numerical model’s catalogue, quasi-static and dynamic tests could be conducted on varying brain and skin simulants. 3D models of the foam and other components along with more complex material models should be utilised to improve the accuracy of the numerical results. Implementing damage criterions for the ‘bone’ and ‘brain’ within the FE model should help quantify any damage occurring within the system.

Acknowledgments

I would like to thank A.D. Brown, P.J. Hazell, J.P. Escobedo and JL. Liow for their expertise and assistance over the course of this project, it has been greatly appreciated.

References


