The dynamical behaviour of a model material for brain tissue

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The dynamical behaviour of a model material for brain tissue

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Coaches: • dr. ir. P.H.M. Bovendeerd
         • ir. D.W.A. Brands
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1. General introduction

The head is identified as the body area most frequently involved in life threatening injuries in traffic accidents [ESTC-report, 1993] and is considered to be the most critical body region in crash-situations because of the often irreversible nature of injuries to the central nervous system [Brands, 1997].

It is not yet completely understood in which way an external mechanical load, applied on the human head, leads to head and brain injury. For this we must know which phenomena take place inside the human head during impact situations. Unfortunately the local response cannot be recorded in-vivo, and for this reason numerical models of the human head under impact conditions are developed [Claessens et al., 1997]. Developing a numerical model requires knowledge of the constitutive behaviour of human brain tissue and information on relevant dynamical phenomena during impact situations (e.g. pressure, shear strain, and wave phenomena). Mimicking the constitutive behaviour of brain tissue can take place by the use of model materials. On the other hand, a model material can be applied in an experimental validation of a numerical model. Moreover, the use of a physical model instead of actual brain tissue has several advantages, such as reproducibility, availability and (extended) storage life.

The mechanical properties can be determined by loading the brain tissue. Shear loading experiments are a common method to characterize both viscous fluids and soft solid tissues (like brain tissue). Moreover, shear loading is an interesting area of research, because shear strains are supposed to be most damaging to the brain [Holbourn, 1943], because of its relatively low shear modulus and its nearly incompressible behaviour.

In the quest for a suitable model material for brain tissue, in this report the mechanical properties of various gelatine mixtures will be investigated on the basis of some simple rheological experiments. A 20% weight gelatine mixture has already been applied in various tests by NATO in order to mimic the dynamical behaviour of soft tissue. Now it has to be determined whether the dynamical behaviour of the gelatine correlates with the dynamical behaviour of human brain tissue.

In chapter two an overview of the theoretical background will be given for understanding the experimental results. In chapter three the mechanical properties of brain tissue as found in literature are presented. Chapter four deals with the experiments on some gelatine mixtures. In chapter five we will look for alternative materials and in chapter six another experimental technique will be investigated. Finally, in chapter seven the experiments will be evaluated and conclusions will be drawn. Also recommendations for further investigation will be given.
2. Small strain material functions

2.1 Introduction

Rheological experiments of (biopolymeric) gels can be subdivided into two groups:
- small deformation tests, to investigate the viscoelastic properties within the linear regime;
- large deformation tests, to acquire complete stress-strain relations and to probe failure criteria.

We now will concentrate on the former, i.e. the small deformation tests. Within a certain regime, where the strain $\gamma$ does not exceed a critical value $\gamma_c$, the gelatin will behave as a linear viscoelastic material. In order to show linear viscoelastic behaviour, a material must meet two requirements: *proportionality* and *superposition*. These requirements will be elucidated briefly.

Proportionality: This requirement is met when an increasing excitation results in a proportional increase of the response.

Superposition: The superposition principle states that the response to a set of combined excitations is equivalent to the sum of responses to each excitation separately.

On the basis of these two criteria a general relation between stress $\sigma(t)$ and strain $\gamma(t)$ can be formulated:

$$\sigma(t) = \int_{t'=0}^{t} G(t-t') \dot{\gamma}(t') dt'$$  \hspace{1cm} [2-1]$$

where $G(t)$ is the relaxation modulus and $\dot{\gamma}(t) = \frac{\partial \gamma}{\partial t}$.

2.2 Relaxation experiment

The physical significance of $G(t)$ is best explained by means of a relaxation experiment. In such an experiment the strain is changed stepwise from zero to $\gamma_0$ at time $t = t_0$. This is illustrated in figure (2-1) and (2-2).

![figure (2-1) strain step $\gamma_0$](image1)

![figure (2-2) typical stress behaviour](image2)
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The relation between stress \( \sigma(t) \) and strain \( \gamma(t) \) for a viscoelastic material now is given by

\[
\sigma(t) = \gamma_0 G(t)
\]  
[2-2]

and consequently

\[
G(t) = \frac{\sigma(t)}{\gamma_0}
\]  
[2-3]

In case of a liquid \( G(t \to \infty) = G_m = 0 \), in case of a solid \( G_m > 0 \).

A small change in strain \( d\gamma \) leads to a small change in stress \( d\sigma \) according to

\[
d\sigma = G \cdot d\gamma
\]  
[2-4]

This can be rewritten as

\[
d\sigma = G \cdot \frac{d\gamma}{dt} \cdot dt = G \cdot \dot{\gamma} \cdot dt
\]  
[2-5]

Integrating this expression brings us back to equation [2-1].

2.3 Dynamical experiments

In dynamical experiments the load applied on the test sample varies in time. Often harmonic loading functions are used. In this paragraph some dynamical experiments will be discussed.

2.3.1 Sinusoidal oscillations

An important dynamically applied load is a harmonically varied strain according to:

\[
\gamma(\omega, t) = \gamma_0 \sin(\omega t)
\]  
[2-6]

Where \( \gamma_0 \) is the maximum amplitude of the strain.

Because of the energy dissipated by the (partial) viscous behaviour, a phase difference \( \delta(\omega) \) between stress and strain arises:

\[
\sigma = \sigma_0 \sin(\omega t + \delta)
\]  
[2-7]

Differentiation of eq. [2-6] yields

\[
\dot{\gamma}(\omega, t) = \omega \gamma_0 \cos(\omega t)
\]  
[2-8]
Substitution of equation [2-8] in equation [2-1] and some rewriting results in:

\[ \sigma(\omega, t) = \gamma_0 \left( G'(\omega) \sin(\omega t) + G''(\omega) \cos(\omega t) \right) \]  

where

\[ G'(\omega) = G_\infty + \omega \int_0^\infty [G(t) - G_\infty] \sin(\omega t) dt = \frac{\sigma_0}{\gamma_0} \cos(\delta(\omega)) \] 

\[ G''(\omega) = \omega \int_0^\infty [G(t) - G_\infty] \cos(\omega t) dt = \frac{\sigma_0}{\gamma_0} \sin(\delta(\omega)) \]

In these expressions \( G' \) is called the elastic or storage modulus: it determines the elastic part of the stress, according to the storage of mechanical energy; \( G'' \) is called the viscous or loss modulus: it determines the viscous part of the stress, according to the dissipation of mechanical energy.

Furthermore \( \frac{G''}{G'} = \tan(\delta(\omega)) \).  

In case of a pure elastic behaviour \( \delta = 0 \), in case of a pure viscous behaviour \( \delta = \frac{\pi}{2} \).

The complex modulus \( G^* \) is defined as:

\[ G^* = G'(\omega) + i G''(\omega) \]

Furthermore

\[ |G^*| = \sqrt{(G')^2 + (G'')^2} \]

\( |G^*| \) is often referred to as the dynamic modulus \( G_d \).

### 2.3.2 Sinusoidal strain rate

Another way to view the same experiment is in terms of a sinusoidal strain rate \( \dot{\gamma}(t) \) according to

\[ \dot{\gamma}(t) = \dot{\gamma}_0 \cos(\omega t) \]

Substituting this expression in equation [2-1] and some rewriting results in

\[ \sigma(\omega, t) = \dot{\gamma}_0 \left( \eta'(\omega) \cos(\omega t) + \eta''(\omega) \sin(\omega t) \right) \]

In this case a complex viscosity \( \eta^* \) can be defined as:
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\[ \eta^* = \eta' - i\eta'' \] [2-17]

where \( \eta' \) is the dynamic viscosity and \( \eta'' \) is the elastic part of the complex viscosity

Furthermore

\[ |\eta^*| = \sqrt{(\eta')^2 + (\eta'')^2} \] [2-18]

Comparing equations [2-9] and [2-16] it can be seen that

\[ \eta' = \frac{G^*}{\omega} \] [2-19]

and

\[ \eta^* = \frac{G'}{\omega} \] [2-20]

The following relation between \( G^* \) and \( \eta^* \) can be formulated:

\[ G^* = i\omega\eta^* \] [2-21]

and also:

\[ \tan(\delta(\omega)) = \frac{G^*(\omega)}{G'(\omega)} = \frac{\eta'(\omega)}{\eta^*(\omega)} \] [2-22]

The dynamic viscosity \( \eta' \) and the steady flow viscosity \( \eta \) (which is the viscosity in case \( \dot{\gamma} \) is constant), are nearly identical at low frequencies or rates of shear.
3. Literature study on the dynamical properties of brain tissue

In this chapter the mechanical properties of brain tissue are presented as found in literature. It has been shown [Galford and McElhaney, 1970] that brain tissue behaves like a nonlinear viscoelastic material. Nevertheless the storage modulus $G'$ and the loss modulus $G''$ often are presented as linear material parameters.

In order to investigate the global behaviour, our main interest is the order of magnitude of the storage modulus $G'$ and the loss modulus $G''$.

3.1 Reported brain tissue properties

Figure (3-1) below shows a graphical survey of the reported values of the storage and loss modulus of brain tissue. The values are derived from human brain tissue, porcine brain tissue and brain tissue from a calf.

![Graphical survey of reported brain tissue shear properties](image-url)
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The values are acquired by Dynamic Torsional Shear (DTS) methods as well as Dynamic Simple Shear (DSS) methods. See also table 3.1 below.

<table>
<thead>
<tr>
<th>Reference</th>
<th>Method</th>
<th>Reported values [kPa]</th>
<th>Notes</th>
</tr>
</thead>
</table>
| [Fallenstein et al., 1969]¹ | DSS | $G' = 0.6-1.1$  
$G'' = 0.35-0.60$ | Human brain tissue  
• Frequency 10 Hz |
| [McElhaney et al., 1972]² | DSS | $G' = 0.437-0.95$  
$G'' = 0.35-0.60$ |  
• Frequency 10 Hz |
| [Shuck et al., 1970]³ | DTS | $G' = 0.83-1.38$  
$G'' = 0.3-3.8$ |  
• Frequency 2-400 Hz |
| [Shuck and Advani, 1972]⁴ | DTS | $G' = 7.6-33.9$  
$G'' = 2.7-81.0$ |  
• Frequency 10-350 Hz |
| Arbogast et al., 1985]⁵ | DSS | $G' = 0.35-0.95$ | Human brain tissue  
• Frequency 20-100 Hz |
| [Arbogast and Margulies, 1997]⁶ | DSS | $G' = 1.25-1.75$  
$G'' = 0.50-2.10$ |  
• Frequency 20-200 Hz |
| [Thibault and Margulies, 1996]⁷ | DSS | $G' = 1.00-1.50$  
$G'' = 0.28-1.60$ |  
• Frequency 20-200 Hz |
| [Meulman, 1996]⁸ | DTS | $G' = 0.145-0.40$  
$G'' = 0.028-0.167$ |  
• Frequency 0.159-15.9 Hz |

Table 3.1 Summary of reported brain tissue shear properties

¹: Fallenstein et al. (1969). Samples were placed between two oscillating shear plates using an electromagnetic exciter. Because of the relatively large mass of the apparatus only experiments at a resonant frequency of about 10 Hz were performed.

²: McElhaney et al. (1972). Dynamic simple shear experiments were performed.

³: Shuck et al. (1970). Dynamic torsional shear experiments were performed.

⁴: Shuck and Advani (1972). A comparison was made between the complex modulus estimated for samples from the corona radiata (purely white matter) and the thalamus (grey matter). Shear moduli are obtained from torsional response experiments. It appeared that white matter could be considered isotropic. On the other hand, grey matter showed variations in storage modulus for different cut directions and was also different from white matter. The loss modulus showed negligible variation in grey matter and small variation in white matter:

<table>
<thead>
<tr>
<th>Cut direction:</th>
<th>1</th>
<th>2</th>
<th>3</th>
</tr>
</thead>
<tbody>
<tr>
<td>White $G'$ [Pa]</td>
<td>$7.65 \times 10^3$</td>
<td>$6.96 \times 10^3$</td>
<td>$4.34 \times 10^3$</td>
</tr>
<tr>
<td>$G''$ [Pa]</td>
<td>$2.62 \times 10^3$</td>
<td>$3.24 \times 10^3$</td>
<td>$3.51 \times 10^3$</td>
</tr>
<tr>
<td>Grey $G'$ [Pa]</td>
<td>$10.6 \times 10^3$</td>
<td>$6.27 \times 10^3$</td>
<td>$4.13 \times 10^3$</td>
</tr>
<tr>
<td>$G''$ [Pa]</td>
<td>$1.45 \times 10^3$</td>
<td>$1.51 \times 10^3$</td>
<td>$1.38 \times 10^3$</td>
</tr>
</tbody>
</table>

Table 3.2 Values for storage modulus and loss modulus for different cut directions  
[Shuck and Advani, 1972]

The reported differences were considered small enough (about 7%), so that average values for white and grey matter in all directions were used in reporting the results.
5: Arbogast et al. (1995) Tests using porcine brainstem samples were performed on a stress relaxation shear device, at strain rates $>1\ \mbox{s}^{-1}$, to three levels of peak strain (2.5%-7.5%). Experiments with samples originating from the pons were performed. In this part of the central nervous system (CNS) the fibres are supposed to be organised mainly in longitudinal direction (transversely isotropic). The results of the experiments suggested that regions of the CNS that are characterized by a predominant axonal fiber direction possess different material behaviour parallel and transverse to this direction. This suggestion was not accompanied by a histological investigation of the samples.

6: Arbogast and Margulies (1997). The material response of two regions of the porcine central nervous system (brainstem and cerebrum) was measured in order to examine any regional variation. Complex shear moduli were calculated over a range of frequencies at two engineering strain amplitudes (2.5% and 5.0%) At a 2.5% strain, the complex modulus $G^*$ showed no statistical difference between regions. For 5.0% strain, the regional difference in $G^*$ was statistically evident. The storage modulus of the brainstem samples was approximately 100% greater than that of the cerebral samples. According to the authors these region specific differences, when combined with a numerical model, can contribute to an enhanced understanding of traumatic brain injury mechanisms.

The moduli presented in table 3.1 are the averages of the values for brainstem and cerebrum.

7: Thibault and Margulies (1996). Tests were performed in order to quantify the age-dependent material properties for porcine brain tissue. Brain tissue samples were obtained from 2-3 day old and 1 year old domestic pigs. The complex shear modulus $G^*$ was measured in a custom-designed oscillatory shear testing device, at a shear strain amplitude of 2.5% from 20-200 Hz. The elastic and viscous components of the complex shear modulus change significantly with the development of the cortical region of the brain. The dynamical properties of pediatric and adult porcine brain tissue differ up to 100%, as the moduli increase with age.

The moduli presented in table 3.1 are the averages of the values for adult and pediatric brain tissue.

8: Meulman (1996). Torsional shear experiments with calf brain tissue were performed for frequencies between 0.159 and 15.9 Hz at a shear strain amplitude of 1.0%. The brain tissue was considered to behave as a linear viscoelastic isotropic material. In order to obtain information about the constitutive behaviour at frequencies relevant for impact (20-1000 Hz), the application of time/temperature superposition was investigated and found to be applicable. See also [Peters et al., 1997].

Ljung (1975) investigated the material properties of brain matter in terms of the shear modulus $G$ and the kinematic viscosity $\nu$. Values were determined experimentally in transient rotation yielding the values $G = 1700\ \mbox{Pa}$ and $\nu = 0.009\ \mbox{m}^2/\mbox{sec}$. These data are not presented in figure (3-1), as this method involved a rotational acceleration pulse and not a continuous sinusoidal excitation.
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4. Shear experiments on gelatine samples

4.1 Experimental setup

The material properties of the gelatine test samples are investigated by means of a Rheometrics Fluids Spectrometer II (RFS II). A schematic view of the RFS-II is given in figure (4-1). The RFS II enables measurements of the viscous and viscoelastic properties of the samples. It applies a steady rotational or dynamic oscillatory shear to the sample, which is placed between two parallel plates. By measuring the torque transmitted through the material it is possible to determine values for $G'$ and $G''$ [Meulman, 1997], according to

$$G'(\omega) = \frac{2HM_0}{\pi R^4 \alpha_0} \cos(\delta(\omega)) \quad [4-1]$$

$$G''(\omega) = \frac{2HM}{\pi R^4 \alpha_0} \sin(\delta(\omega)) \quad [4-2]$$

where $H$ is the sample height, $M_0$ the amplitude of the measured torque, $R$ the sample radius, and $\alpha_0$ the amplitude of the angular displacement of the lower plate.

![Schematic overview of the RFS II](image)

Figure (4-1) Schematic overview of the RFS II

The frequency range is limited by inertial effects in the RFS II. Shear experiments take place at frequencies between 0.0159 and 15.9 Hz (0.1-100 rad/s).

The experimental protocol consists of the following stages: preparation, handling and measurements. In the following sections these stages will be discussed.
4.2 Preparation of the gelatine samples

The gelatine mixture was prepared following a recipe from T.N.O. Prins Maurits Laboratorium. The gelatine powder was supplied by Gelatine Delft N.V. In appendix A a description of the gelatine preparation is given, for both 20% and 4% gelatine.

4.3 Handling

20% gelatine mixture

After preparation the gelatine was poured into a test tube with an inner diameter of 10 mm, was sealed air-tight, and then was cooled down. By carefully breaking the test tube a cylindrical sample with a diameter of 10 mm was acquired. Slices with a thickness of 1-2 mm were cut out of this cylinder in order to enable a direct comparison with the results of Meulman [Meulman, 1997].

4% gelatine mixture

In the case of a 4% gelatine mixture it was no longer possible to break the test tube, as the gelatine was damaged immediately as a result of strong adhesion to the tube wall. To solve this problem at first the inside of the test tube was covered with Teflon®. In this way the gelatine sample could carefully be pushed out of the test tube. Cutting the slices however was no longer possible, due to the low stiffness of the material. To avoid the above mentioned problems, the gelatine mixture was poured between the two shear plates of the rheometer, directly after preparation (i.e. in liquid condition at 50 °Celsius). The samples were then cooled down to the testing temperature of 10 °Celsius. Using this method, samples with a diameter of 50 mm were tested.

4.4 Measurements

First the boundaries of the linear regime were examined by means of a dynamic strain sweep default test. In this test the sample is subjected to an increasing strain amplitude at a constant frequency.

Secondly, the gelatine test samples are subjected to a dynamic frequency sweep test. In this test the samples are sheared at frequencies between 0.0159 and 15.9 Hz (0.1-100 rad/s), at a strain level of 1%. In the following two sections the experimental results of both 20% and 4% gelatine mixture samples will be presented.

4.4.1 Experimental results for 20% gelatine samples

Measurements of the complex modulus $G^*$ as function of the strain amplitude demonstrated that both parameters showed no correlation up to a strain level of at least 1% (see also appendix B). It therefore can be concluded that a strain level of 1% the gelatine sample behaves as a linear viscoelastic material. This statement holds for both 20% and 4% gelatine mixtures.

As stated in the general introduction, we will start the experiments with 20% gelatine. Results for $G'$ and $G''$ are presented in figure (4-2) and (4-3) respectively. Figure (4-4) shows the behaviour of the phase angle $\delta(\omega)$.
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Figure (4-2) Elastic modulus $G'$ as function of the frequency for 20% gelatine

Figure (4-3) Viscous modulus $G''$ as function of the frequency for 20% gelatine
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Clearly, experimental values for $G'$ are about two decades higher than those for brain tissue. Values for $G''$ are of the same order of magnitude as can be found in literature. In other words, the 20% gelatine does possess the viscous behaviour of brain tissue, but is too stiff. We therefore adjusted the material properties by changing the mixture composition.

4.4.2 Experimental results for 4% gelatine samples

The results from experiments with eight different samples are presented in figures (4-5), (4-6), and (4-7) for $G'$, $G''$, and $\delta$ respectively.
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Figure (4-5) Elastic modulus $G'$ as function of frequency for 4% gelatine

The solid lines in the figures represent the average values for all samples, dots represent each measured value.

Figure (4-6) Viscous modulus $G''$ as function of frequency for 4% gelatine
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Figure (4-7) Phase angle $\delta$ as function of frequency for 4% gelatine

4.5 Discussion

We started our experiments with a 20% gelatine mixture, as it had already been applied by NATO in various tests. From the experiments we can conclude that a 20% gelatine mixture indeed is a viscoelastic material, but with an elastic modulus that is about two decades too high in comparison with human brain tissue. We therefore altered the mixture composition and experimented with a 4% gelatine mixture. Our first goal, to decrease the value for the elastic modulus, was reached. From figure (4-6) it can be seen that the elastic modulus $G'$ decreases to a value of about 10 kPa, which is comparable with data from literature.

A simultaneous effect however was a substantial decrease of the viscous modulus, which resulted in a very small phase angle. Figure (4-6) shows values for the viscous modulus $G''$. According to figure (4-7) the phase angle $\delta$, which can be seen as the ratio of the viscous modulus and the elastic modulus, becomes very small. From this we can deduct that we are dealing with an almost perfectly elastic material, as the viscous properties become negligible.

Figures (4-6) and (4-7) however also indicate experimental errors, directly related to the resolution of the angular transducer. Values for the phase angle are of the same order of magnitude as the resolution of the angular transducer. Because of this, values for the phase angle can become negative, which accounts for the suddenly decreasing values for $G''$. 
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Also the effect of temperature was investigated. Figure (4-5) shows the different behaviour of the gelatine at 10° Celsius and 15° Celsius. With increasing temperature, the elastic modulus decreased with an amount of about 25%.

Because of inertial effects of the RFS II rheometer, the maximum testing frequency is 100 rad/s, or 16.9 Hz. At higher frequencies (above 45 rad/s or 7 Hz) inertial effects already are playing a role and are thus troubling the outcome of the experiments. This can be seen in the presented figures where values for $G'$ and $G''$ show an unexpected increase. Moreover, we are interested in the dynamic properties at relevant frequencies for impact situations, in which case the order of magnitude is [kHz].

In short, two major problems have occurred during the experiments:

- The dynamical properties of the model material used in the experiments do not get near the properties of actual (human) brain tissue. Variation of the mixture composition offers no solution.
- The testing device RFS II is not equipped for simulating relevant (impact) situations. When approaching the maximal frequency of the RFS II, the obtained data are no longer reliable.

In the following section we will look for solutions for the above mentioned problems. Another model material for brain tissue will be investigated and an alternative experimental technique, ultrasonic testing, will be evaluated.
5. Reported properties of model materials for brain tissue

In the preceding chapter the need for an alternative model material became clear. Literature study revealed the frequent use of a silicone gel as a model material for brain tissue.

[Thibault et al., 1982] analyzed the strains in physical model materials as a result of inertial loading. The surrogate brain material was Sylgard Gel (Dow Corning). A baboon skull was filled with the gel. The polymer/catalyst ratio used in the reported experiments was 50/50, which seemingly best approximated the mechanical behaviour of fresh brain tissue. Further explanation for the use of this material is lacking and no references are mentioned.

[Margulies et al., 1985] used the Dow Corning Silicone Gel System as a surrogate brain material to fill an (idealized) physical skull model. They studied the intracranial content deformations due to impulsive, distributed, angular accelerations applied to the head. Experiments were carried out with a pure-slip boundary and a no-slip boundary condition (to simulate strong skull/brain adhesion), with the goal of finding an appropriate boundary condition for the modelling of Diffuse Axonal Injury (DAI). It appeared that the no-slip boundary model restricted large strains to the periphery, while the pure-slip boundary model produced significant strains in the deep hemispheric region. In order to reproduce the pathological lesion pattern of DAI, a pure-slip model with a more life-like skull configuration would have to be developed. [Margulies et al., 1990] developed such a model with the use of the Dow Corning Silicone Gel System, mixed in a 1:1 ratio of polymer to catalyst. Real baboon and human skulls were filled with the gel. Indentation experiments with a small rigid indenter were performed on the gel to determine the Young’s modulus of the model. The values of the gel varied by a factor of three within the range of reported properties of brain tissue in literature. No exact data are mentioned.

[Meaney et al., 1994] also used the Dow Corning Sylgard Gel as a surrogate for brain tissue. In their experiments a physical skull model was filled with the gel. The silicone gel material was used because (1) it exhibited mechanical properties similar to those of brain tissue, (2) it was optically transparent and (3) it was self-adherent, so that a series of layers could be cast to form a continuous material. For the first property it was referred to [Blum et al., 1985], but no information whatsoever about the mechanical properties of the gel is found there.

[Viano et al., 1997] simulated brain responses in closed-head impact situations. During head impact the motion of grid-points in the silicone brain gel was filmed by high speed cameras. From the digitized films the gel displacement and Green-Lagrange strain were calculated. A transparent silicone gel from Dow Corning (Dielectric Gel Q3 6527) now marketed as Dow Corning Sylgard A/B was used to simulate the brain. The gel consisted of two components A and B that were mixed 50%/50%. Testing of the gel determined that $G' = 1.0 \text{kPa}$ and $G'' = 0.75 \text{kPa}$ at 2-4 Hz. $G'' = 0.3 \text{kPa}$ at 2 Hz increasing linearly to 0.5 kPa at 5 Hz. The authors state that these properties are close to the values for brain as determined by [Ljung, 1975] and [Fallenstein et al., 1969] as described in section 3.1. The properties did not vary between tests at room temperature and $35^\circ \text{C}$. The brain model consisted of a cylindrical vessel with a tight-fitting lid and radius comparable to the average radius of the skull in different planes.
The two components of the gel were mixed, poured into the vessel up to half of the depth of the skull-vessel, and cured at room temperature. Then markers were placed on the surface of the gel and finally a top layer of gel was prepared and poured filling the container. In the experiments again a distinction is made between a free-slip and a non-slip situation.

[Arbogast et al., 1997] experimented with samples of viscoelastic Sylgard silicone gel (Dow Corning) in a high frequency shear device. Three Sylgard gel mixtures with consistencies similar to soft biological tissue were prepared and tested: 1:1, 1:2, and 1:3 ratios of polymer to catalyst, respectively. Increasing proportions of catalyst produced a material of increasing stiffness. The master curve (after time-temperature superposition) of the complex shear modulus at a reference temperature of 20°C for a 1:1 gel shows values for the complex modulus $G^*$ from $9 \times 10^2 \text{ [Pa]}$ at a frequency of 0.16 Hz up to $6 \times 10^3 \text{ [Pa]}$ at a frequency of 160 Hz. An increase in sample thickness above a critical level erroneously increased the measured complex shear modulus due to inertial effects.

Although the Sylgard gel is a frequently used brain surrogate material, only few authors provide data of the material properties in order to justify the use of it. [Viano et al., 1997] and [Arbogast et al., 1997] do provide these data, that are of the same order of magnitude as found in literature (see chapter 3). Therefore further research on the use of Sylgard gel as a brain surrogate material is recommended.
6. Ultrasonic testing techniques

Numerical models strongly depend on values for the material properties of the brain tissue. Because of the low stiffness and highly viscous behaviour of brain tissue, the parameters are difficult to assess using standard mechanical testing techniques.

Ultrasound techniques are pre-eminently suited for non-destructively assessing the properties of (bio-)materials. While the magnitude of the deformation in ultrasonic tests does not approach that seen in closed-head injuries, the high frequency of the measurements results in a high strain rate, which cannot be matched using (traditional) large deformation techniques without causing damage to the tissue.

When dealing with small time intervals, as is the case during impact situations, stress and strain are considered in terms of wave propagation. The velocity of an ultrasonic wave through a material is related to the density of the material and its elastic properties.

For a homogenous, isotropic, elastic material the longitudinal and transverse wave propagation velocity (\( c_L \) and \( c_T \) respectively) can be expressed as [Hempenius, 1995], [Lin et al., 1997]:

\[
\begin{align*}
  c_L &= \sqrt{\frac{E(1-\nu)}{\rho(1+\nu)(1-2\nu)}} \quad \text{or} \quad c_L = \sqrt{\frac{1}{\rho} \left(K + \frac{4}{3}G\right)} \\
  c_T &= \sqrt{\frac{E}{2\rho(1+\nu)}} \quad \text{or} \quad c_T = \sqrt{\frac{1}{\rho} G}
\end{align*}
\]  

[6-1]

and

\[
\begin{align*}
  c_L &= \sqrt{\frac{E(1-\nu)}{\rho(1+\nu)(1-2\nu)}} \quad \text{or} \quad c_L = \sqrt{\frac{1}{\rho} \left(K + \frac{4}{3}G\right)} \\
  c_T &= \sqrt{\frac{E}{2\rho(1+\nu)}} \quad \text{or} \quad c_T = \sqrt{\frac{1}{\rho} G}
\end{align*}
\]  

[6-2]

where \( E = \) elastic modulus or Young's modulus  
\( \rho = \) density of the medium  
\( \nu = \) Poisson's ratio  
\( G = \) shear modulus  
\( K = \) bulk modulus

According to [6-2] transverse ultrasound is directly related to the shear properties of (brain) tissue.

[Stelwagen, 1998] carried out experiments with transverse ultrasound in 1.5% gelatine-in-water-solutions. The transverse ultrasound propagation velocity was experimentally determined by both pulse-echo and through-transmission methods. Pulse-echo methods employ a single transducer to measure the time of flight of a reflected wave. Through-transmission techniques employ a pair of transducers to measure the transit time for the wave in a single direction.

At a frequency of 1 MHz it appeared to be impossible to detect a signal of a transverse ultrasonic pulse in the gelatine solution. It was assumed that the damping of transverse ultrasound at 1 MHz was too high to detect a signal.
According to [Sutilov, 1984] the attenuation $\alpha_r$ of transverse waves in viscous liquids can be expressed as:

$$\alpha_r = \frac{\pi f \rho}{\eta_s} \text{ [dB/m]}$$  \[6-3\]

where $\eta_s = \text{dynamic viscosity} = \eta'$, see also equation [2-19]

From [6-3] it followed that the attenuation of transverse ultrasound (at 1 MHz) in low-viscous liquids is extremely high ($>10^4 \text{ dB/mm}$), which accounted for not detecting a transverse pulse in the gelatine solution.

As the attenuation $\alpha_r$ is proportional to the square root of the frequency, [Stelwagen, 1998] stated that it should be possible to detect transverse ultrasound at lower frequencies. Further it is questioned whether [6-3] is applicable to gelatine-in-water-solutions.

Attenuation of an ultrasonic pulse results from a combination of reflection, scattering and absorption mechanisms [Lin et al., 1997].

- **Reflection** is dependent on the difference in the acoustic impedances of two adjoining material types. It occurs at the boundary between a specimen and the coupling medium or at the interface between the specimen and a large included inhomogeneity.
- **Scattering** results from the interaction of ultrasonic waves with small inhomogeneities.
- **Absorption** is due to the conversion of mechanical energy to thermal energy, as the wave travels through the medium. It is related to the damping properties of the material.

[Etoh et al., 1994] studied bovine brain tissue properties by the measurements of longitudinal ultrasonic wave propagation and the mechanical transfer function. The ultrasonic attenuation of bovine brain tissue increased monotonously as the frequency was decreased. This is confirmed by findings of [Lin et al., 1997]. It is stated that the large ultrasonic attenuation must arise from some relaxation phenomena or from particular structures in the brain tissue, e.g. the existence of lamellar structure.

The longitudinal mechanical transfer function $Z(\omega)$ is equal to the complex Young’s modulus:

$$Z(\omega) = E' + iE''$$  \[6-4\]

The imaginary part of Young’s modulus is expressed as:

$$E'' = \omega \eta \tilde{\eta}$$  \[6-5\]

where $\tilde{\eta}$ is the viscosity of the longitudinal mode.

Measurements of $Z(\omega)$ lead to the conclusion that both the real and imaginary part show finite values at the low frequency limit.
A finite value for the real part of Young's modulus is reasonable, because brain tissue is a soft but solid material. The finite value for the imaginary part however is anomalous; it would mean a divergence of viscosity at zero frequency, as lead from [6-5].

Again it should be noted that the layer compression modulus of lamellar structures such as myelin sheath may also contribute to Young's modulus of brain tissue. The ultrasonic velocity and the real part of Young's modulus indicated some dispersion, suggesting the distribution of relaxation phenomena.

As already stated the application of ultrasound is a promising technique in measuring brain tissue properties. Longitudinal ultrasound provides information on the bulk and shear moduli of the material, as transverse ultrasound directly is related only to the shear properties. In order to overcome the problems described in this section further research has to be done.
7. Conclusions

In our quest to find a suitable model material for brain tissue, we started our experiments with a 20% gelatine mixture, which had been used by NATO as a surrogate material for soft tissue. The 20% gelatine appeared to be too stiff, i.e. the elastic modulus was about two decades too high. We therefore adjusted the mixture components and experimented with a 4% gelatine mixture. The elastic modulus (as expected) decreased, but also the viscous modulus decreased. From the phase angle it could be concluded that we were dealing with an almost perfectly elastic material. Clearly, by changing the mixture composition, we could not compose a suitable material with desirable properties.

Furthermore, measurements with the Rheometrics Fluids Spectrometer II (RFS II) were problematic because of the limited frequency range and the unreliable data at the upper end of the frequency limit.

Summarizing, two problems have arisen during this project: gelatine mixtures are not a suitable model material for brain tissue and measurements with the RFS II are problematic.

As an alternative model material, a frequently used material is Sylgard gel (Dow Corning), a silicone gel system. The earliest literature references provide no arguments for the use of this gel. In the article of [Thibault et al., 1982] it is stated that the gel, in a ratio of 1:1 of polymer to catalyst, shows brain tissue-like behaviour. Other authors refer to this article without further research. [Viano et al., 1997] and [Arbogast et al., 1997] however provide data of measured properties that justify the use of the Sylgard gel, as it possesses mechanical properties close to those of actual brain tissue. Their results reveal potential good behaviour when compared with figure (3-1). Further research on the use of Sylgard gel as a brain surrogate material therefore is recommended.

Alternative experimental methods are ultrasonic measurements. Experiments with longitudinal and transverse ultrasound have been carried out. Longitudinal ultrasound provides information on the bulk and shear moduli of the material. Transverse ultrasound is an interesting area of research, because of the direct relationship with the shear properties of a material. Further investigation has to be done whether transverse ultrasound is a suitable way of measuring, as damping seems to be a problem.
The dynamical behaviour of a model material for brain tissue

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Appendix A

Preparation of 20% and 4% gelatine mixtures

- Measure off cold water and gelatine powder in a mass ratio of 4:1 or 24:1 (for 20% and 4% gelatine mixtures respectively)
- Add the gelatine powder to the water in a (small) cup
- Let the mixture expand for 45 minutes
- Put the fluid in a warm water bath and heat it up to about 50° Celsius
- When a clear transparent fluid has formed, it can be poured into the final form (e.g. a test tube)
- Remove possible bubbles (ultrasonic)
- Seal the test tube (air-tight) and let the mixture cool down
- Place the test tube in a refrigerator (preferably at 10° Celsius)
- Keep the samples cooled as long as possible and do not remove them from the refrigerator until just before the experiment
- Preferable testing temperature is 10° Celsius
Appendix B

Determination of the linear viscoelastic regime

In the experimental protocol the gelatine test samples first are subjected to a dynamic strain sweep default test. In this test the samples are subjected to an increasing strain amplitude at a constant frequency of 1.59 Hz (or 10 rad/s). The complex modulus $G^*$ was measured as a function of the strain amplitude $\gamma_0$. Results for the 20% gelatine are presented below in figure (B-1).

![Graph](image)

Figure (B-1) Dynamic strain sweep default test

At a strain level of 1% the complex modulus $G^*$ and the strain amplitude $\gamma_0$ show no correlation. Therefore we can conclude that the gelatine sample behaves as a linear viscoelastic material at a 1% strain level.