# Infant brain subjected to oscillatory loading *

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***SUMMARY:*** *Past research into brain injury biomechanics has focused on short duration impulsive events as opposed to the oscillatory loadings associated with Shaken Baby Syndrome (SBS). A series of 2D finite element models of an axial slice of the infant head were created to provide qualitative information on the behaviour of the brain during shaking. The test series explored variations in subarachnoid cerebrospinal fluid (CSF) thickness and geometry. A new method of CSF modelling based on Reynolds lubrication theory was included to provide a more realistic brain-CSF interaction. The results indicate that the volume of subarachnoid CSF, and inclusion of thickness variations due to gyri, are important to the resultant behaviour. Stress concentrations in the deep brain are reduced by fluid redistribution and gyral contact. These results provide direction for future 3D modelling of SBS.*

## 1 INTRODUCTION

In Australia, 150 out of 100,000 people per year are hospitalised for head injuries due to impact or high accelerations (Australian Institute of Health and Welfare, 2002), leading to substantial treatment and ongoing care costs, in addition to the personal and community ramifications. The search for improved diagnosis, prevention and treatment has spurred biomechanical investigation of the response of the head and its contents to various loading conditions.

Finite element (FE) modelling offers a means to safely investigate the intracranial biomechanics of a shaken infant. Current research provides some guidance to this model's creation, but there are several key differences to take into account, including brain matter material property selection, and the representation of the subarachnoid space (brain-membrane interface).

The mechanics of the subarachnoid space (brain-membrane interface) are critical to the deformation experienced by the brain matter and the developed injuries. This interface has been modelled in a variety of ways, mainly based on specific simplifying assumptions without detailed theoretical consideration. Partial validation has been achieved through the use of neutral density targets, which are implanted throughout cadaveric brains prior to testing with impact (Zhang et al, 2001), but direct observation of this interface behaviour is yet to occur. Theoretical investigation of this interface is needed to support the validity of these common assumptions, and to investigate its behaviour under different loading regimes, including shaking.

In this study, a series of 2D FE models based on an axial slice of the infant head were subjected to an oscillatory acceleration of a magnitude commensurate with infant dummy testing. A novel method for representing the cerebrospinal fluid behaviour was included. The results provide interesting qualitative information on the behaviour of the infant brain during shaking, and indicate areas of modelling importance for extension to 3D.

## 2 MODELLING TECHNIQUES

### 2.1 Mesh generation

Two-dimensional plane strain FE models of a thin axial slice of the infant human head were generated and analysed using MSC MARC FE code. The model geometry is based on MRI scans of a three-month-old infant (figure 1). This age lies within the range commonly associated with Shaken Baby Syndrome (SBS) diagnoses. One brain hemisphere was used for the analysis, assuming geometric and loading symmetry corresponding to sagital plane shaking. A typical mesh was composed of approximately 4000

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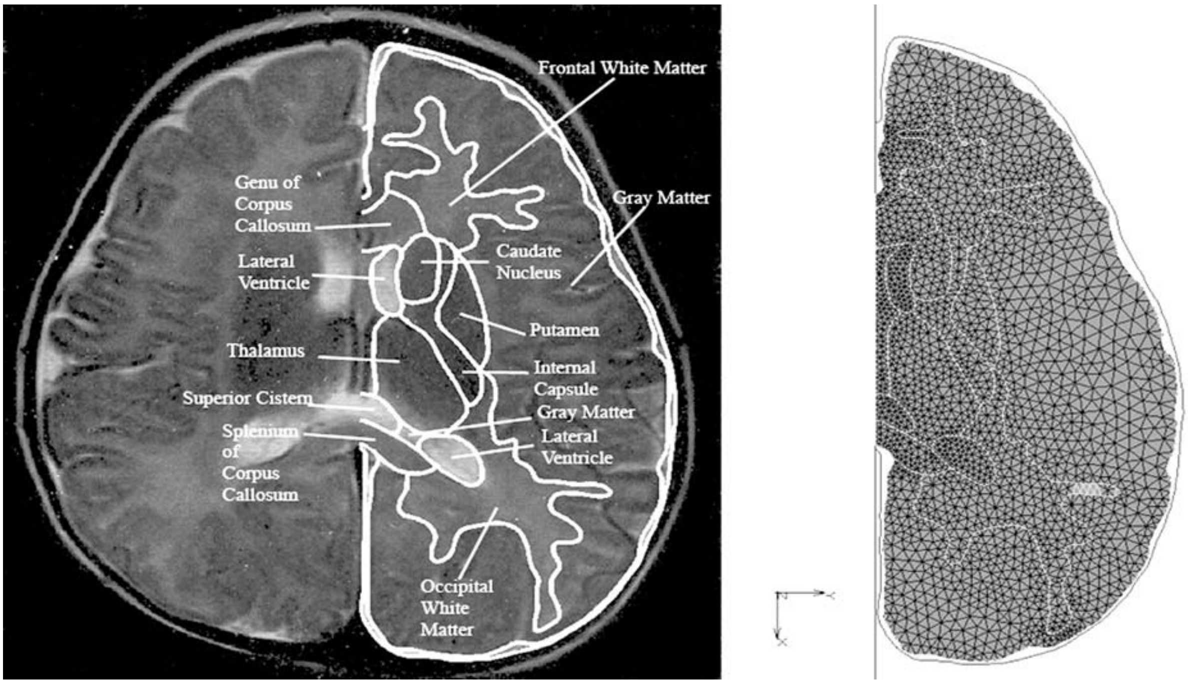


**Figure 1:** Delineation of FE mesh components and typical mesh.

elements. Care was taken to ensure element size was appropriate to the local component curvature.

Plane strain elements were deemed more representative of the lateral confinement of a 2D slice than plane stress elements, though in reality neither is strictly correct. As the study is qualitative, this was considered acceptable. Due to the relatively incompressible nature of brain matter (Zhang et al, 2001), Herrmann (1965) type triangular elements were used. These elements are formulated with a variational principal splitting the deviatoric and volumetric parts of the energy equation, and are well suited to large deformation problems (commonly used in the analysis of elastomers). For all models, the interior of the cranial bone and the dura mater and falx were represented as rigid bodies.

### 2.2 Applied accelerations

For a 2D axial model of a sagital plane shake, only anterior/posterior accelerations can be applied as a representation of sagital plane motion. This suffices for the purposes of this study, which is focused on the qualitative mechanics of the head as it undergoes oscillatory motion. We have chosen to apply a sinusoidal acceleration of 60 m/s$^2$ with a frequency of 4 Hz as a body force. This is based on physical testing results (Duhaime et al, 1987; Prange et al, 2003; Wolfson et al, 2005; Couper & Albermani, 2005), which indicate the frequency of shaking and the tangential acceleration at this head level are of this order. This is applied in the reference frame of the head; consequently, an acceleration equal and opposite to the acceleration in the global reference frame is applied to the brain matter elements and cerebrospinal fluid (CSF) representation.

### 2.3 Brain matter material properties

At this point, quasilinear viscoelastic (QLV) models are widely accepted as an appropriate form of constitutive relationship for brain matter. These models imply a strain-time separation, which has substantial support over a wide-strain and strain-rate range. With this methodology, a constitutive relationship is composed of an infinitesimal viscoelastic relationship for the shear modulus ($G(t)$) in combination with a time independent large strain relationship, producing a convolution integral (equation (1)). In alignment with the work of other researchers (Miller & Chinzei, 2002; Prange & Margulies, 2002; Velardi et al, 2006, we have elected to use a large strain relationship of the Ogden kind (equation (2)).

$$\sigma_{ij} = \int_0^t G_{ijkl}\left(t-\tau\right)\frac{\partial f\left(\tau\right)}{\partial \tau}\partial \tau \qquad (1)$$

$$W = \frac{2\mu}{\alpha^2}\left(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3\right) \qquad (2)$$

where $\sigma$ = Cauchy Stress, $f$ = large strain relationship derived from strain energy density function, $W$ = strain energy density, $\mu$ = Ogden stiffness parameter (infinitesimal shear modulus), $\alpha$ = Ogden non-linear parameter, and $\lambda_i$ = $i^{\text{th}}$ principal stretch. A non-linear parameter $\alpha$ = –4.7 has been adopted for general use in this paper (Miller & Chinzei, 2002).

For infinitesimal strain behaviour, we have adopted a multimode Maxwell model (equation (3)) for general brain matter as listed in table 1.

$$G_R\left(t\right) = G_\infty \sum_{k=1}^{n} g_k e^{-t/\tau_k} \qquad (3)$$

**Table 1:** Infinitesimal shear behaviour of brain matter used in models.

| **Stiffness** | **Dissipation** | | |
|---|---|---|---|
| **$G_0$ (Pa)** | **Mode** | $g_k$ | $\tau_k$ |
| 15000 | 1 | 0.700270 | 0.000122 |
| | 2 | 0.083459 | 0.001128 |
| | 3 | 0.076857 | 0.005872 |
| | 4 | 0.047485 | 0.045694 |
| | 5 | 0.030398 | 0.283730 |
| | 6 | 0.020524 | 1.734800 |
| | 7 | 0.016917 | 12.11300 |
| | 8 | 0.024090 | ∞ |

where $G_R$ = relaxation shear modulus, $G_\infty$ = long term shear modulus, $g_k$ = modal multiplier, and $\tau_k$ = modal relaxation constant. All materials used the bulk modulus ($K$) of water, 2.19 GPa, which is a consistent approach amongst previous analyses.

## 2.4 Cerebrospinal fluid modelling

CSF is a saline fluid that immerses the central nervous system (CNS), supplying nutrients and removing waste, in addition to providing shock absorption to injury susceptible tissues. CSF has similar mechanical properties to water, with a bulk modulus of 2.19 GPa, and Newtonian viscosity of around 0.0007 to 0.001 Pas at body temperature.

The brain/skull interface is of much importance to the overall biomechanics of the head contents, and has been modelled by many researchers, albeit by differing methods. The dural membrane is firmly adhered to the cranial bone, while the connection between the dural and arachnoid membranes is relatively tenuous. The pia matter is firmly adhered to the outer surface of the brain. The subarachnoid space is filled with CSF, while the arachnoid and pia are connected by capillaries and bridged by several veins. In terms of local mechanics during loading, sliding motion between the pia and arachnoid is resisted by shear stresses within the CSF, and the connecting capillaries, whose influence remains unquantified. Normal motion is resisted by the squeezing required to move the CSF within the head. General motion is restrained by the tethering effect of bridging veins and nerves. Locally, the resistance offered to normal motion is certain to be substantially larger than the resistance offered to sliding. In summary, researchers have desired to represent the incompressibility, high separation and low slip resistance aspects of this interface. For low Reynolds number flow regimes (Re < 2000), viscous forces are dominant, and inertial effects negligible. CSF flow in the subarachnoid space has the potential to exist in this flow regime due to the low flow thickness. If the flow is assumed unidirectional (for 2D), the rate of change of thickness slow, and pressure constant through the depth, the flow regime can be described using standard equations of lubrication theory (Williams, 2005). This theory was implemented for analysis purposes in MSC MARC through the use of several user subroutines (FORCEM, FORCDT). Body forces were included through the addition of a hydrostatic buoyancy pressure to the pressure determined through lubrication theory. The end result is a system of subarachnoid CSF resistance in which: (1) hydrostatic pressure loads simulate an effective density of the brain matter, but also result in brain matter internal distortion due to hydrostatic pressure variation over the boundary; and (2) the hydrodynamic pressure and shear loads simulate the resistance to relative motion between the cranium and brain matter.

## 2.5 Summary and test regime

A simulation series with four CSF arrangements (2 mm, 1 mm, 0.5 mm and actual thickness, represented by T1, T2, T3, T4, respectively) was prepared and run using a desktop computer. Each simulation was run for a duration of 2 s (eight shaking cycles), which was considered sufficient to approximate steady behaviour of the system. A total Lagrangian formulation with implicit Houboult time integration employing a variable time step was used, with displacement and force residual convergence tolerance of 0.001. The CSF fluid model used a density of 1000 kg/m$^3$ and a viscosity ($\mu$) of 0.00035. While testing by Bloomfield et al (1998) indicated actual CSF viscosities are around 0.0008, a reduced value was chosen to partially offset the increased flow constriction of 2D relative to 3D lubrication. A density of 1040 kg/m$^3$ was used for brain matter.

# 3 RESULTS

Considering T1, the resulting motion approaches steady state, and consists of a damped progression from posterior to anterior brain impact and vice versa (figure 2), with the motion lagging the peak accelerations by approximately 115°. The CSF resists the direction of motion of the brain matter relative to the cranium. The magnitude of

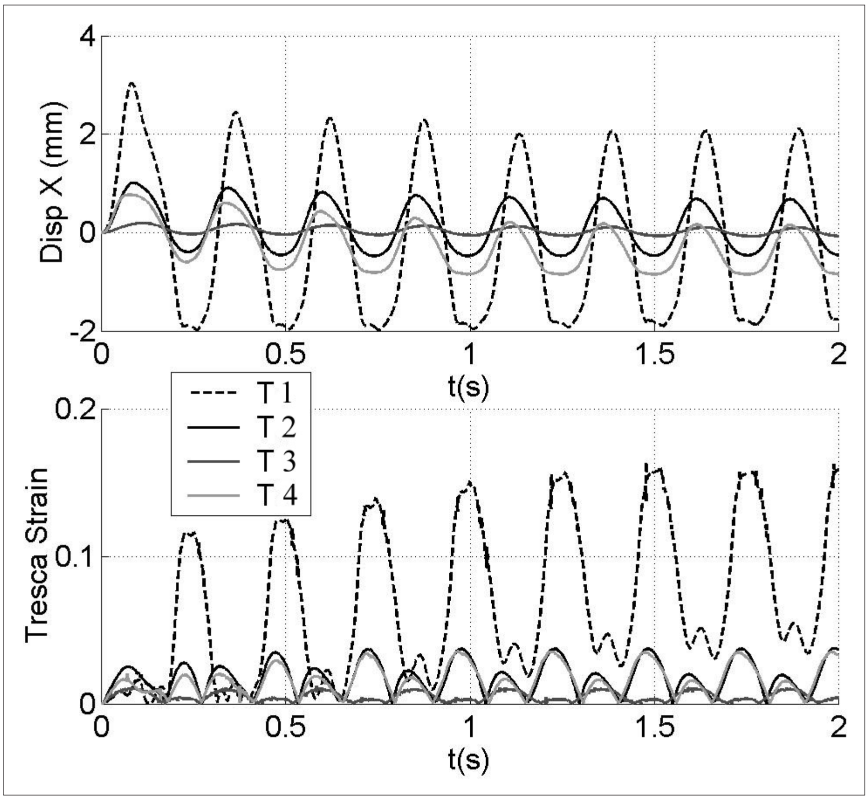


**Figure 2:** X displacement and Tresca strain for genu of corpus callosum.

this resistance is proportional to the velocity and inversely proportional to a higher power of the flow thickness. Consequently, the largest overall resisting force is obtained when a high degree of posterior or anterior squeezing is occurring (figure 3, 8b). Conversely, when the brain matter is centered, the force is lowest.

The brain-CSF interaction leads to the evolution of specific motions. The impact at either end is damped by the CSF film squeezing effect. As the impacting end is brought to a stop, the opposite end continues to compress for a short period afterward. This leads to a springing back effect, but different to what is observed during typical contact. In this case, the suction generated behind the impacted end "holds" on and slows down the motion, stretching the brain matter out. The degree of this flattening is controlled by the amount of fluid that must be shifted in the impacting area. The anterior end is "sharper", and generates large resistance only when the fluid layer is very thin, hence a more acute impact. The posterior end experiences a drawn out impact as a consequence of the larger volume of fluid to be shifted.

There is a rapid increase in the resistance to motion as the volume of subarachnoid CSF decreases, leading to predominantly smaller deformations and stresses in the brain matter (figure 2), and smoother displacement curves during peak anterior and posterior excursions. The variation in CSF thickness during peak movement is reduced; this is consistent with a more even pressure distribution between posterior and anterior ends, and thus a greater proportion of the total resistance is through relative suction pressures. For T2 and T4, the phase change moves towards 90°, as the CSF damping is able to arrest the motion of the brain soon after the acceleration changes direction (figure 2).

As for T1, strains are concentrated around the corpus callosum due to the lateral movement of the ends, with further concentration around the ventricles. Significantly, for T4, there is an additional concentration in the frontal lobe where contact occurs with the cranium. This contact occurs due to the bulge of the gyrus included in T4; locally, the approach of the brain towards the cranium is made easier, while maintaining the resistance of the system as a whole. It is interesting to note the peak Tresca strains in the corpus callosum are alleviated by this contact. Overall, strains are substantially alleviated by reductions in the CSF thickness. Over the course of the 2 s run, T4 demonstrates a movement to a steady position more removed from its initial state than the models with even CSF thickness, T1-T3. This indicates a redistribution of the initial CSF to balance the resistance of its movement in either direction.

## 4 CONCLUSIONS

It is clear from the test series T1-T4 that the CSF volume is critical to the resulting displacements and maximum strain of the brain matter, potentially the main indicators of brain injuries such as traumatic axonal injury and subdural hematoma. The relationship is quite non-linear, with a halving

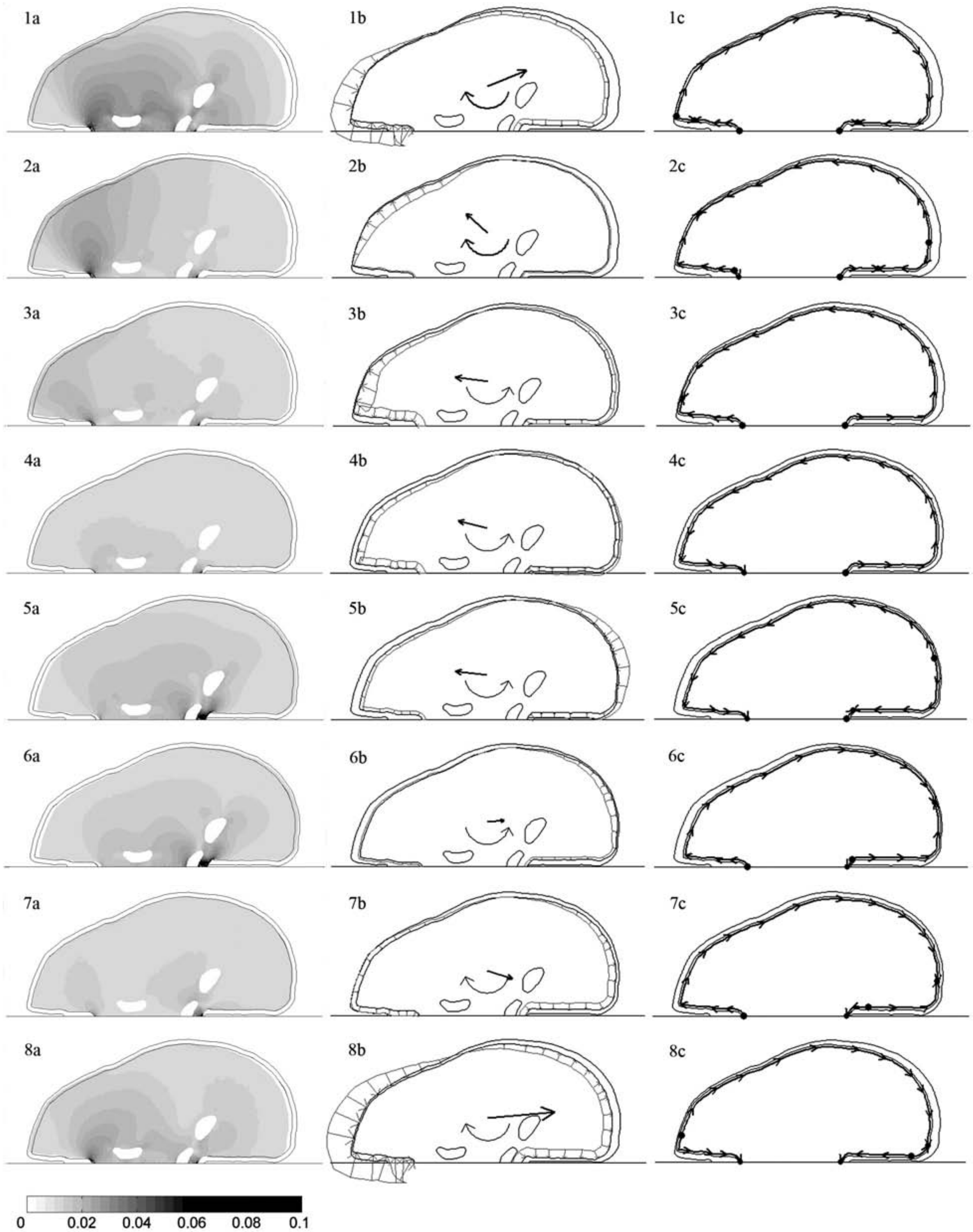


**Figure 3:** Simulation case T1 – (a) Tresca strain, (b) CSF squeezing pressure and force/moment resultants, and (c) flow direction from $t$ = 1.75 to $t$ = 1.96875 s, even spacings of 0.03125 s.

of the volume of subarachnoid CSF leading to a 80% reduction in the peak strains in the corpus callosum. In terms of SBS, the potential for a child to sustain injury will be heavily affected by the pre-existing CSF characteristics. Using the actual profile of the brain for FSI purposes reveals some important considerations. During cyclic loading, it appears as if the brain matter will move to an equilibrium position, which tends to balance the resistance of the fluid in each direction. This redistribution of the fluid may be a key part of the injury mitigation tactics evolved by the brain. Additionally, the ability of the sulci to contact the cranium and attract load while retaining the fluid resistance of the system as a whole means that the load is spread away from the concentrations (in this case, the corpus callosum) but not to the detriment of the contacted areas. Figure 2 illustrates how the system of T4, with a higher volume of subarachnoid CSF (5.1% of area) than the uniform CSF of T2 (4.8%), has lower magnitudes of stress concentration. In this way, the brain maximises its mechanical protection while still providing the volume needed to cater for the physiological processes following brain injury. This behavioural mechanism is of key importance, and has been overlooked by researchers due to the limitations of the methods employed. In conclusion, it is evident that the brain-CSF interaction is highly dependant on both the subarachnoid CSF volume, and the thickness variations associated with protusions of the gyri. These protusions alleviate deep brain stress concentrations and hence aid injury mitigation. Further teleologic protection is provided by the tendency for fluid to redistribute to balance directional resistance. Future research should take these considerations into account.

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