On the potential importance of non-linear viscoelastic material modelling for numerical prediction of brain tissue response: test and application

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ABSTRACT – In current Finite Element (FE) head models, brain tissue is commonly assumed to display linear viscoelastic material behaviour. However, brain tissue behaves like a non-linear viscoelastic solid for shear strains above 1%. The main objective of this study was to study the effect of non-linear material behaviour on the predicted brain response. We used a non-linear viscoelastic constitutive model, developed on the basis of experimental shear data presented elsewhere. First we tested the numerical implementation of the constitutive model by simulating the response of a silicone gel (Sylgard 572 A&B) filled cylindrical cup, subjected to a transient rotational acceleration. The experimental results could be reproduced within 9%. Subsequently, the effect of non-linear material modelling on computed brain response was investigated in an existing three-dimensional head model subjected to an eccentric rotation. At the applied external load strains in the brain were approximately ten times larger than was expected on the basis of published data. This is probably caused by the values of the shear moduli applied in the model. These are at least a factor of ten lower than the ones used in head models in literature but comparable to material data in recent literature. Non-linear material behaviour was found to influence the levels of predicted strains (+20%) and stresses (-11%) but not their temporal and spatial distribution. The pressure response was independent of non-linear material behaviour. In fact it could be predicted by the equilibrium of momentum, and thus it is independent of the choice of the brain constitutive model.

KEYWORDS – non-linear constitutive model, Finite Element head model, physical head model, brain tissue material.

INTRODUCTION

To obtain insight into brain injury mechanisms, the internal mechanical response of the head under impact conditions has to be known. Since it is impossible to determine this response using in-vivo models, Finite Element (FE) modelling of the head is often applied (Bandak & Eppinger, 1994; Claessens et al., 1997; DiMasi et al., 1991; Kang et al., 1997; Ruan et al., 1991; Trosseille et al., 1992; Turquier et al., 1996; Zhang et al., 2001; Zhou et al., 1996). Intracranial pressure, predicted by these models, has been found to correlate favorably with experimental data (Nahum et al., 1977). The computed brain tissue shear response has not been confronted with reality directly, due to the lack of experimental data. However, the finding that computed spatial patterns of stresses and strains coincide with patterns of brain injury, found in experiments, has been considered an indirect test of the predicted brain response (Miller et al., 1998, 1999).

From this comparison, injury thresholds ranging from 0.06 to 0.32 nominal principal strain were found, depending on skull-brain boundary condition assumed (Miller et al., 1998). This strain range agrees with injury thresholds found in ex-situ in-vivo isolated axon studies (0.15 to 0.21 engineering strain) (Bain & Meaney, 2000; Galbraith et al., 1993; Thibault et al., 1990). In in-vitro cell models even higher injury thresholds are found (0.63 in (Smith et al., 1999)). For this reason, strains exceeding 0.15 can be expected to have been present in the brain, during impacts resulting in injury.

It has been shown that brain tissue behaves like a non-linear viscoelastic solid for shear strains above 0.01 (Brands et al., 2000b; Darvish & Crandall, 2001; Peters et al., 1997; Prange et al., 2000). In brain tissue stress relaxation experiments, approximately quasi-linear viscoelastic strain softening material behaviour was observed for strains exceeding 0.01, i.e. shear stiffness decreases as function of strain and relaxation behaviour was nearly nearly independent of strain (Arbogast et al., 1995; Bilston et al., 2001;
Brands et al., 2000b; Prange et al., 2000). Fully non-linear behaviour for frequencies above 44 Hz was reported by Darvish & Crandall (2001). For this reason, it can be concluded that brain injury will occur in a strain region where the material behaves non-linear.

Although current FE head models vary in degree of complexity with which the geometry and skull-brain interaction is modelled, they all assume linear shear material behaviour for brain tissue. It is unknown to what extent neglecting of non-linear behaviour of brain tissue in current FE head models affects the results of these models. Given the shear softening observed, it might be expected that the strains they predict at a given external head load underestimate the real tissue strains. Consequently, the injury threshold, derived from comparison of computed strain patterns and measured injury pat-tems, might be to conservative. Applying non-linear material behaviour also might lead to different spatial strain patterns than in the linear case, as wave propagation velocity will be influenced by the time and strain dependent stiffness.

Assessment of the influence of material non-linearity in a FE head model requires a non-linear constitutive model. In Brands (2002), a non-linear viscoelastic constitutive model is developed on the basis of experimental data presented in Brands et al. (2000b).

The constitutive model was implemented in an explicit FE code (MADYMO 5.4 (TNO-Automotive, 1999)). Special attention was paid to implementation of the volumetric response: since the bulk modulus of brain tissue is about 10 times higher than the shear modulus (Etch et al., 1994; Goldman & Hueter, 1956; Lin & Grimm, 1998; Lin et al., 1997), numerical artifacts are easily introduced in the solution (Brands, 2002; Nusholtz & Shi, 1998). In (Brands, 2002; Brands et al., 2002) the constitutive model and its numerical implementation was tested against brain tissue stress relaxation data obtained from rotational rheometer experiments by Brands et al. (2000b). It was found that non-linear viscoelastic behaviour of brain tissue could be predicted realistically for shear strains up to 0.2 and strain rates up to 8 s−1. However, the numerical implementation of the constitutive model was not tested for more impact like deformations.

The main objective of this study is to investigate the effect of non-linear material behaviour on the predicted brain response by applying the constitutive model in an existing three-dimensional head model. However, before doing this, the numerical implementation of the non-linear material model is tested for more impact like conditions. This is the second objective of this study.

Testing the numerical implementation of the constitutive model using a realistic FE head model is problematic. Firstly, the strain levels, predicted in the brain tissue, are not only subject to errors in the implementation of the constitutive model, but also to other model errors, e.g. in the descriptions of the geometry or the skull-brain interface, or a poor mesh quality. Secondly, no accurate experimental data is available to serve as benchmark data.

To avoid these problems, we develop a physical model to test the quality of the numerical implementation. Such models have been used before to get an indication of strain levels inside the brain under impact conditions (Margulies et al., 1990; Meaney et al., 1995; Viano et al., 1997), and to perform parametric studies (Bradshaw et al., 2001; Ivarsson et al., 2000). In our model design we emphasise a straightforward translation to a numerical model rather than prediction of the real head impact re-sponse. Errors that arise from translating the physical model to a numerical model are eliminated as much as possible by using a simple, well defined geometry and known boundary conditions. As in the physical model studies, mentioned above, we use Dow Corning Sylgard 527 A&B silicone gel as brain tissue model material. The experiments are simulated with a FE model that includes the constitutive model developed in Brands (2002) and results are compared.

Subsequently, we apply the constitutive model in an existing FE head model presented in (Claessens, 1997; Claessens et al., 1997). The model is loaded with a transient eccentric rotation in the sagittal plane. The effect of nonlinear material behaviour is assessed by performing the simulations both with and without inclusion of non-linear material parameters in the same constitutive model. The brain response is investigated in stress and strain quantities commonly associated with injury.

METHODS

Description of constitutive model

A brief description of the non-linear viscoelastic constitutive model is provided here. A more comprehensive description is beyond the scope of this paper and is presented elsewhere (Brands et al., 2000a; Brands, 2002; Brands et al., 2002).

For accurate modelling of nearly incompressible materials the Cauchy stress, \( \sigma \), is additively composed of a volumetric part \( \sigma^v \) and multiple deviatoric contributions \( \sigma^d \), which depend on change of shape only (i.e. not on change of volume).
\[ \sigma = \sigma^r + \sum_{i=0}^{n} \sigma^d_i \] 

(1)

**Deviatoric part** The deviatoric part of the stress is modelled non-linearly viscoelastic. It is made up of a number of contributions, \( \sigma^d_i \), each with its own time constant. In a one-dimensional analogue, these modes can be regarded as a number of spring-dashpot elements in parallel. Non-linear behaviour of the model is achieved by replacing the linear springs by non-linear ones while linear dampers are used. In the three-dimensional model, the deformation gradient tensor, \( F \), is split multiplicatively into an elastic part (c.f. elongation of spring), \( F_{e,i} \), and an inelastic part (c.f. elongation of dashpot), \( F_{p,i} \) for each mode, \( i \) (e.g. Leonov (1976)):

\[ d\vec{x} = F \cdot d\vec{x}_0, F = F_{e,i} \cdot F_{p,i} \]

(2)

where \( d\vec{x}_0 \) and \( d\vec{x} \) represent a material line element in undeformed and deformed state respectively.

The elastic behaviour, for each mode, is modelled by a third order non-linear Mooney-Rivlin strain energy density function,

\[ W = C_{10,i} (\bar{I}_{1,i} - 3) + C_{01,i} (\bar{I}_{2,i} - 3) + C_{20,i} (\bar{I}_{1,i} - 3)^2 + C_{02,i} (\bar{I}_{2,i} - 3)^2 + C_{30,i} (\bar{I}_{1,i} - 3)^3 + C_{03,i} (\bar{I}_{2,i} - 3)^3 \]

(3)

in which \( \bar{I}_{1,i} \) and \( \bar{I}_{2,i} \) are the first and second invariants acting on the isochoric elastic Finger tensor in mode \( i \). For clarity we omit the index \( i \) in subsequent equations. Using the definition of a deviatoric tensor, \( \mathbf{A}^d = \mathbf{A} - \frac{1}{3} \text{trace} (\mathbf{A}) \mathbf{I} \), this yields the deviatoric part of the stress \( \sigma^d \) in each mode,

\[ \sigma^d = \frac{2}{\sqrt{3}} \left[ C_{10} (\bar{I}_1 - 3) + 2C_{20} (\bar{I}_1 - 3)^2 + 3C_{30} (\bar{I}_1 - 3)^3 \right] \mathbf{B}^d - \frac{2}{\sqrt{3}} \left[ C_{01} (\bar{I}_1 - 3)^2 + 2C_{02} (\bar{I}_2 - 3)^2 + 3C_{03} (\bar{I}_2 - 3)^3 \right] \mathbf{B} \]

(4)

To illustrate the non-linear behaviour of each mode in this model, we elaborate equation (4) for simple shear,

\[ \sigma = 2(C_{10} + C_{01}) \gamma + 4(C_{20} + C_{02}) \gamma^2 + 6(C_{30} + C_{03}) \gamma^3 \]

(5)

with \( \gamma \), the elastic part of the shear strain.

The inelastic, time dependent, behaviour of each mode is modelled like a simple Newtonian fluid (or linear damper). It provides the inelastic rate of deformation, \( D_{p,i} \), from the deviatoric part of the stress obtained using equation (3), and a viscosity, \( \eta_i \),

\[ D_{p,i} = \frac{\sigma^d}{2\eta_i} \]

(6)

This inelastic rate of deformation is used to update both elastic and inelastic parts of the deformation. To prevent numerical inaccuracies, which can easily be present due to the nearly incompressible material behaviour, a strain evolution equation stated in rotation invariant strain(rate) quantities (for details refer to Brands (2002)).

**Volumetric part** The volumetric part of the stress is modelled using a linear elastic relation between the hydrostatic compression \( I_3^{1/2} \) and a bulk modulus, \( K \),

\[ \sigma^v = K \left( I_3 - 1 \right) \mathbf{I} \]

(7)

**Physical model: experimental setup**

The numerical implementation of the constitutive model was tested by simulating the response of a silicone gel filled cylindrical cup, subjected to a transient rotational acceleration. The experimental setup, shown schematically in Figure 1, consists of three parts: the actual physical model, the spring driven loading device and the deformation recording section, consisting of camera and post processing unit.

**Physical model** The physical model consists of a cylindrical polymethylmetacrylate (PMMA) cup. For practical reasons, the cup dimensions are chosen smaller than real human head dimensions. The cup inner diameter was set to 70 mm and its depth equals 26 mm.
The cup is filled with Dow Corning Sylgard 527 A&B silicone gel. A and B components of the gel were mixed in a one-to-one mass ratio and cured in the cup, for at least two weeks at room temperature, before experiments commenced. The gel sticks to the PMMA walls of the cup, thus providing a well defined, no-slip, boundary condition.

Two geometry variants are used. In the first one, the cup is left open providing a stress free top surface of the gel. Simulation of the open cup experiments provides insight into the model performance at large strains. In the second geometry variant, the top surface is closed by a transparent cover. This closed cup provides insight in the model performance at isochoric deformations.

**Loading device** The cup is rotated along its axis of symmetry by a pre-tensioned spring, connected to the cup via a ratchet wheel. The maximum angle over which the cup is accelerated is set to 2.2 rad. As to obtain a gel response, representative for impact, the amount of pre-tension in the spring, was adjusted such that strain and strain rate levels associated with injury are obtained (refer to appendix A).

**Deformation recording** Histories of both gel deformation and cup rotation were determined using marker tracking. 4-mm diameter black plastic markers were placed in the gel (gel markers) and on the cup edge (cup markers) (Figure 2). For the open cup, gel markers were placed on top of the silicone gel while for the closed cup gel markers were placed at half cup height in a plane parallel to the top plane. The mass density of the markers is less than that of the gel. In-plane marker positions were recorded at 4500 frames per second by a high-speed video camera (Kodak Ektrapro ES Motion Analyser) at a resolution of 256X256, 8-bit gray-scale pixels. The optical axis of the camera was positioned as to approximately coincide with the axis of rotation of the cup (refer to Figure 1). Diffuse lighting equipment utilizing infrared filters has been used to reduce object heating and reflections.

The markers were tracked using the MATLAB Image Processing Toolbox (MathWorks Inc., 1999) and their positions were computed in polar coordinates. A
typical noise level seen in the results is on the order of 0.03 mm. Note that in this manner out-of-image-plane marker displacements are not measured.

**Numerical model of physical model experiment**

*Model assumptions* It is assumed that no slip occurs between the gel and the cup and that the cup and its cover behave rigidly. For this reason only the gel is modelled. For the closed cup simulation, radial and axial motion of all outer surfaces are suppressed while circumferential motion is prescribed according to the experimentally obtained rotation history. For the open cup simulation, the same boundary are applied except for the upper surface, which is considered stress free.

The silicone gel shows linear viscoelastic behaviour for strains up to 0.5 and frequencies up to 460 Hz in oscillatory experiments by (Brands et al., 1999, 2000b). This behaviour was modelled with the viscoelastic material model presented before, with non-linear material parameters set to zero (refer to Table 1). The bulk modulus, \( K \), is determined from ultrasonic data, as \( K = \frac{c p^2}{\rho} \) with ultrasound velocity, \( c_p = 1048 \text{ m/s} \) (Stelwagen & Boer, 2000) and mass density \( \rho = 970 \text{ kg/m}^3 \). This modulus equals about half the bulk modulus of brain tissue. Figure 3 shows that the model fits the experimental data well.

**Numerical implementation** Spatial discretisation is achieved using 28 brick elements over diameter, with linear interpolation functions and reduced integration. To keep the initial element shape close to cubic, twelve element layers are applied over the cup height. This mesh density is such that decreasing element size to 50% leads to a 4.0% increase of peak values only and does not alter the time course (refer to appendix B for more details). Time integration is achieved by the, conditionally stable, Central Difference Method. Miklowitz & Achenbach (1977) showed that the type of FE models used here provides the most accurate solution when the maximum time step which provides a stable solution, provided by the Courant criterion, is applied. To prevent instable solutions, this maximum time step is multiplied by 0.85 and kept constant during the complete simulation (0.001 ms).

**Post processing** Nodes are selected with radial distances coinciding as close as possible to real marker radial position (maximum difference 1.3%) but different angular coordinates. This latter is allowed as the model and its response are axisymmetric. Trajectories in the x-y plane were determined and are presented in polar coordinates for comparison with the experimental results.

**FE head model**

The head model used was originally developed in the implicit Finite Element Code MARC (MARC,1994) by (Claessens, 1997; Claessens et al., 1997) and has been evaluated using intra-cranial pressure responses obtained from a cadaver experiment by Nahum et al. (1977). Next, the model has been used in an accident

### Table 1

<table>
<thead>
<tr>
<th>Mode</th>
<th>Bulk Modulus ( K ) [Gpa]</th>
<th>Mooney-Rivlin parameters</th>
<th>( C_{10,j} = C_{01,j} ) [Pa]</th>
<th>( \eta_j ) [Pa s]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>1.065</td>
<td>54.10</td>
<td>( \infty )</td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>-</td>
<td>30.58</td>
<td>37.89</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>-</td>
<td>94.70</td>
<td>9.360</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>-</td>
<td>429.3</td>
<td>3.702</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>-</td>
<td>6870</td>
<td>2.689</td>
<td></td>
</tr>
</tbody>
</table>
reconstruction study by Verhoeve (1998). In that study improvements with respect to element size and shape were carried out. Furthermore, the anatomical detail of the model was increased by including a more realistic geometry near the skull base and falx-cerebri and by including an element layer between cerebral hemispheres and skull to model the dura mater.

For the present study, the model is transformed to MADYMO 5.4.1 (TNO-Automotive, 1999). In the next sections the model will be presented briefly.

Geometry
The geometry of the model is obtained from the Visible Human data set (U.S. National Library of Medicine, 1996). Head length (195 mm), width (155 mm) and height (225 mm) have been scaled to 50th percentile measures according to Pheasent (1986). The anatomical structures, included in the model, are grouped in three components: the cranium, the meningeal layers and CSF, and the brain tissue (Figure 4). All structures are assumed to be rigidly connected to each other. The motion of brain tissue through the foramen magnum is suppressed. The head is meshed in one continuous mesh consisting of 14092, 8-node, brick elements. Mesh quality has been checked for element warpage, Jacobian values, aspect ratio and skew angles using Hypermesh 4.0 software (Altair Computing, Inc.). It was found that more than 82% of the elements qualified for the default accuracy parameters Verhoeve (1998). For practical reasons the mesh density has not been varied to check numerical accuracy. Instead it is estimated in appendix B that this mesh density will provide peak strain values which might deviate but correct time courses / trends. Furthermore, Spatial integration is achieved using one integration point per element only (reduced integration). Hourglass control is used to suppress the zero energy deformation modes that exist for reduced integrated brick elements (TNO-Automotive, 1999). During the simulations, the hourglass suppression routine contributed less than 4.5% to the total internal energy of the model.

Time discretisation
As in the physical model simulations, the maximum time step by the Courant criterion is multiplied by 0.85 and kept constant during the complete simulation. This yields a time step of 0.4 μs, which results in 75000 time increments to simulate a 30 ms impact.

Constitutive assumptions
In this study we will not model a direct contact impact. For this reason the cranium is assumed to be rigid. As in the original model by Claessens et al. (1997), the meningeal layers were modelled linearly elastic with material parameters taken from Ruan et al. (1991) (Young’s modulus E=31.5 MPa, Poisson’s ratio, ν=0.45 and mass density ρ=1130 kg/m³). The brain tissue is modelled using the newly developed material as explained before. Brain tissue material parameters taken from Brands (2002) and are shown in Table 2. Figure 5 is reprinted from results of simulated stress relaxation experiments in Brands (2002) to illustrate the correctness of the material parameters. In the reference simulation, the brain tissue is modelled with linear material parameters only (LIN). The simulation including shear softening (SOF) is like LIN but with higher order terms added. The mass density is set to 1040 kg/m³.

![FIGURE 4. Overview of the head model (Verhoeve, 1998). The right half of the head model is shown with the brain, the meningeal layers (dura mater with falx and tentorium) and the skull separated. In the model, these structures are connected to each other, resulting in a continuous Finite Element mesh.](image1)

![FIGURE 5: Simulation results of stress relaxation experiment at three strain levels versus experimental results (apparent stresses sample G2 in Brands et al (2000b)). Num = MADYMO result, Exp = Experiment. Results taken from (Brands, 2002).](image2)
Loading conditions An eccentric rotation in posterior-anterior direction has been applied to the model. The axis of rotation is positioned at 155 mm below the anatomical origin of the model, located in the ear hole projected to the sagittal plane. The eccentricity resembles a typical neck length (Thunnissen et al., 1995). The rotational acceleration time history, obtained from Bandak & Eppinger (1994), consists of two sine functions shown in Figure 6. The initial acceleration part equals twice the amplitude and half the duration of the subsequent deceleration part. The total rotational pulse duration was set to 30 ms whereas the maximum angular velocity was set to 5 s\(^{-1}\) as to obtain strain values inside the brain tissue model within the strain range of experimental data to which the constitutive model parameters were fitted, i.e. 20% shear strain approximately.

Post processing The effect of non-linear material behaviour on the dynamical brain response is investigated in terms of quantities associated with injury. Maximum shear strain (Holbourn, 1945; Margulies et al., 1990; D.F.Meany et al., 1994; Ueno et al., 1995; Viano et al., 1997), maximum principal strain (Bandak & Eppinger, 1994; Bain & Meaney, 2000; Miller et al., 1998; Schreiber et al., 1997; Thibault et al., 1990), Von Mises stress (Miller et al., 1998; Schreiber et al., 1997; Ueno et al., 1995) and pressure (Bandak et al., 1999; Nusholtz et al., 1996; Viano et al., 1997; Young & Morphey, 1998; Zhang et al., 2001) have been identified by various authors as potential causes for brain injury. These can be described by the following three quantities,

- Von Mises, or effective, strain, 
  \[ E_{VM} = \sqrt{\frac{1}{2} \text{trace} \left( E^d \cdot E^d \right)} = \sqrt{-\frac{3}{2} I_2 \left( E^d \right) } \]
  , with \( E^d \) the deviatoric Green-Lagrange strain. It is a measure of distortional deformation and provides a measure of maximum principal strain differences present in the material. We did not use maximum principal strains as we wanted a consistent separation of deviatoric and hydrostatic response as in the material model description.

- Von Mises stress, \( \sigma_{VM} = \sqrt{\frac{1}{2} \text{trace} \left( \sigma^d \cdot \sigma^d \right)} \), depends both on strain and strain rate as viscoelastic material behaviour is assumed.

- Pressure, \( p = -\frac{1}{3} \text{trace} (\sigma) \), is a quantity that has been used in many studies for evaluating head models Claessens et al. (1997); Kang et al.
For each element in the brain mesh, maximum values appearing during the impact time history are determined and shown in contour plots of various para-sagittal cross-sections. Areas of interest will be determined and the time history of the quantities in these areas will be plotted.

RESULTS

Test of numerical method

Open cup Figure 7 shows a sequence of frames of the open cup during the acceleration part of the loading (right). On the left, numerical results are shown in which elements approximately at marker positions have been high-lighted. These pictures depict the qualitative agreement between numerical and experimental results.

Rotation (\(\phi\)) and angular velocity (\(\omega\)) of the cup are shown versus time in the upper left plots in Figure 8. The angular velocity of the cup increases to a value of 90 rad/s approximately. The upper right plot of Figure 8 shows trajectories of three selected markers in the gel as well as the trajectories of the markers on the cup. The rotation of the cup is axisymmetric, as expected. In the lower left plot the rotation of the gel markers, relative to a cup-fixed co-rotating coordinate system, is shown. The markers lag behind in motion with respect to the cup. Also, the relative marker rotation increases for increasing radial distance from the edge of the cup. This is expected since there is decreasing influence of edge boundary conditions towards the centre. Peak relative rotation occurs first at about 40 ms for marker 1 close to the cylinder edge and occurs a few ms later for markers 2 and 3.

The time history of the radial displacements, in the lower right plot of Figure 8, shows that the radial displacements remain approximately zero until 30 ms. After this, they increase to 2 mm approximately. The radial displacements are related to the three-dimensional deformation present during this experiment as shown in Figure 9. The lower plots in Figure 8 also show numerical results. The general shape of the curves is the same for both experiment and simulation. The numerical model however, overestimates the relative rotation of the markers by 17%, 14% and 24% for markers 1, 2 and 3, respectively. The numerically predicted radial displacements are within range of the experimental results.

FIGURE 7: Sequence of frames starting at t = 20 ms (upper) to 53 ms (lower). Experimental results shown (right column) as well as numerical results with elements high-lighted close to marker positions} for qualitative comparison (left column). Cup rotation is clock wise. Note: marker positions in numerical results are illustrative as nodes with different angular coordinates but accurate radial coordinates have been used for quantitative analysis (refer to main text).

FIGURE 9: Side view of open cup experiment showing the three-dimensional deformation field present. Time history from left to right.
FIGURE 8: Results of open cup experiment. Top left: cup rotation and angular velocity. Top right: Marker trajectories: initial positions of gel markers numbered 1 to 3, initial positions of cup markers labeled *. Bottom left: Circumferential displacements of gel markers relative to cup circumferential displacement (both experimental and numerical results shown). Bottom right: Radial displacements of gel markers (both experimental and numerical results shown).

FIGURE 10: Results of closed cup experiment. Top left: cup rotation and angular velocity. Top right: Experimentally obtained marker trajectories: initial positions of gel markers numbered 1 and 2, initial positions of cup markers labeled *. Bottom left: Relative circumferential displacements of gel markers. Bottom right: Radial displacements of gel markers. Both numerical (Num) as well as experimental (Exp) results shown.
Closed cup Figure 10 shows that the load applied to the closed cup is similar to the one applied to the open cup. The maximum relative rotations equal 0.12 and 0.24 rad for markers 1 and 2. The radial displacements of the markers are on the order of 0.1 mm.

The numerical model over-predicts the amplitude of the angular displacements by 62% and 46% for marker 1 and 2 respectively, but still provides similar time histories. The predicted radial displacements are in the same order of magnitude as the experimental results (0.1 mm) but show different time courses.

Influence of non-linear material behaviour in the FE head model

Figure 11 shows contour plots of the maximum Von Mises strain and stress values during the impact in various para-sagittal cross-sections in the brain for both the linear and the non-linear model. The Von Mises strain ranges up to about 0.3 while the Von Mises stress ranges up to approximately 400 Pa. Both stresses and strains show similar, concentric, contour

![Contour plots of maximum Von Mises strain and stress values](image)
patterns with minimum values concentrated in the center of the cerebral hemispheres. Maximum values appear at the upper part of the cerebral cortex (cross-sections 2 and 3 in Figure 11), in the mid-brain section (cross-section 1) and superior to the tentorium cerebelli (cross-sections 3 and 4). Another concentration of maximum values is found in the concave region at the base at the lower part of the cerebrum near the temporal lobes. However it is believed that these are not accurate due to the presence of degenerated elements in these regions. Therefore these maximum values are not taken into account in the next discussion.

The SOF variant provides approximately the same spatial distribution of maximum Von Mises stresses and strains than the LIN model. However, maximum strain values increase by approximately 21% whereas maximum Von Mises stresses decrease by 11%.

The Von Mises stress and Von Mises strain time histories for an element located in the mid-brain region are shown in Figure 12. Von Mises strain and Von Mises stress time histories show the same tendencies for both models. The upper plot shows that the maximum strain level increases in the SOF simulation whereas it occurs later in time. The stress history in the lower plot, shows that shear softening leads to decreasing maximum stresses which again appear later in time than in the LIN simulation.

The pressure response of the LIN model is shown in Figure 13. The upper contour plot shows the pressure distribution at $t=5$ ms, which is at maximum positive angular acceleration. Positive pressures arise in the posterior region while negative pressures are present in anterior region. The lower contour plot is taken during maximum deceleration of the head, at $t=20$ ms. It can be seen that pressures in anterior and posterior region changed sign while absolute values are lower than during acceleration. In both plots the spatial pressure gradient in A-P direction is nearly constant. This is also true for other moments in time not shown in Figure 13.

The pressure time history in two elements, located on posterior (P) and anterior (A) regions is plotted in the upper part of Figure 13 for the reference simulation. As observed in the contour plots, the pressures differ in sign. Their time behaviour follows the rotational acceleration input history of Figure 6. The largest pressure difference between P and A element occurs at 5.0 ms and equals 27.8 kPa. The effect of shear softening on the pressure response is negligible (maximum difference 0.3 Pa) and is not shown.
DISCUSSION

The main objective of this study was to study the effect of non-linear material behaviour on the predicted brain response in an existing three-dimensional head model (Claessens, 1997; Claessens et al., 1997). To do so, we used a non-linear viscoelastic constitutive model for brain tissue developed in (Brands, 2002; Brands et al., 2002). In (Brands, 2002; Brands et al., 2002) the constitutive model and its numerical implementation was tested against brain tissue material experiment data obtained from rotational rheometer experiments by Brands et al. (2000b) with shear strains up to 0.2 and strain rates up to 8 s⁻¹. However, its numerical implementation was not yet tested for more impact like deformations. For this reason we tested the numerical implementation of the model first.

Test of numerical implementation

Experiment A physical model was used to test the numerical implementation of the constitutive model in an deformation state expected during impact. To do so, the physical model was designed to minimise the number of modelling assumptions and to maximise experimental accessibility. Furthermore a known brain tissue mimicking material has been used, Dow Corning 527 A&B silicone gel.

The mechanical behaviour of the silicone gel resembles that of brain tissue in that it possesses a shear stiffness on the same order as that of brain tissue, displays viscoelastic behaviour and has a bulk modulus to shear modulus ratio of 10⁶ (Brands et al., 2000b). Especially this nearly incompressible material behaviour poses special demands on numerical implementation. The gel does not replicate exact brain tissue mechanical behaviour as the gel shows linear viscoelastic behaviour for strains up to 0.5 and frequencies up to 460 Hz whereas brain tissue behaves non-linear (Brands et al., 1999, 2000b). However, the silicone gel has some additional properties which makes it valuable for testing the numerical implementation of the model: good reproducibility, good adherence to the cup providing a well defined no-slip boundary condition and transparency.

Simulation The experiments were simulated with an FE model and results were quantitatively compared. Angular displacements were over predicted up to 25% and 62% relative difference for open and closed cup experiments respectively. However, time histories were similar in experimental and numerical results.

A potential source for the overestimated angular marker displacements is the fact that material parameter were determined from a different gel specimen than the one used in this experiment. Spread in small strain oscillatory strain data of three different silicone gel samples equals approximately 0.2 to 20% depending on the frequency used (Brands et al., 2000b). Repeating the open cup simulations with material parameters increased by 10%, revealed that maximum relative numerical errors in angular rotation decreased from 17%, 14% and 24% to 8.9%, 0.23% and 2.5% for markers 1, 2 and 3 respectively.

Increasing the material parameters in the closed cup simulations by 10% reduces relative errors from 62% and 46% to 40% and 25% for markers 1 and 2 respectively, but does not explain the remaining difference. The fact that the overestimation in this isochoric deformation state is more pronounced than in the open cup simulations suggests a second explanation. In the constitutive model, shear behaviour is assumed to be independent of the hydrostatic deformation. If, in reality, some coupling between shear and hydrostatic deformation is present, implementation of this coupling in the constitutive model may affect the simulation results. This effect will be different in the closed cup simulation than in the open cup simulation due to different, volumetric, boundary conditions applied. No material data on the presence of such partial coupling exists today. This topic remains to be investigated (e.g. by measuring the normal force during a simple shear experiment or by applying a different deformation, for example an unconfined compression).

The radial displacements in the open cup simulations were predicted quite well. However in the closed cup simulations radial displacements were of the same amplitude (0.1 mm) than the experimental results but showed different time courses. To some extent this is due to experimental errors. Analysis of the displacement of the cup markers indicated a misalignment of the cup axis of symmetry and the reconstructed axis of rotation of 0.3 mm. Thus the experimental accuracy is too low to draw conclusions on the accuracy of predicted radial displacements in the closed cup simulations.

Quality of numerical implementation In conclusion, the numerical implementation of the constitutive model is capable of providing realistic predictions of deformation histories in silicone gel, subjected to transient rotational loading. In particular the 6 orders of magnitude difference between shear stiffness and bulk stiffness, which is typical for both the gel and brain tissue, is dealt with adequately. Obviously, the capability of simulating non-linear behaviour was not addresses in this test. However, non-linear behaviour was tested before in (Brands et al., 2002; Brands,
and is considered to be not a major problem of the numerical implementation.

Influence non-linear material behaviour

After testing the numerical model implementation, we applied the constitutive model, with and without non-linear material behaviour in an existing FE head model to investigate the effect of non-linear material behaviour on the predicted brain response. To our knowledge, this is the first application of a constitutive model predicting the shear softening observed in stress relaxation experiments, in a FE-head model. Other non-linear constitutive models applied in FE (animal) head models predict linear shear behaviour (Miller et al. 1998, 1999) or were used for predicting low velocity indenter simulations only (Miller et al., 2000). In none of these studies a comparison with linear shear behaviour has been made. Effects of non-linear behaviour on deviatoric and hydrostatic response are discussed in separate sections followed by a model limitations section.

Deviatoric response We found that, due to shear softening, the maximum Von Mises strain values increased by approximately 21% whereas maximum Von Mises stress values decreased by approximately 11%. However, both temporal as well as spatial distributions did not change much. Also we observed shear strains on the order of 0.2 at a rotational velocity of 5 rad/s. In contrast, Bandak & Eppinger (1994) applied the same time history as we used but with maximum rotational velocity set to 35 rad/s (acceleration: 5000 rad/s²) and found principal strains on the order of 0.1 only (corresponding with a shear strain of 0.2 approximately in simple shear). However, Omori et al. (2000) applied a similar loading history (maximum rotational velocity 35 rad/s but duration 15 ms) in a half-circular disk model and found shear strains up to 0.79.

Probably, this discrepancy can be explained by the fact that the shear moduli typically used in head models in literature are on the order of 10 kPa or more (Bandak & Eppinger, 1994; Claessens et al., 1997; Kang et al., 1997; Miller et al., 1998; Zhang et al., 2001; Zhou et al., 1996), which is about a factor of ten higher than the shear moduli used in this study. However, the values used in this study and those by Omori et al. (2000), are in range of data published in literature by various authors (Arbogast & Margulies, 1997; Darvish & Crandall, 2001; Fallenstein et al., 1969; McElhaney et al., 1972; Peters et al., 1997). Another explanation for the high strains in the model are other model assumptions present in the existing model, such as skull-brain interface conditions or meninges material parameters.

Hydrostatic response The pressure response was not influenced by applying non-linear material behaviour. This was to be expected as non-linear behaviour is applied to deviatoric behaviour only, which is independent of hydrostatic (pressure) behaviour in the model. In fact, the pressure response in Figure 13 indicates that the posterior-anterior pressure difference at a certain moment in time, \( \Delta p(t) \), can be estimated from the balance of momentum as,

\[
\Delta p(t) = \rho \ddot{x}(t) l
\]

in which \( \ddot{x}(t) \) equals the linear acceleration, \( \rho \) the mass density and \( l \) a typical length measure.

To check this, the linear acceleration is estimated from the product of the angular acceleration and the neck length (155 mm), while the head length (195 mm) serves as typical length measure, \( l \). Inserting the maximum angular acceleration 785 s⁻², at \( t=5 \) ms, provides an estimated \( \Delta p \) of 25 kPa which agrees closely with the maximum pressure difference in Figure 13, \( \Delta p=28 \) kPa.

As pressure history is independent of the assumed material behaviour and accurate prediction of a constant pressure gradient poses few demands on the spatial and temporal discretisation used (Miklowitz & Achenbach, 1977), we conclude that pressure is not a critical measure for testing the quality of both the constitutive model and the numerical implementation during traffic related impacts.

For this reason, validation of head models for traffic related impacts should be based on deviatoric strains, and not on commonly used pressure data, e.g. reported by Nahum et al. (1977). Promising studies to supply such data are high speed x-ray marker tracking techniques applied in cadaver impacts (Al-Bsharat et al., 1999; Hardy et al., 2001) or the, non-invasive, MR tagging method applied to in-vitro animal studies (Dougherty et al., 1999).

Limitations of FE head model We applied the constitutive model in an existing FE head model. This FE-head model contains a reasonable detailed geometry description, but a very simplified description of the skull brain interface; all structures are tied together. In reality some displacement between the skull and brain is possible (Hardy et al., 2001). Application of such relative motion will influence the strain state in the model and might influence the effect of non-linear material modelling. Furthermore, we applied a rigid skull which makes the model not suitable for modelling direct, contact impacts. Also only one loading condition has been investigated, a rotation in posterior-anterior direction.
For this reason it would be interesting to investigate the effect of non-linear material behaviour in simulations using different loading directions.

CONCLUSIONS

A three-dimensional non-linear viscoelastic constitutive model, that describes the shear softening observed in brain tissue in simple shear stress relaxation experiments, has been applied in a FE head model. First the capability of this model of describing the transient behaviour of nearly incompressible viscoelastic materials under impact conditions was illustrated using a simple physical model experiment. Subsequently, the effect of non-linear material behaviour on the predicted brain response in a three-dimensional head model has been investigated by applying both linear and non-linear viscoelastic variants of the same constitutive model in an existing FE head model.

It was concluded that the addition of non-linear brain tissue material characteristics in current FE head model subjected to a posterior-anterior eccentric rotation, increases deviatoric strain values by approximately 21% and decreases stress levels by 11%. Their temporal and spatial distribution showed only minor changes. The pressure response remained unaffected by application of non-linear behaviour. In fact, we saw that the pressure gradient is completely determined by the equilibrium of momentum, and thus independent of the choice of the brain constitutive model. The final observation was that strains in the brain model were approximately ten times larger than was expected on the basis of published model simulation results at higher external loading. This is probably caused by the shear stiffness used in our model which was a factor ten lower than those used in head models in literature but comparable to recent material measurement data on brain tissue.

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REFERENCES


APPENDIX

A. STRAINS IN THE PHYSICAL MODEL

This appendix contains the strain(rate) requirements and the verification of these requirements in the physical model response.

Strain(rate) requirements for physical model

Animal injury data suggests strain levels ranging from 0.15 to 0.21 to be responsible for axonal injury (Bain & Meaney, 2000; Galbraith et al., 1993; Thibault et al., 1990) and breakdown of the blood brain barrier (Schreiber et al., 1997). The strain rate associated with injury is unknown but is estimated from data by Thibault et al. (1990). An angular acceleration, $\omega$, of 15 krads$^{-2}$ was associated with Diffuse Axonal Injury at a change of angular velocity, $\Delta\omega=150$ rad/s (Thibault et al., 1990). The typical time duration for this impact is estimated as, $T_{\text{impact}}=\Delta\omega/\dot{\omega}=10$ ms. If we assume that indeed shear strain values of 0.2 occurred during these experiments and that a constant strain rate was present, a lower bound estimate of the typical strain rate becomes, $\dot{\gamma}=\frac{1}{T_{\text{impact}}} = 20$ s$^{-1}$.

Strain verification

A lower bound estimation of the strain(rate) levels present during the physical model experiment is obtained by assuming that the marker motion is two-dimensional in the image plane and that a simple shear exists between the marker and the cup edge. From this the principal strain histories are derived analytically (Brands, 2002) in Figure 14. For the open
cup the maximum principal strain values range from 0.10 to 0.49 while for the closed cup experiments strain values of 0.09 and 0.15 were found. The maximum strain rate values were determined using numerical differentiation and equal 45 s\(^{-1}\) for the open cup and 20 s\(^{-1}\) for the closed cup. Although these values are rough estimates, it is concluded that both maximum strain as well as strain rate in the physical model are within the range associated with the occurrence of injury.

B. MESH DENSITY CONSIDERATIONS

The effect of spatial discretisation on the numerical results is investigated by repeating the simulation of the open cup experiments with both a more coarse and a more refined mesh. The coarse mesh is obtained by merging eight elements of the reference mesh into a single, larger element. Mesh refinement is achieved by splitting every element in the reference mesh into eight new elements. The results in Figure 15 show similar time courses as obtained with the reference mesh. The average maximum value of the angular displacements decreases by 9.7% when the coarse mesh is used. Refining the reference mesh provides a 4.0% increase of peak values. The moment of time on which these maxima are reached also increases with mesh density -3.9% (-1.6 ms) and +1.0% (+0.44 ms) with respect to the reference value respectively.

The radial displacements show an increasing trend when the mesh density is increased but remain in the experimental range. This indicates that the results of the reference mesh can be used to indicate where and when peak strain or stress values occur.

FE-head model

The mesh density in the FE head model is characterized by the number of elements over the head diameter. For the brain section in the model 20, 22 and 22 elements are used in anterior-posterior, superior-inferior and lateral direction respectively. The physical model simulations showed that typically 28 elements over the diameter are needed for accurate prediction of the dynamical response of a physical model. As a result, the computed solutions will not be at maximum accuracy. However, the physical model simulations showed that a coarse mesh (14 elements per diameter) did not alter the nature of the response. For this reason it is believed that the response predicted by the model will reflect trends due to different constitutive modelling correctly.