

# The Analysis of Pressure Response in Head Injury

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## ABSTRACT

A new approach to generating physical and numerical models of the human head is presented and we aim to investigate whether it is possible to predict the response of the head for a particular impact scenario using these modelling techniques. Finite element (FE) and rapid prototyped (RP) models were generated from the conversion of 3D image data. Both the numerical and physical models were used to validate an approximate analytical model based on full 3D elasticity equations as developed by one of the authors. Good agreement was observed between the three modelling techniques and large transient pressure amplification at the site of impact was observed for impacts of low duration.

## INTRODUCTION

The study of head injuries is of obvious importance with approximately 1 million head injury cases reported in the UK each year. 60% of all such injuries are caused by motor vehicle accidents but even a simple fall or blow to the head can result in serious damage. A number of theories have emerged in an attempt to explain the injuries that occur inside the skull as a result of an impact, the most popular being the development of pressure gradients (Hardy *et al.*, 1994). As early as 1944, the concept of dynamic stresses produced by pressure gradients was explored by Gurdjian, (1944) who discovered compression and high pressure at the coup site and tension and low pressure at the contrecoup site in dogs. Gurdjian *et al.* studied this concept further by measuring the acceleration and intra-cranial pressure (ICP) response, also in dogs. Unterharnscheidt and Sellier, (1966) showed that a long duration impact to a fluid filled spherical container (relative to the pressure wave transit time) resulted in a linear pressure gradient from the pole to the anti-pole. Goldsmith, (1972) isolated three major and "relatively independent", injury mechanisms, one of which is the development of pressure gradients with a large positive pressure at the site of impact and negative pressure of equal magnitude

at the opposing pole. Thomas *et al.*, (1967) highlighted the existence of a pressure gradient across a gel-filled, dry cadaveric human skull. It was demonstrated that both acceleration and compression caused a change in ICP where high positive pressures were found to exist at the pole and a pressure gradient continued linearly through the skull to equally high negative pressures at the anti-pole. The existence of negative pressures at the site of impact during short duration impacts was not, however, examined. Other authors that subscribe to linear pressure gradient theory include Kenner and Goldsmith, (1972) and Khalil *et al.*, (1974) who examined the response of water filled, aluminium spherical shells.

Conversely, negative pressure theory relates primarily to the occurrence of cavitation, or fluid vaporisation, at either the pole or the anti-pole (Hardy *et al.*, 1994) and it is thought that brain injuries arise as a result of cavitation collapse. Cavitation is thought to occur either in the CSF or between the dura and the skull (Nusholtz *et al.*, 1996); (Brennen, 2003) and were explained as being due to the formation of a vacuum under the membranes of the head (Denny-Brown and Russell, 1941). Negative pressure at the contrecoup site was hypothesised as early as 1948 by Ward *et al.*, (1948) and has since been investigated by Gross, (1958); Suh *et al.*, (1972); Hickling and Wenner, (1973); Engin and Akkas, (1978) and Lubock and Goldsmith, (1980) to name but a few.

In a substantial autopsy study, (Fujiwara *et al.*, 1986) found that of all the cases who had received a blow to the head, only one presented a contre-coup lesion so although negative pressures are observed at the contrecoup site, they often may not be high enough to cause damage to the brain. This is a theory upheld by many such as Lindenburg, (1960) and Edberg *et al.*, (1963) who believe that brain damage occurs in locations of positive and not negative pressure. Nusholtz *et al.*, (1984) noted that although negative pressures were present during impact, they were often non-injurious and Stålhammar, (1975a); Stålhammar, (1975b) and Stålhammar and Olsson, (1975) found that negative

pressures appeared not to be important to the injury mechanism. Contre-coup lesions were in fact more common in fall incidents, (Hardy *et al.*, 1994) but even the single incident found by Fujiwara *et al.* (1986), demonstrates that under the right impact conditions, contre-coup injuries can occur from a blow rather than a fall. These findings suggest possible cavitation as discussed by Nusholtz *et al.*, (1995). It is the belief of many that negative pressures can cause brain injury as a result of cavitation collapse, (Hardy *et al.*, 1994), described as the formation of a vacuum underneath the membranes in the skull, (Denny-Brown and Russell, 1941). Whilst impacting fluid filled flasks, Gross, (1958) suggested that cavitation collapse, as a result of negative pressure, destroyed some of his glass vials. Lubock and Goldsmith, (1980) noticed that cavitation formation coincided with negative pressure transients in the formation of bubbles in the cranial fluid.

After a notable lack of information with regards to the role of skull deformation in the causation of cerebral contusion, Fujiwara *et al.*, (1989) studied the contributions of skull deformation and acceleration. They concluded that when an impact force is not severe enough to cause significant skull deformation or fracture, the resulting contusion is most likely to be caused by an accelerated motion of the head induced by the impact and that a negative pressure may cause a contre-coup contusion. With significant force to induce skull deformation, however, most of the force energy is transferred to the deformation or fracture of the skull thus reducing the accelerated motion of the head. In Fujiwara's study, the recorded positive pressures were immediately followed by negative pressures, in both the coup and the contre-coup sites.

It was suggested (Hardy *et al.*, 1994), that injuries may develop from short duration, high acceleration, high pressure impacts or long duration, low acceleration, low pressure impacts. Ljung (1975) found that although soft tissue can undergo large deformations quasi-statically without experiencing structural failure, this ability is reduced with dynamic loading, (Margulies and Thibault, 1989). This is highly relevant as short impact durations have previously been associated with the onset of high pressure and shear strain gradients in the brain (Young and Morfey, 1998) with potentially deleterious effects.

Young and Morfey, (1998) also showed that the onset of dynamic pressure effects for fluid-filled spherical shells can be predicted by the ratio of the period of oscillation ( $T_\Omega$ ) of the first ( $n = 2$ ) spheroidal mode of vibration of the fully free shell, and the duration of impact ( $T_p$ ). A  $T_p/T_\Omega$  ratio greater than 4, provided a hydro-static pressure response of linear gradient from a peak positive pressure ( $P_{max}$ ) at the impact pole to a peak negative pressure ( $-P_{max}$ ) at the anti-pole. For impacts of shorter duration ( $T_p/T_\Omega < 4$ ), the response was observed

to be dynamic at both the pole and the anti-pole with elevated peak pressures at up to 8 times of those predicted in the aforementioned case.

An approximate analytical model based on full 3D elasticity equations was developed by one of the authors (Young, 2003) and implicit analytical expressions proposed to predict the response to blunt impact. Young, (2003) noted that when the ratio of impact duration to period of oscillation was greater than 4 ( $T_p/T_\Omega > 4$ ), the pressure response in the fluid was essentially hydrostatic with a linear pressure gradient ranging from a maximum positive peak pressure ( $P_{quasi}$ ) observed under the site of impact (pole) to a minimum negative peak pressure at the anti-pole ( $-P_{quasi}$ ).

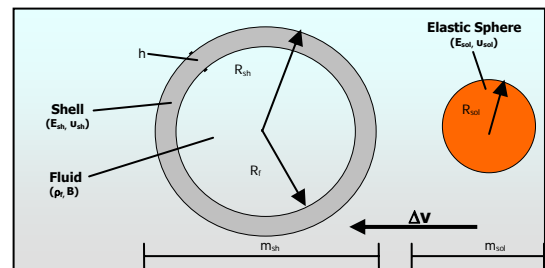
$$P_{quasi} = \frac{R_f \rho_f F_{max}}{m_{sh}} \quad \text{[Equation 1]}$$

where  $R_f$  = Radius of the fluid (inner radius of shell),  $\rho_f$  = density of fluid,  $F_{max}$  = peak force of impact and  $m_{sh}$  = mass of shell/fluid system.

The dynamic magnification of the maximum negative pressure at the pole ( $-P_{max}$ ) over the quasi-static pressure ( $-P_{quasi}$ ), in the case of a Hanning force-time history applied radially over a small spherical cap, is approximately given by:

$$\frac{P_{max}}{P_{quasi}} = -32e^{-2\left(\frac{T_p}{T_\Omega}\right)} \quad \text{[Equation 2]}$$

The analytical model is based on significant geometric and material simplifications (**Figure 1**) and so in order to test the validity of the assumptions on which the analytical model is based, experimental and numerical models were developed for comparison.



**Figure 1: Illustrative representation of the analytical shell model**

Physical head injury models have included, fluid-filled spheres (Kenner and Goldsmith, 1972) and head replicas (Edberg *et al.*, 1963); (Mason *et al.*, 1972); (Fujiwara *et al.*, 1986) for the study of stress patterns, impact pressure and acceleration. Using a fluid-filled

sphere as a means of modelling the human skull is a means of simplifying the problem and studying the head injury parameters one at a time but the use of head replica models permits the study of a more complex and therefore more realistic geometry. These modelling techniques have provided information about pressure and acceleration but the findings have often been contradictory. The purpose of the current study, therefore, is not to compare experimental results with previous findings but to utilise a set of physical models over which the experimenter has full knowledge of the model parameters. Numerical models, such as finite element, can provide means of assessing the biomechanical response of the brain using more complex geometric and material properties and hence have gained increasing popularity.

The predictions from the analytical model were validated using finite element (FE) and rapid prototyped (RP) models generated from MRI scans obtained *in vivo*. Both the numerical and physical models were generated from the conversion of 3D image data. The authors have previously reported (Johnson and Young, 2005) on the excellent agreement between the three modelling modalities for the measurement of global response characteristics such as impact duration, acceleration and peak force. The emphasis of this current paper is, therefore, on the pressure response of the intracranial fluid.

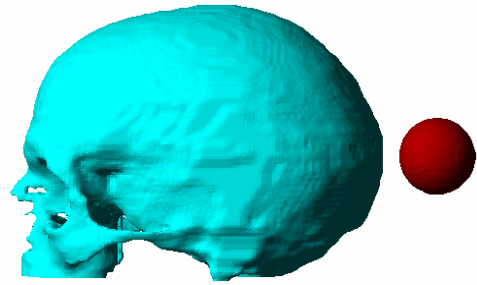
## METHODOLOGY AND RESULTS

High resolution T1-weighted whole head MRI scans of normal volunteer young males were obtained *in vivo*. Three-dimensional patient specific finite element models were generated automatically from the 3D data sets using ScanFE software from Simpleware Ltd.

The models were generated using a technique adapted from the marching cubes approach which automates the generation of meshes based on 3D scan data and allows for a number of different structures (e.g. skull, scalp, brain) to be meshed simultaneously. The resultant meshes are perfectly conforming at the interfaces (i.e. no gaps or overlaps) and the user can also elect to generate contact surfaces at these interfaces.

The resulting models are geometrically very accurate (see Figure 2) and were used to explore the intra-cranial response to impact. A rapid prototyped model of the finite element mesh was also generated in parallel using ScanFE to provide experimental corroboration for some of the finite element results obtained. Previously developed approximate closed form analytical expressions were also used to provide additional comparison results

The finite element models generated were solved using LS Dyna3D.



**Figure 2: Rendered ANSYS finite element model of fluid filled skull and impacting projectile.**

Low velocity impacts to the back of the head (occipital region) with spherical balls of 4 cm radius were simulated. The mass of the ball was varied between 0.01 Kg and 100 Kg by varying its density and the Young's modulus of the ball was kept constant  $E = 13.8$  GPa (same as skull bone). The skull was modelled filled with inviscid fluid with properties of water (Bulk modulus  $B = 2.18$  GPa and density  $1000$  kg/m<sup>3</sup>) and an impact at a relative velocity of 2 m/s was initially simulated; numerical results are given in Table 1.

Mass [kg]	$T_{imp}$ [ms]	$F_{max}$ [kN]	$P_{max}$ [Gpa]
0.01	0.16	0.38	$1.0 \times 10^{-4}$
0.05	0.31	1.08	$2.0 \times 10^{-4}$
0.2	0.54	2.42	$2.3 \times 10^{-4}$
10	1.56	9.65	$2.8 \times 10^{-4}$
100	1.72	10.78	$3.2 \times 10^{-4}$

**Table 1: Impact duration, peak force transmitted and peak pressure in the brain under the site of impact for impacts with projectiles of varying mass at 2 m/s.**

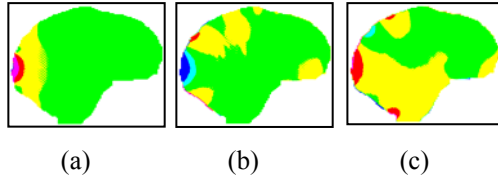
As can be seen from Table 1 as the mass of the impactor drops the impact duration also decreases which has been shown to cause potentially deleterious high pressure transients in the brain. The intra-cranial response is quasi-static for impacts with large mass projectiles with a peak positive pressure at the pole (site of impact) varying linearly to a peak negative pressure at the anti-pole as shown in Figure 3.



**Figure 3: Typical pressure response in the brain for impacts with 'heavy' projectiles.**

As the mass of the impactor decreases below a threshold value a dynamic/transient response is

observed with a resultant pressure distribution which is qualitatively different and which varies throughout the impact. At early stages after contact a high pressure transient is observed under the site of impact which is followed by a negative pressure transient and then a high positive pressure transient as shown in Figure 4 (a), (b) and (c).

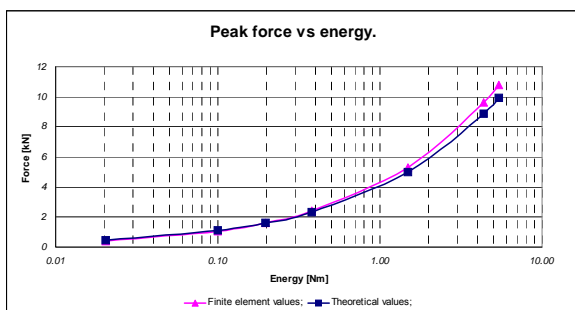


**Figure 4: Intra-cranial pressure response (red positive pressure, blue negative pressure) during impact.**

This pattern, inverted, is then observed some time later at the anti-pole (directly opposite to the site of impact).

For constant impact energy, the pressure observed for low mass projectiles (short duration impacts) is not only qualitatively different but also quantitatively much higher. In other words for the same impact energy an impact with a light projectile can cause pressure transients within the brain which are up to ten times larger than impact with a heavy projectile (e.g. impact with the ground). This may explain the existence of qualitatively different injury patterns for impacts with projectiles of differing masses.

Analytical expressions based on modelling the head as a spherical shell filled with inviscid fluid impacting a solid spherical projectile have been developed by one of the authors (Young, 2003). These were used to obtain approximate comparison values for peak force transmitted based on key geometric and material and impact parameters. Results are shown in Figure 5 for increasing impact energy as obtained using the finite element model and the analytical expressions.

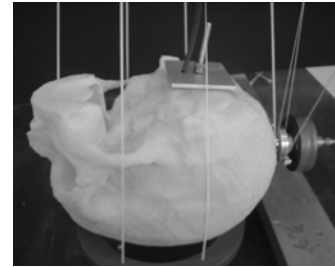


**Figure 5: peak force transmitted with increasing impact energy; analytical and finite element model results**

From Figure 5 we can clearly see that remarkably good agreement between theoretical and numerical results is achieved.

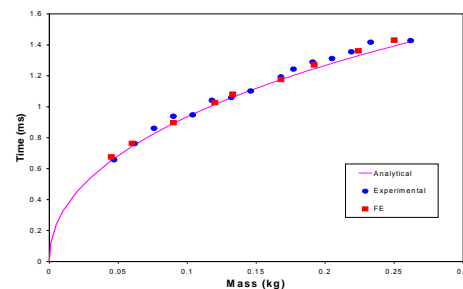
Comparison impact force and durations were computed using the finite element model and values were also obtained using analytical expressions.

In order to provide experimental corroboration the RP model was used in impact tests to characterize the response of the human head. The impact tests were carried out using the same pendulum swing system and related components as described in earlier study (Johnson and Young 2005) (Figure 6).



**Figure 6: Photographic illustration of RP skull impact rig**

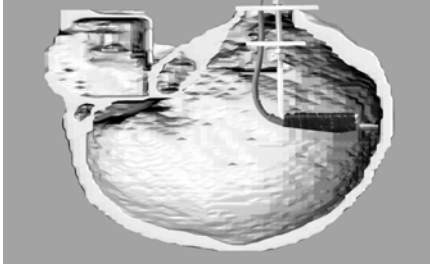
By introducing the material properties of the rapid prototype material, rather than typical values for the skull, into the finite element and analytical models results could be compared with experimental results obtained from impacting the rapid prototyped model filled with fluid. In Figure 6 the measured impact duration with different impactor masses at a constant impact velocity of 2.0ms<sup>-1</sup> are plotted. The analytical model was used to predict impact duration for a range of masses from 1g to 262g and the FE and experimental models were run using a range of masses from 45-262g. The results from the three models are shown in the graph and it is clear that the experimental results agree remarkably well with both the analytical and the FE predictions.



**Figure 7: Impact duration with change in impactor mass using analytical, experimental and FE models (constant velocity of 2.0 m/s)**

The pressure response is qualitatively different for short and long impact durations. In order to test this

experimentally a polyurethane rubber impactor, being less stiff than steel, was used in replacement of the steel ball bearing in order to produce a quasi-static response. A Kistler 211B3 General Purpose Miniature Piezotron Pressure Sensor was used to measure the transient pressure response inside the skull, immediately under the site of impact (pole) (**Figure 8**) and at the opposing site (anti-pole).



**Figure 8: Illustration of how the pressure transducer was placed inside the RP skull, during an impact to the pole**

Pressure responses obtained from the experimental test were qualitatively similar to those predicted numerically and analytically- above a threshold impact duration the response was quasi-static and below this threshold (for low impact masses)-the response led to high positive and negative transients at both pole and anti-pole.

## CONCLUSIONS

Analytical, numerical and experimental models were used in parallel to explore the pressure response of the human head as a result of low velocity impact. The patient specific, experimental and numerical models used were generated semi-automatically based on an MRI data set obtained *in vivo*. Good agreement was achieved between the results obtained from these approaches, providing a three way validation of these findings.

Elevated pressures were observed at both the pole and the anti-pole which confirms the view that impacts of low mass and short impact duration are often associated with pressures of high amplification.

Although the study was based on simplified models, the three-way validation provides confidence in results which might be obtained from models with increased bio-fidelity. Beyond its significance in the area of head impact biomechanics, the study has demonstrated that numerical models generated from 3D medical data in parallel with exact physical replica models can be used effectively to simulate physical processes. This is particularly useful when considering the risks, difficulties and ethical issues involved when using cadavers.

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