A two-dimensional analysis of the biomechanics of frontal and occipital head impact injuries

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Abstract: A two-dimensional plane strain finite element representation of the head was constructed and incorporated a single layered skull, cerebrospinal fluid layer, and the brain. The model that was taken in an off-centre, mid-sagittal plane in an anterior-posterior direction was used to investigate the dynamic response of the human head when subject to direct translational impact events. The physiological consequences of modelling a viscoelastic brain instead of an elastic brain were established. The influence of the stiffness of the brain matter upon the response of the head and the effect of different impact sites were also modelled. The foramen magnum was shown to have a direct influence upon the magnitude of the strains predicted to develop within the intracranial contents.

The first resonant natural frequency of the head was predicted by the model as being 126 Hz. The mode shape associated with this first resonant frequency indicated that the brain mass would rotate around a point that was located at the centre of a circle which circumscribed the boundary of the skull in the sagittal plane.

Compressive and tensile strains were predicted at the coup and contrecoup sites for both frontal and occipital impact events by both elastic and viscoelastic analyses. These correspond directly to coup and contrecoup injuries similar to those witnessed in clinical studies that involve translational deceleration.

The effects of damping and the absorption of impact energy are more accurately modelled by using viscoelastic constitutive properties for the brain material. The initial response is no different from that predicted using elastic properties, namely, compression (negative strain or positive pressure) develops at the site of coup contusion with tension (positive strain or negative pressure) developing at the contrecoup site immediately following impact. Elastic properties fail to account for the transient response by which strain levels attenuate with time and tensile and compressive waves reverberate between the contrecoup and coup sites within the brain.

NOTATION

G_0	short-term shear modulus
G_∞	long term shear modulus
β	decay factor
t	time

INTRODUCTION

The engineering aim of injury biomechanics research is to develop ways of understanding damage that can occur from impact by applying the principles of mechanics. This can only be achieved by first understanding the mechanisms of impact injury and the biomechanical response of the head-brain system to a variety of loading conditions. Brain injury mechanisms are described in terms of the mechanical and physiological changes that result in anatomical and functional damage. Due to its complex nature, functional damage is difficult to quantify and it is for this reason that a variety of alternative mechanisms have been proposed for explaining the development of brain injury. All theories, however, agree that injury is related to acceleration of the head regardless of whether the impact is applied directly or indirectly. The brain is loosely coupled to the skull with the effect that its motion will lag that of the skull, resulting in 'bruising' of the brain as it impacts against the interior surface of the skull. Stretching of the tethering blood vessels, which arises as a result of this lag in motion, can cause them to strain excessively, rupture, fail and bleed. The brain tissue itself may be damaged by normal and shear forces that develop during translational and rotational accelerations or decelerations.

Experimental studies have been carried out using animals, cadavers, dummies and other physical models. The general method is to measure forces, accelerations and/or displacements and relate these physical parameters to tolerances, severities, types and criteria of traumatic injury. The findings of such experimental studies may also be used to validate mathematical models, where the impact conditions are pre-defined and a solution attempts to predict injuries associated with hypothetical accidents. Many tests have been conducted on cadavers to provide information concerning the various processes involved in impact of the head. The primary advantage of cadaveric studies is that they have the same geometry and mass distribution as a living human, although pathophysiological changes, and therefore information pertaining to the mechanisms of damage, cannot be studied with such tests. Cadaver skulls have been studied where geometry has been of importance [1], although it is unclear how bone properties would deteriorate due to drying or embalming. Perhaps the most widely recognised cadaveric study was performed by Nahum et al. [2] who investigated two series of cadaver head impact experiments, the first consisting of single impact experiments and the other focusing on repeated impacts of different energy levels to a single specimen. The specimens were prepared in order that *in vivo* fluid pressures within the cerebrospinal fluid space and cerebral blood vessels could be measured. The seated, stationary cadaver subjects were impacted by a rigid mass travelling at a constant velocity. The blows were delivered to the frontal bone in the mid-sagittal plane in an anterior-posterior direction. Both the input force time histories and the intracranial pressure time histories were recorded.

The constitutive properties of both the skull and brain characterise the system response to mechanical loads and must be known if the physical response of the skull-brain system is to be predicted accurately. This has posed significant difficulties for researchers in recent decades following the development of adequate computational resources, since the properties of living human tissue deviate from those of cadavers and primates and no harmless non-invasive procedures for establishing such properties *in vivo* have been established. The *in vivo* dynamic constitutive properties of the human brain were investigated under conditions of pure shear [3] with the conclusion that, due to the small microstructure of its cells, brain tissue can be expected to behave as an isotropic material on a macroscopic scale. There has been definite evidence that the brain exhibits viscoelasticity and numerical modelling has incorporated estimates of an appropriate shear stress relaxation modulus [3, 4, 5]. Zhou et al. [6] used measured data [3] to deduce the shear modulus of viscoelastic brain tissue. Shear modulus parameters were determined from a logarithmic plot where G_0 is the short-term shear modulus, G_∞ is the long term shear modulus and β is the decay factor:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t} \quad [1]$$

Analytical modelling, which has largely been superseded by finite element methods within the past two decades, represents the head/brain either as a fluid filled shell or a combined mass-spring-damper system. Lumped parameter models merely define the gross rheological and/or structural response. Continuum models are more desirable for defining head injury since they can predict local phenomena such as displacements, stresses, strain rates and wave propagation. Misra and Chakravarty [7] performed an analysis that considered the eccentricity of the cranium by modelling the head as a prolate spheroidal shell (i.e., ellipsoid) filled with linear viscoelastic material representing the brain. The effect of an angular acceleration pulse was studied. The numerical investigation was based on a finite difference approach. It was calculated that an angular acceleration of magnitude greater than 20 $x10^3$ rad/s^2 can be responsible for human brain injury. The model however, considered the skull brain system to be floating whereas in reality the motion of the head is influenced by the neck and its muscles and ligaments, which limits the applicability of this model.

The diverse analytical approaches have largely used closed form solutions based on idealised geometry and boundary conditions. The skull has often been modelled as a sphere, but in reality it is closer to an ellipsoid that is narrower towards the front of the head. It has been suggested by many [8, 9, 10] and confirmed analytically and computationally that this eccentricity should be taken into account. The frontal bone provides greater resistance to stress due to its curvature. If the skull is represented by a discrete distribution of spring masses, the stiffness of the skull would be different for each impact scenario because the structural response is dependent on all of the springs involved. In addition, the radius of curvature of the skull is different throughout: in some areas it is much more rounded while in others it is much flatter.

The cerebrospinal fluid acts as an important energy damping mechanism during impact of the skull brain system. Higher contrecoup injuries have been witnessed when high accelerations before impact are involved, such as develop in a fall where the head is rapidly decelerated upon contacting the stationary ground. This has been attributed to the theory that, during the fall, the cerebrospinal fluid moves towards the impact site leaving the contrecoup site relatively deficient. This has been confirmed experimentally [12]. Cerebrospinal fluid has been seen to reduce the magnitude of the shear strains near the falx, the partition between the cerebral hemispheres, when the head is subject to rotational impact about the horizontal plane. Also, the theory of cavitation largely depends on the presence of cerebrospinal fluid near the brain. All this implies that careful modelling of the cerebrospinal fluid is essential when analysing the response of the brain to an impact. The subarachnoid space, in which the cerebrospinal fluid is contained, is non existent over the surfaces of the gyri, is relatively small where the arachnoid bridges over small sulci and is much larger in certain locations where it bridges over large surface irregularities. Such regions, containing a considerable volume of cerebrospinal fluid are called subarachnoid cisterns. To date, most of the sophisticated computational models have treated the cerebrospinal fluid as a layer of uniform depth [11], whereas in reality, it is a dispersed collection of discrete pockets that are distributed abundantly along the longitudinal fissure.

Boundary conditions, such as the foramen magnum (the opening at the base of the skull) must also be considered carefully in any representative model. The presence of the foramen magnum allows the intracranial pressure to change. Representing the foramen magnum by a force free opening in combination with a no-slip interface [13] is likely to impose too strong a constraint on the movement of the brain through the opening for the spinal cord. Sauren & Claessens [14] suggested that using a slip condition at the skull-brain interface, whilst also including a small opening around the spinal cord to represent the foramen, might provide a more realistic solution. The kinematic neck boundary condition must also be modelled to allow the head respond realistically to impact. The most reasonable boundary condition probably depends on the impact condition (i.e., the distance from the head neck junction, regardless of whether the accelerations are translational or rotational) and these would allow some simplifying assumptions.

In this paper a two-dimensional plane strain finite element model is introduced and validated against clinical findings and used to simulate frontal and occipital impact scenarios.

DESCRIPTION AND VALIDATION OF FINITE ELEMENT MODEL

Model Description

A two-dimensional plane strain model was constructed in the mid-sagittal plane of the head in an anterior-posterior direction; the outline of this model is shown in Figure 1. The geometry of the model incorporates some of the rises and falls or gyri and sulci of the brain. A single layered skull encompassing the brain was modelled. The two-dimensional slice was generated slightly off centre in order to avoid the falx cerebri and corpus callosum. The length of the model measures 192 mm from anterior to posterior and, as such, represents an average male human head.

The mesh for the finite element model was generated in MSC/Patran [15] using four noded isoparametric quadrilateral elements, all of which were of unit thickness. The finite element model, which is shown in Figure 1, contained 12081 elements and 25766 degrees of freedom. Boundary conditions were chosen to simulate restraint against large rigid body motions at the base of the cranium by constraining the chosen nodes in all three degrees of freedom. The material properties of the model were assumed to be homogenous, isotropic and linear elastic and are listed in Table 1. These values were obtained from a combination of experimental results and parametric analyses which were used to fit results to the experimental cadaveric data [2].

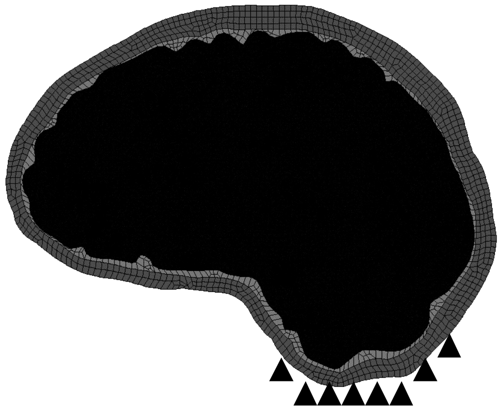

Figure 1. Two-dimensional finite element mesh of the head model including skull (shown in grey), cerebrospinal fluid (light grey), and brain (dark grey) containing 12081 elements. Nodes at the base of the skull (marked out by solid triangles) identify the constrained region of the model.

Table 1. Isotropic constitutive properties used in elastic analysis of head impact (taken from Ruan [11]).

Material	Young's modulus **[kPa]**	Density **[kg/m^3]**	Poisson's ratio
Brain	558	1040	0.485
CSF	148.5	1040	0.499
Skull	6.65 x 10^6	1410	0.22

Model Validation

Since any finite element model involves a degree of simplification, it was first necessary that the present human head model be validated against experimental data in order to ascertain its usefulness and its potential for accurate predictions. Most of this validation comes from cadaver studies, despite the fact that without in vivo data it is impossible to ascertain how the cadaver's response is affected once the blood supply has ceased.

The frequency response of a head model can be used to indicate the validity of that model. The vibration response of any elastic structure is related to the excitation and depends on the natural frequencies of the structure involved. In a critical review of finite element modelling of head impact, Khalil and Viano [16] concluded that any correct finite element model must predict resonant frequencies and mode shapes that are in close agreement with those determined clinically. Furthermore, since it has been possible to measure the natural frequency of the head by in vivo test methods, this proves to be a viable method of validation.

Furthermore, knowledge of the natural frequencies can also play an important role in protecting the brain. The design of head protection devices should ensure that impact pulses of frequencies close to the natural frequencies of the brain are fully absorbed and not transmitted to the skull brain system.

A natural frequency analysis of the plane strain model of the head system was performed using ABAQUS [17]. The analysis prescribed no specific fixed boundary conditions in order that the natural frequency of the independent system could be calculated. A second modal analysis to determine the natural frequencies of the intracranial contents was also performed in the absence of the skull and cerebrospinal fluid. The first natural frequency and corresponding mode shape was recorded for each of these two analyses.

For this frequency analysis the brain material was treated as elastic. As has been discussed by Willinger et al. [18], this is appropriate for modal analysis calculations as the brain's damping characteristics have a low effect on resonant frequencies and on mode shapes.

The resonant frequencies corresponding to the first five natural modes of vibration of the entire head and intracranial contents and of the brain solely were calculated using the model. The results are shown in Figure 2 below: the first natural frequency of the head was calculated at 126 Hz which is in close agreement with in vivo studies that have measured this to be 120 Hz [18-21].

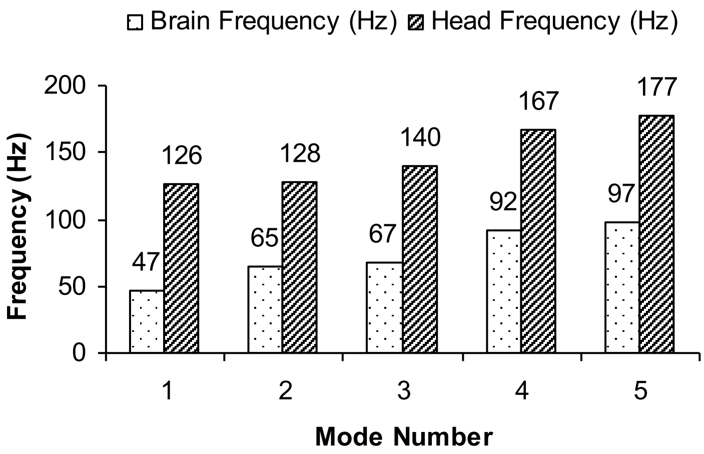

Figure 2. First five natural frequencies of the head and brain models calculated using ABAQUS.

RESULTS AND DISCUSSION

Having validated the present two-dimensional finite element model by means of the preceding modal analysis investigations, a series of progressively more sophisticated analyses was carried out to simulate the biomechanical response of the head to frontal and occipital impact conditions. These simulated experiments involved a viscoelastic analysis of a frontal impact, which corresponded to the cadaveric experiments of Nahum et al. [2], and a viscoelastic analysis of an occipital impact. These cadaveric experiments had examined the consequences of a blow to the head of a seated human cadaver. The impact was directed to the frontal bone in the mid-sagittal plane in an anterior-posterior direction. The intracranial pressure time histories had been recorded by Nahum's use of accelerometers in the frontal bone adjacent to the impact area and in the occipital bone. These pressure locations had been termed coup and contrecoup respectively.

Within the computational simulations, the impact was represented by an approximate half sine wave pulse of magnitude 8000 N and duration 4 ms. Transient dynamic analyses were performed for 10 ms and these are discussed below.

Viscoelastic Analysis of Frontal Impact

The effects displayed by a time-dependent constitutive model of the brain were simulated by defining viscoelastic material properties for the brain material which were the same as those used by Zhou et al. [6] with G_0 = 41 kPa, G_∞ = 7.6 kPa and β = 700 s^{-1}. All other material properties were as defined in Table 1.

Figure 3 details the distribution of longitudinal strains (ε_x, anterior-posterior direction) that develop after 3 ms within the head. Compressive strains exist in both the cerebrospinal fluid and brain tissue at the impact or coup region, while tensile strains develop in the opposite or contrecoup region. Physically, this corresponds to the brain being compressed against the front of the skull, compacting subarachnoid matter, and moving away from the back of the skull, thus inducing tension in this region.

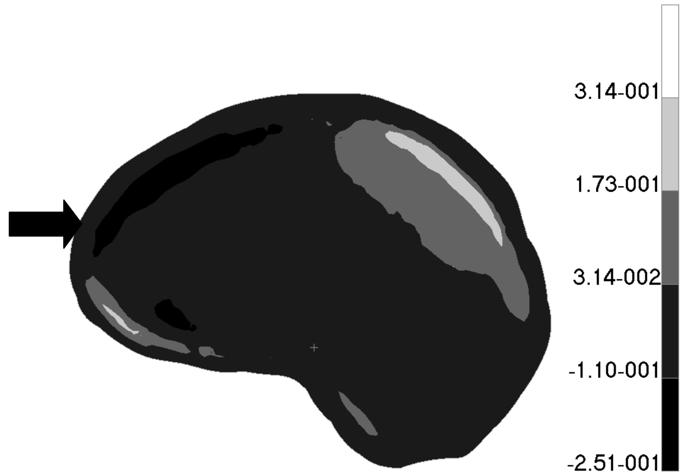

Figure 3. Response of the head to frontal impact after 3 ms. Contours indicate absolute values of longitudinal strain. Negative compressive strains are seen in the cerebrospinal fluid and brain tissue at the impact (coup) site, positive tensile strains are seen in the cerebrospinal fluid and brain tissue at the opposite (contrecoup) site.

This same two-dimensional model was used in the analysis presented in [22, 23], where elastic properties (Table 1) were used for brain tissue instead of the present viscoelastic properties. This elastic analysis [22, 23] also predicted the development of compressive and tensile strain at the coup and contrecoup locations. However, more severe levels of strain were found to occur within the brain tissue when the present viscoelastic properties were used. The distribution and magnitude of the predicted strains up to a time of 2 ms were similar to those predicted using the elastic brain properties as shown in Figure 4. However, for the remainder of the viscoelastic analysis the tensile and compressive strains did not extend as widely as possible throughout the brain. Compressive strains were always concentrated at the coup site while tensile strains were consistently present at the contrecoup site. This agrees more closely with clinical data than the results of the elastic analyses and indicates that viscoelastic properties of the brain are important if transient and post-impact strains and trauma are to be predicted using such computational models. Figure 4 below compares the variation of maximum strains within brain tissue as predicted assuming elastic or viscoelastic constitutive properties.

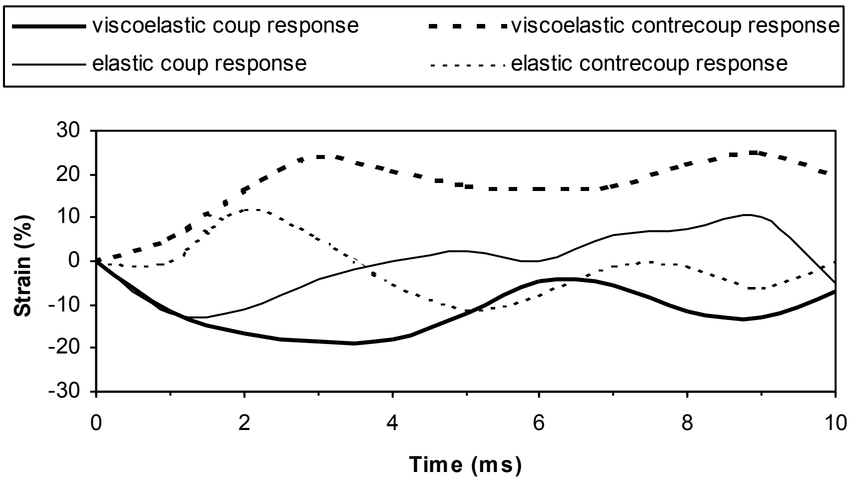

Figure 4. Maximum longitudinal strains within the coup and contrecoup regions of brain tissue due to a 4 ms frontal impact of magnitude 8000 N. Both viscoelastic and elastic brain properties were assumed in two separate analyses.

Viscoelastic Analysis of Occipital Impact

To investigate the response of the head to an occipital impact, an identical impact pulse to that used in the frontal analysis was imposed on the model at the occipital side of the skull in a posterior-anterior direction. Boundary conditions and material properties were identical to those used in the previous frontal analysis and a similar 10 ms transient analysis was performed.

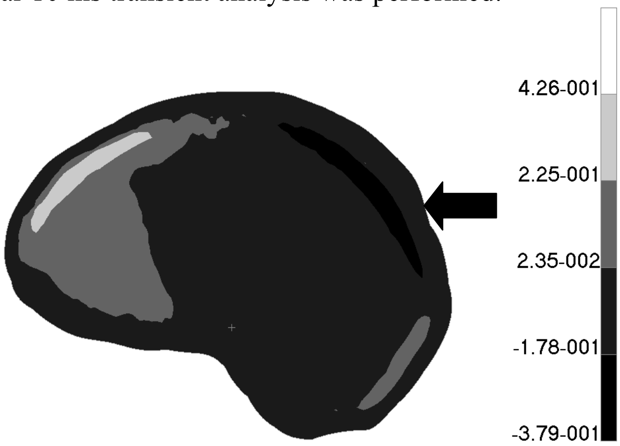

Figure 5. Response of the head to occipital impact after 3 ms. Contours indicate absolute values of longitudinal strain. Negative compressive strains are seen in the cerebrospinal fluid and brain tissue at the impact (coup) region, positive tensile strains are seen in the cerebrospinal fluid and brain tissue at the opposite (contrecoup) region.

A comparison between the viscoelastic occipital and frontal analyses of Figures 3 and 5 indicates that the quantitative difference is in the magnitudes of strain that develop after the impact. For the occipital impact conditions, higher magnitudes of coup and contrecoup strains were observed to develop within the brain than for the frontal impact analysis. This difference is believed to be due to the greater constraint and consequent amplification provided by the relative narrowness of the frontal region of the skull. Figure 6 below presents the results of an identical occipital impact scenario but using elastic brain tissue constitutive properties. As was shown similarly for the frontal impact simulation, the use of elastic brain properties fails to account for the transient response by which strain levels attenuate with time and the manner in which tensile and compressive waves reverberate between the coup and contrecoup sites within the brain tissue.

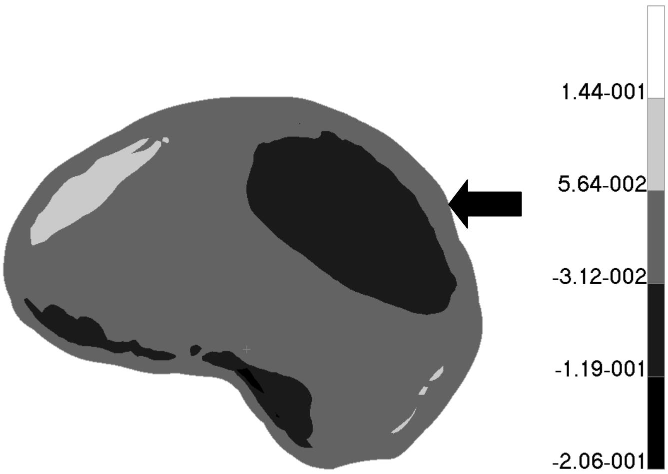

Figure 6. Response of the head to an occipital impact after 3 ms with elastic properties. Contours indicate absolute values of longitudinal strain. Negative compressive strains are seen in the cerebrospinal fluid and brain tissue at the impact (coup) region, positive tensile strains are seen in the cerebrospinal fluid and brain tissue at the opposite (contrecoup) region.

Influence of Foramen Magnum on Intracranial Response

The presence of the foramen magnum allows any increase of intracranial pressure due to an impact to be relieved. Therefore, in order to model this, the skull elements from this region of the model were removed and the previous frontal impact analysis was repeated.

This particular viscoelastic frontal impact simulation predicted the development of similar strain distributions within both the brain tissue and cerebrospinal fluid as had been predicted previously by the model that did not include the foramen magnum. However, the magnitudes of these predicted strains were significantly reduced from the levels that were seen when the foramen magnum was not included in the model. The response of this model at t=3 ms is shown in Figure 7, while Figure 8 indicates that, by not modelling the presence of the foramen magnum, the levels of strain, and consequently pressures that develop within the brain tissue are significantly over predicted.

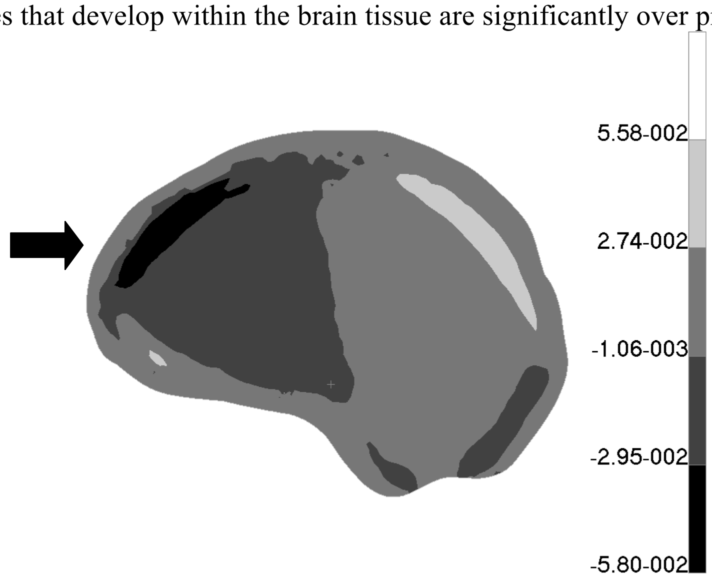

Figure 7. Response of the head to frontal impact after 3 ms. This model represents the foramen magnum explicitly, unlike that shown in Figure 3. Contours indicate absolute values of longitudinal strain. Negative compressive strains are seen in the cerebrospinal fluid and brain tissue at the impact (coup) region, positive tensile strains are seen in the cerebrospinal fluid and brain tissue at the opposite (contrecoup) region.

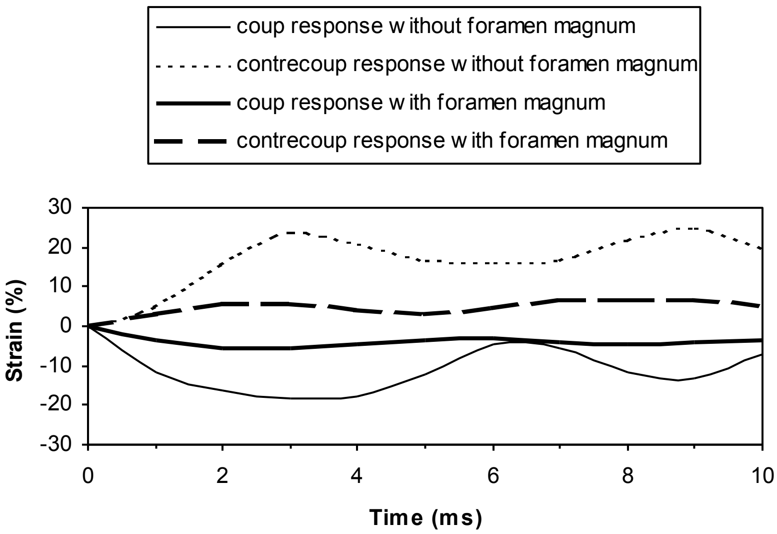

Figure 8. Maximum strain levels predicted for the coup and contrecoup regions within the elastic and viscoelastic brain under the influence of a 4 ms frontal impact, with inclusion of the foramen magnum.

Figure 8 and a comparison between Figures 3 and 7 show that the presence of the foramen magnum serves to reduce the magnitude of the strains that develop within the brain. Consequently, it can be concluded that it is essential that correct consideration be given to the skull geometry if a model is to predict correct strain levels due to an impact.

CONCLUSIONS

A two-dimensional model of the skull-brain system was created. This incorporated a single layered skull, cerebrospinal fluid and brain tissue. The model was validated by a resonant frequency analysis which gave a first natural frequency of the head as 126 Hz and the first resonant frequency of the brain as 47 Hz.

Frontal and occipital impact accidents were simulated using viscoelastic brain tissue properties. The strain histories of the viscoelastic brain at the coup and contrecoup sites were compared against those of an analysis in which the brain was assumed to have elastic properties. It was seen that viscoelastic properties for the intracranial contents are a more realistic representation than elastic properties, although this difference is only apparent after approximately 2 ms. However, both elastic and viscoelastic models successfully predicted the development of compressive and tensile strains in both simulated impact accidents that agreed with the clinical occurrence of coup and contrecoup contusion.

The model was also modified to include the foramen magnum. In a frontal impact analysis, this model predicted strain levels that were significantly lower than in those models that did not account for the foramen magnum. This modified model is believed to simulate the alleviation of any pressure build up through the opening at the base of the skull in a way that is more realistic than a model that does not represent the foramen magnum.

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