The IST Programme
Project No. 10378

SimBio

SimBio - A Generic Environment for Bio-numerical Simulation

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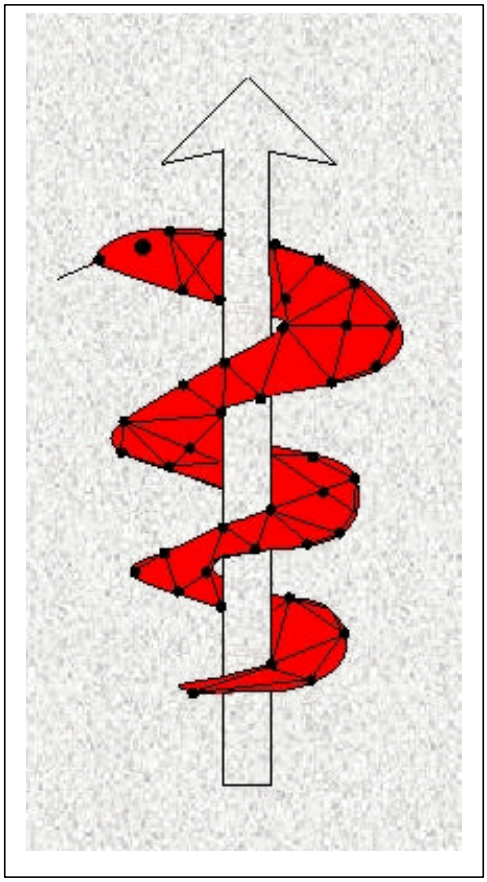

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Material Database Design Report

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1. Introduction

Modelling and simulations for biomedical applications is a demanding task, which requires an adequate and reliable material database to achieve meaningful results. So workpackage 2 ‘Compilation of the Material Database’ will provide the basis for all the other work performed in is this project. Since the projected applications include as well mechanical as electrical simulations the database has to include detailed data about both these tissue properties. We started our work by collecting information about human tissue properties and experimental approaches to gather this data. As will be discussed later available existing knowledge and data is in no terms sufficient for simulations of living systems. So the main part of the work in this workpackage has to be the development of appropriate experimental set-ups and execution of the according experiments to generate the needed base of normal, physiological data.
Furthermore the project aims to model electrical and mechanical properties of human tissue of individuals for diagnostical and therapeutical purposes which means that applied methods need to be non destructive and as non invasive as possible. For this reasons magnetic resonance strain imaging (MRSI) and diffusion tensor imaging (DTI) have been chosen for the basic approach. Both methods basically rely on magnetic resonance imaging (MRI) and have been developed very recently. Their major advantage in comparison to previous methods is the possibility to make investigations in-situ and in-vivo. Data acquired and evaluated in this manner provide real life information for the envisaged models and simulations in contrast to the so far mainly effected ex-vivo measurements.

2. Living tissue properties

Common biomechanical experiments are designed to gather data about visco-elastic material properties as Young’s modulus, Poisson’s ratio, shear modulus etc.. For this purpose mainly devices as traction/compression machines, viscometer, … are used [Fun93]. All these methods demand the isolation of representative parts of the tissue under investigation and its handling in the experimental environment. This means essentially that the part under investigation has to be separated (disconnected) from its environment: the inspected subsystem is isolated, input and output of energy, material and information is interrupted. Oversimplified: Experiments are performed on dead tissue and can only provide data for this regarded state only. To understand better the meaning of this we will have a closer look at living tissues properties.
Mechanical properties of bone and cartilage are governed by hard substructures exhibiting mainly solid state elastic properties which are due to long term changes only. That is why for this type of tissue extraction and isolation of probes doesn’t change their properties significantly. In contrast soft tissue presents itself as an agglomeration of three well separated, even so closely interconnected and interwoven subspaces formed by more or less weak membranes and filled up with slightly different liquids.
These subspaces are the

- *intravascular*,
- *interstitial*, and
- *intracellular*.

The *intravascular* volume can be regarded in a first approximation as a pure fluid exhibiting viscose properties. It’s volume depends on the equilibrium between artery influx and venous efflux res.; pressure.
Due to the high whole-body interconnectivity of the vascular network every change in intra- or extravascular pressures result in a local displacement of the blood filling up this subspace and thereby a change in volume and composition of the regarded volume of tissue. These changes occur with time constants in the range of milli-seconds. Even more if the blood flow rests as is mandatory if one takes a tissue probe this subvolume of up to 10% changes its properties due to the intra vascular coagulation process from fluid to visco-elastic one which means a pronounced change in the overall mechanical properties.

The *interstitial* volume is balanced by capillary filtration and reabsorption and lymphatic efflux. Due to the limited transport-capacity, volume changes of this subspace occurs in the minutes regime or even longer.

Finally, the *intracellular* fluid has to be taken into account. It is mainly governed by the difference in pressure - mainly the osmotic - between the intra- and extracellular space. Since this intracellular subspace forms the last link in the above presented chain, forces applied externally to a tissue sample will result finally in a reduction of cell water and thereby cell volume also.
So, applying an external force to a soft tissue sample leads to a time depending reduction in volume due to expression of water. Own experiments showed after application of a physiological load during ca. 30 minutes a continuing loss of up to 50% of volume before a quasi-static state was reached. Investigation of mechanical properties make sense only under defined boundaries. This state can be reached in this sort of experiments only by establishing a quasi static state here after 30 minutes after a pronounced structural change.

In conclusion, one has to state that most standard methods used for determination of mechanical properties of viscoelastic solids may be appropriate for investigation of bones, cartilages, and structures rich of collagen as blood vessel walls but will give more or less meaningless results for soft tissue if one aims for information about the living state. Even if accepted and done without choice further problems arise in soft tissue investigation: the application of traction like machines requires that at least part of the probe is of sufficient stability to apply forces in a well defined manner. E.g. for brain tissue especially of interest in this project, Miller and Chinzei state that [Mil94]: 'Brain tissue is very soft and adheres upon contacting almost any material. Consequently it was very difficult to obtain an exact cylindrical shape.'
Further problems are the dependency of soft tissue properties in:

- temperature,
- content of water in general,
- enzymatic activity: pronounced changes in properties occur due to persistent enzyme-activity,
- anisotropy e.g. of white matter - strain-hardening
- nonlinearity due to the complexity of biological structures in general.

All this problems together are the reason that biomechanical literature is full of data for ‘solid’ tissue but lacks reliable information for real soft tissue. [Fun93]
So, since we aim to generate a database of living soft tissue properties we have finally to carry out experiments with living tissue in situ. Such a method is not yet available at least not for tissue deeper inside the body e.g. as the brain. Here elastography methods and especially MRSI offers a promising approach[Bil98, Fow95].
The ultimate goal of our work is therefor to develop an elastography method based on MRSI for in situ, in vivo evaluation of soft tissue mechanical properties in the normal physiological state, which might be applied for individual diagnostics also. Steps we plan to take to realise such a method and to apply it for generation of a data base will be described in the following chapter in detail.

3. Bio-mechanical material database

Detailed knowledge about individual material behaviour of human tissue will be measured, calculated and included in the material database. Parallel to acquiring new material knowledge by applying innovative magnetic resonance techniques, data of material properties are collected from literature. Particular features required by the SimBio evaluation applications will be given priority.
Steps in realisation will be:

- data extraction from literature,
- small sample measurement using a traction machine
- Magnetic Resonance Strain Imaging (MRSI) of ex vivo coated tissues samples
- MRSI experiments with isolated ex-vivo organs: brains
- MRSI in vivo: strains induced in-vivo for evaluation of in vivo material properties.

3.1 General approach

For the biomechanical part of the material data generation an stepwise approach will be realised: Firstly, a general material database will be extracted from literature. This will provide the basis for the initial simulations performed in WP7.
A method to investigate small samples of soft tissue as harvested by biopsy in clinical routine will be established to provide a calibration tool for the following techniques.
MR Strain Imaging (MRSI) is a method recently invented and until now only applied to some isolated ex-vivo organs as in gel embedded kidneys. So here first experimental steps will be the generation of relevant data by determining material properties of isolated ex-vivo brains, parts, entire, and menisci, respectively [Che98, Eme95, Oph99].
In the final step, in-vivo investigations are planed to perform based in-vivo investigations by inducing defined strain and evaluating the resulting deformation in situ. This third step in material properties acquisition will be the most innovative, advanced, and demanding.
The degree of realisation achievable during the lifetime of the project is a function of the developments made. In any case, 'in-vivo MRSI' will give us directly the mechanical property information we need to realise the goal of our project: to combine the individual geometrical, mechanical, and properties of an individual patient and to introduce them into the individual simulations.

3.2 Literature-based data

It turns out that the biggest problem in data collection and presentation is the big variety of methods and boundaries used to determine soft tissue material properties. Since there are no standards every investigator has to and will define his own method. Boundaries as static – dynamic loads, application and kind and degree of preload, type and range of dynamic force/strain applied, and most important the way of tissue sample harvesting and treatment before experimental evaluation. These experimental conditions are chosen individually depending on the special interest of the project.
This diversity is the reason that experimental data reported in literature is hardly comparable and has to be evaluated in a most critical manner to assure relevance for the here intended use in modelling and simulation of soft tissue. In the appendix we present the most appropriate data collected so far. Further evaluation and verification will be necessary and carried out in light of our own experiments that will be performed in the upcoming project phase. Major attention will be taken in regard to anisotropy of mechanical properties, which until now rarely has been taken into account. The data presented in the Appendix include properties of a variety of soft and hard tissue mainly that to be found in the neuro-cranium.
This first general material database shall provide the basis for the initial simulations performed in WP7.

3.3 Database

3.3.1 Data presentation

The structure of the material database will be adapted to the needs of the different modelling and simulation activities in the project and facilitate rapid inclusion of new data for extended applications. The database format will be defined in conjunction with the interfacing activities of WP 6. Anyway since we start already with a first preliminary data base in this report that is based on literature values only. So, at the moment we have adapted the way of data presentation to the one used generally in literature. Here data is most often given as single values which are evaluated under specific boundary conditions. If the nonlinearity of stress-strain relationship is taken into account also, the data is most often represented in graphs. Anisotropy is normally not treated in a systematic way.

Our data elevation and presentation will be done in different steps. Most reasonable boundary conditions will be chosen and particular experiments will be performed resulting in single values. If possible and reasonable anisotropy will be taken into account also and the data will be presented in tensor form.
This will be done for the exvivo traction experiments as well as for MRSI and as far as realisable for invivo MRSI. For traction experiments this can and will be done for the both, static and dynamic load.

Investigation of dynamic properties will be done in the next future by our traction machine treating small samples of tissue exvivo. While this data will be published in one of the next edition of our data base, it will not be possible during the runtime of the project to perform dynamic experiments by the invivo MRSI method.

3.3.2 Data encoding

To facilitate access and use of the database a tissue property classification scheme will be invented. A code will be used which identifies:

- type of tissue: bone, muscle, …..
- material properties characteristics: linear, static, isotrop, …
- way of sample preparation: invivo, deep frozen, …..
- type of experimental setup used: traction machine, US, …

This code will have the following form:

XXX-xxxx-XXXX-xxxx

It will accompany every material property value given, to enable simple choice of adequate data for a special model. This type of presentation will augment the transparency of given data and should allow to compare values from different sources in a more easy and meaningful way.
We aim to present the complete code description and the first version of our encoded database in the next database release.

3.4 Small sample machine

Direct measurement of mechanical properties of small tissue samples is mandatory since an important step in developing the MRSI technique is evaluating its performance. For this evaluation, an alternate and reliable way of measuring tissue mechanical properties is needed. There are numerous advantages of having a device that can measure these properties of small tissue samples. The most important of which might be the possibility to perform first investigations in anisotropy properties if tissue samples are obtained in an appropriate way.

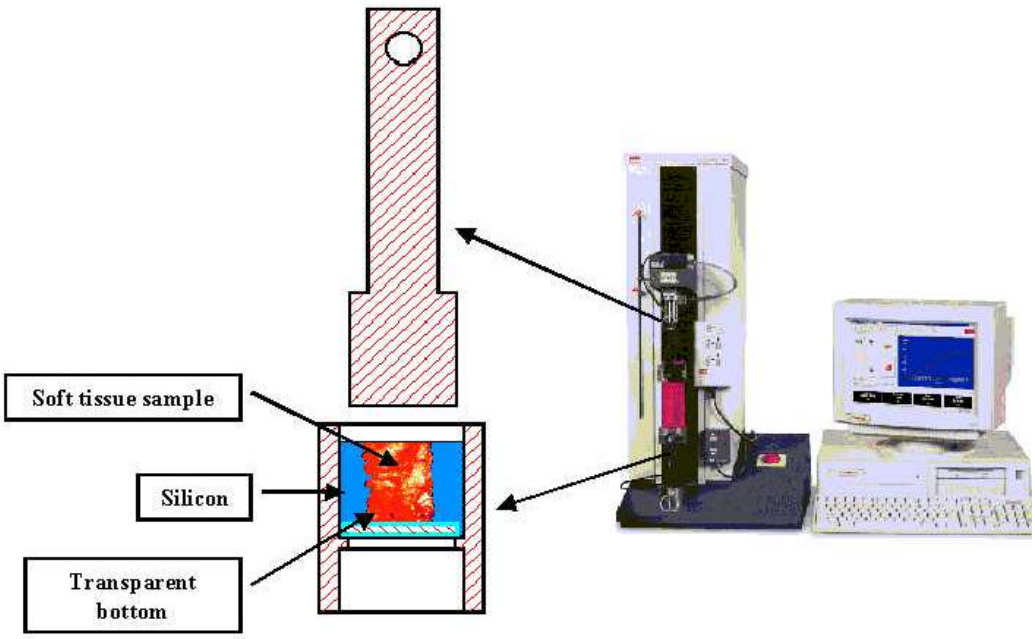

Figure 1: Traction machine modified to perform compression experiments with embedded soft tissue samples.

A system capable of accurate measurements on small (~0.4 cc) cylindrical tissue samples has been integrated in our commercial traction machine and first feasibility studies have been carried out.
To achieve good accuracy, the standard clinical procedure of biopsy is used to harvest the samples. They are placed in a silicone mould by which procedure a large portion of the boundary is fixed in a well defined position by constraining the side boundaries of the sample. The mould is then placed in a cylindrical chamber filled with physiological buffer. Deformations are produced by lowering a circular piston, with a radius bigger than the sample radius smaller than the exterior cylinder, onto the centre of the sample. Deformation and force are applied and evaluated by routines integrated in the traction machine.
The system will be calibrated and tested using rubber samples of known elasticity. In addition an analytical description based on the simple rotational symmetry has been developed, which allows the exact calculation of the results to be expected. These algorithms will be used to confirm the calibration measurements and to investigate the influence of sample geometry imperfections on system accuracy.

3.5 Magnetic Resonance Strain Imaging - Principles

3.5.1 Magnetic Resonance Strain Imaging MRSI - the method

Until now strain imaging techniques have been applied for some isolated 'in vitro' organs (kidneys) only [Erk98, Eme95]. So we will set-up this technique for the "in-vivo brain", which will provide detailed knowledge of material properties of 'in place' substructures in the brain in general.

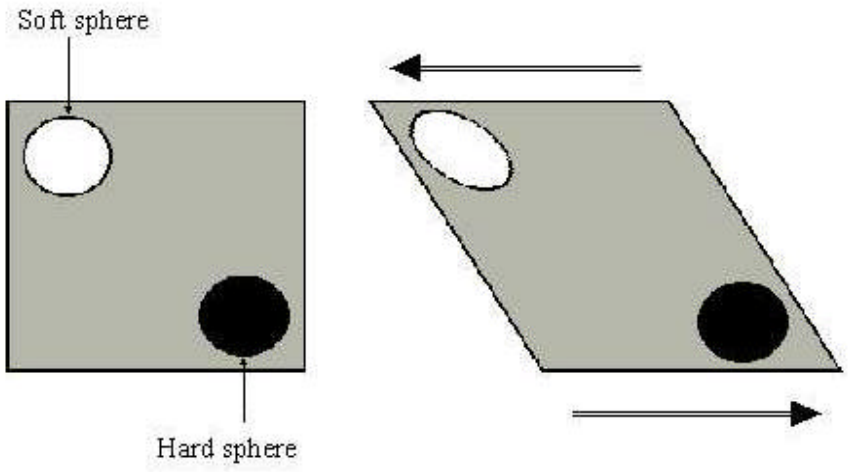

Figure 2: Principle of the Magneto Resonance Strain Imaging MRSI technique. A soft and a hard sphere are embedded in a material of known medium mechanical property. After applying an external force the properties of the unknown spheres can be evaluated by comparising deformations.

Furthermore the projected 'advanced in vivo strain imaging' technique of patients brain will provide most important insights in individual mechanical brain properties and processes for diagnostics and therapy. Development and installation of this method will be a demanding part of the project and at the moment it is unforeseeable which state of realisation will be reachable during the runtime of the project.

3.5.2 Theoretical model - Constitutive equation

To evaluate mechanical properties of a body on the basis of stress-strain experiments firstly a constitutive equation has to be established. A general approach can be developed as follows:
The general equation of equilibrium of a body is given by:

$$\sum_{n=1}^{3} \frac{\partial s_{ij}}{\partial x_j} + f_i = 0, \quad i = 1,2,3$$

Stress-strain relation of compressible media can be written

$$s_{ij} = \left(K - \frac{2}{3} G \right) q d_{ij} + 2Ge_{ij}$$

Stress-strain relation of incompressible media:

$$s_{ij} = pd_{ij} + 2Ge_{ij}$$

In our approach we aim to establish a constitutive modelling as presented by Miller and Cinzei (1997) for investigation of brain tissue by a compression machine with an extension comparable to the experimental set up published by et al. []for shear measurements. This will enable us to apply the above presented equation for compressible media.

3.6 Magnetic Resonance Strain Imaging – Realisation

In the following the steps to be taken to realise and apply the MRSI-method will be discussed in more detail.

3.6.1 Experiments

3.6.1.1 Coating techniques for isolated organs

To achieve an appropriate coating of the biological tissue a material has to be chosen which fulfils the following demands:

- neutral in regard to living tissue,
- perfect coating in the liquid phase,
- short time polymerisation,
- no heating up during polymerisation,
- mechanical properties nearly the same as for the investigated tissue.

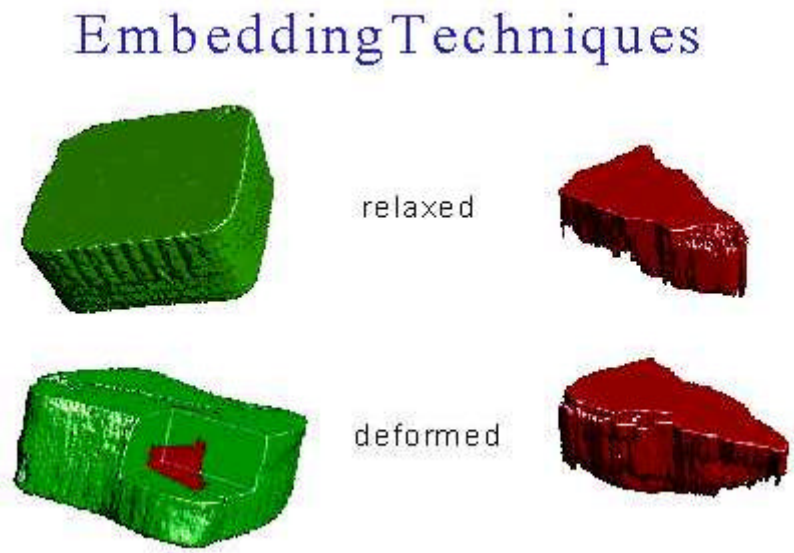

Figure 3: Embedding of a piece of muscle and deformation (preliminary experiment).

Silicon based materials turned out to be one of the first choice. Anyway a number of problems arises in application. First of all the liquid develops during the polymerisation process air bubbles which change the properties of the coating material in an unpredictable manner. Methods will be developed to avoid this problem.

3.6.1.2 Construction of deformation-cubes

A box, deformation-cube, has to be developed, in which the coated objects will be placed. This cube will enable the application of exactly defined strain and/or shear deformations of the coated objects. Since the objects under investigation will be of different size and the relationship between volume of object and coating material should be close to 1, a set of different sized boxes will be fabricated. To enable comparable experiments in three axis these boxes have to be of cubic shape.
Furthermore, care has to be taken to use material only, which can be used inside the MRI machine also.

3.6.1.3 Deformation-cube measurements of isolated homogenous structures

First experiments will be carried out using soft tissue samples of simple geometry which results will be directly comparable to data achieved by experiments performed on the small sample traction machine. This will provide the basis to validate the experimental approach of MRSI in general and calibrate achieved data.

3.6.1.4 Complex structures ex vivo

The next step will be the evaluation of complex structures. We will start with a kidney as already reported in literature to verify our experimental and algorithmic set-up. Substructures as cerebellum, medulla oblongata and a hemisphere will be coated, deformed, and evaluated as described [Eme95].

3.6.1.5 In situ measurement of structures

These experiments will use the experiences gathered by the MRSI-experiments done with the coated objects but will go already beyond. Entire post mortem organs are used where externally induced deformations are evaluated. As such it is a further step in direction of in vivo MRSI.

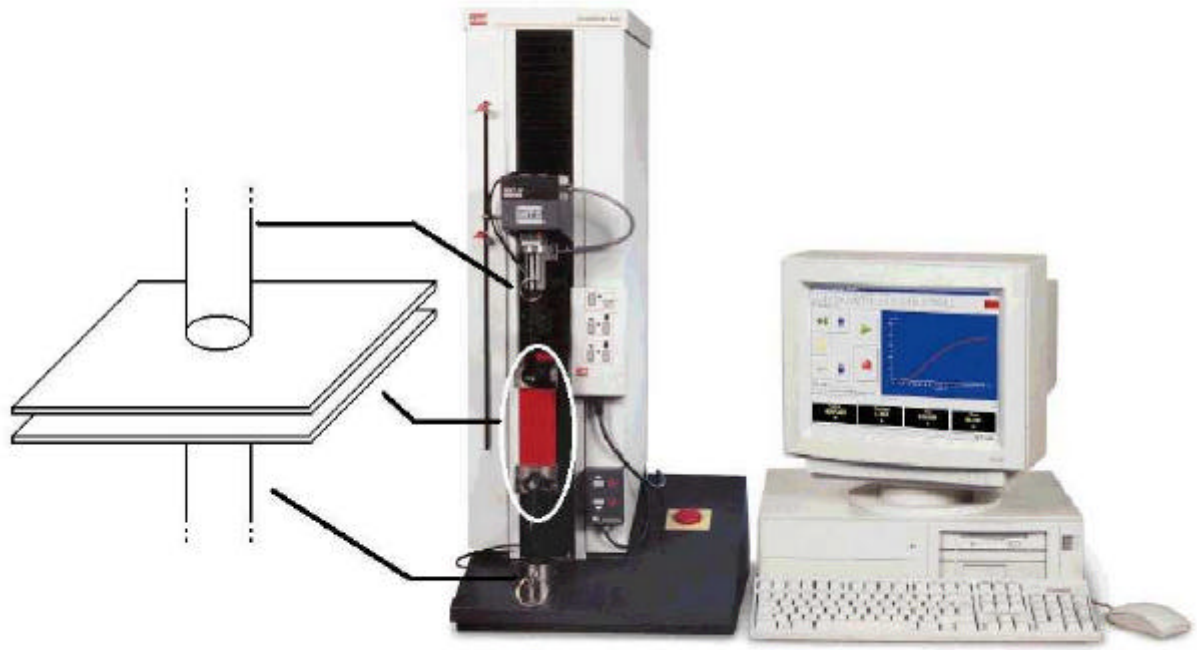

Figure 4: Traction machine equipped to apply compression to coated objects.

To establish models of the mechanical properties of the brain and perform realistic simulations one has to take substructures into account which will probably be not accessible by direct MRSI techniques. These includes the different membranes of which the tentorielle membranes will be the most important ones. To get insight in their influence and if possible to find a way to characterise indirectly their properties, we plan to perform experiments using entire animal heads post mortem. Image acquisition of the brain will be done in the relaxed state and under deformation due to externally induced artificial tumours; e.g. by placing a small balloon inside the cranium.
Of similar importance may be the axons. They are fixed in the grey matter and form bundles which connect different parts of the CNS. So they might be not only the transport system for electrical excitation but even so transmitter for mechanical traction between distant parts of the CNS. Experiments to investigate this hypothesis will be carried out in the same manner as described above.

3.6.1.6 In vivo measurement of structures

This is the technique which is most advanced and could even be introduced in daily clinical diagnostic routine. The basic idea is to apply the so far developed techniques by using forces and thereby introduced deformation which are generated by the body itself. Here the most important ones are

- heart activity with the periodic change between systolic and diastolic pressure and
- respiration which induces at least in the thorax pronounced pressure differences.

If it will be possible to quantify the hereby induced forces the evaluation of the related deformation as developed in the preceding steps will allow the determination of material mechanical properties. Anyway this ultimate goal will hardly be achievable during the runtime of this SimBio project.

3.6.2. Imaging

3.6.2.1 Image acquisition

High resolution MRI of regions of interest

For MRSI it is essential to detect and quantify changes occurring in object geometry due to applied forces as precisely as possible. The resolution needed will be in the region of tenth of mm depending on the kind and degree of deformation. Therefor the acquisition parameters have to be optimised for the actual tissues, which includes parameter optimisation as well as optimisation of antenna [Gow89].

MRI acquisition relaxed and deformed

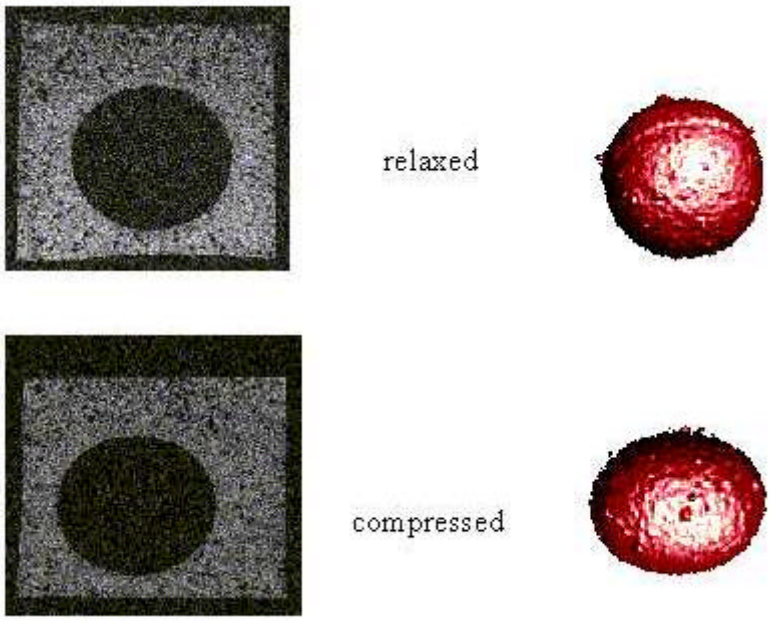

Figure 5: A spherical object has been embedded in silicon of known material property. An external compressing force has been applied. MRI sequences have been acquired in both the relaxed and deformed state. The object has been reconstructed in both states using our VolView-tool.

Synchronisation of MRI-acquisition with spontaneous rhythms: ECG & respiration

For optimisation of image acquisition in in-vivo applications it is most important to avoid movement artefacts. For invivo image acquisition it is therefor important to synchronise the MRI machine with heart activity and /or respiration and to acquire all images at a precisely defined point in time during the cycle. In extension this technique can be used as described in the following paragraph.

MRI acquisition of dynamic deformation

Image acquisition by MRI machines can be triggered by external events. To day a standard, this possibility is mainly used for synchronisation with heart action. All images of a sequence to reconstruct a volume are taken in the same phase of subsequent cardiac cycles. So artefacts due to movement of tissue should be minimised. If one takes in the same manner volume images in different phases of the cycle of an periodically applied external force, the images can be rearranged and composed to time depending sequence of volume-images showing the dynamic response of the object under investigation to the external force. Thus one has the possibility to investigate dynamic mechanical properties also, at least to certain extent.

3.6.3.2 Image pretreatment

Segmentation

For segmentation there are standard procedures available as already implemented in VolView. This include simple thresholding, region growing, texture analysis etc. Due to the demanding type of images we have to deal in this project with, it might be necessary to add special algorithms as developed in the consortium especially to segment brain tissue images.

Identification of ROI's

Here the already implemented standard procedures are e.g. clustering algorithms and special interactive region growing methods. Further improvement by knowledge based algorithms as already developed in the consortium will be applied if necessary.

3.6.3.3 Image treatment

"Matching" of ROI's of different images

This is a most important step since objects which have been identified in image volumes acquired during different states of deformation have to be related to each other. To prepare the following steps corresponding objects have to be positioned separately in a way that there rests no net difference in translation and rotation between them.

Determination of local strain

On base of the above performed matching of identified objects it becomes possible to apply different procedures which in principle all are based on optimisation. The search of global extrema by Monte Carlo algorithms and of local ones e.g. by steepest descent are available from our former work. In addition optimisation by relaxation as proposed by different authors [Zhu99, Fow95, Rob96] and available in the consortium will be tested also. This calculations will result in the needed field of local displacements (Sum95)

Determination of local stress and the tensor field of mechanical properties

The boundaries are given by the externally applied forces. This provides the starting configuration for the following iterative optimisation, which in the first step is based on the assumption of homogenous isotropic elasticity throughout the object as given by the known properties of the coating material. Differences between the model based on this assumption and the measured configuration are used to establish a cost function which is optimised in the iterative process by changes of the assumed material properties [Doy00, Fow95, Kal96].

The result will be the needed tensor field of mechanical properties describing the different objects under investigation.

4. Bioelectric material database

4.1 MR technique called Diffusion Tensor Imaging (DTI)

For the *bioelectric part of the material data generation,* a novel MR technique called Diffusion Tensor Imaging (DTI) approach will be utilized to generate material data.

DTI images of a BOI (especially the human head and for model validation purposes the animal head) will be measured to generate an anisotropic conductivity map. The relation between the water diffusion tensor and the conductivity tensor in various materials and tissue types can be derived from

the mathematical theory of homogenisation, especially the theory of porous media and the effective medium This relation depends on geometric aspects of the considered materials, on cell membrane properties and intra- and extracellular coefficients. Different approaches for the calculation of the conductivity map from measured water diffusion tensor images will be implemented to satisfy the special needs of various materials and tissue types.
MPI has already implemented different sequences for DTI-measurements in the human and its associate BMZ will further implement and test sequences which are especially designed for animal head DTI.

4.2 Development of an individual conductivity tensor map

Multimodal MR-imaging strategies in combination with high level registration and segmentation algorithms and the application of the mathematical homogenisation theory to the voxel-wise measured, anisotropic water diffusion leads to an individual conductivity tensor map of the considered BOI. This map accounts for an exact representation of various tissue types with regard to their specific anisotropic resistance, a prerequisite for modelling in e.g. EEG and MEG (see also Introduction to Biomagnetism). Pathological conditions such as e.g. skull holes and brain lesions like tumours or cerebral ischemia, strongly changing the resistance of the involved head tissues, can now be modelled individually and appropriately. This material modelling is highly innovative compared with today's volume conductor models (e.g based on the boundary element method), where e.g. the head layers brain, skull and scalp can only be assigned constant and isotropic resistance values. This will improve the diagnostic performance of EEG and MEG and will have a large impact not only on basic and clinical research but also on patient outcome in a variety of diseases.

5. Outlook - ***micro-scanning***

During the runtime of the project, advances in other methods of material modelling will be monitored and inclusion in SimBio investigated. One such option is the use of recently developed *micro-scanning* techniques for determining the material properties of hard and soft tissues. The constitution of local bone mechanical properties from sufficiently fine scan data will be programmed, where local bone material density sub-regions, or segments, will be identified from scanned data and their detected structure will be used to generate average local material constitutive data.

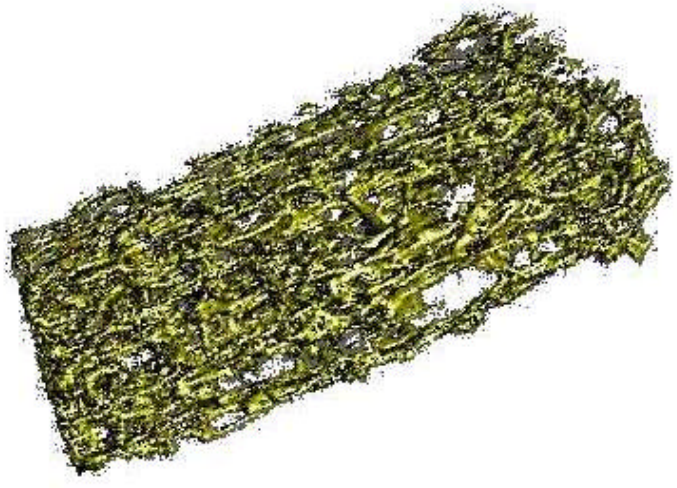

Figure 6: Sample of cortical-bone as acquired by high resolution MRI and reconstructed by VolView-software (CNRS).

This can be achieved, for example, by ‘local virtual material testing’, using fine sub-models of local virtual test pieces, modelled with their available fine grain structures, yielding as output the equivalent average material properties for the coarser finite element grids used in the full models. Similar procedures will be applied for the soft tissues. The advantage of this process is that for constituting average finite element equivalent material data, one can first use locally finer submodels with first principle material properties, such as the properties of the mineral content of bone tissue. These first

principle material properties can be expected to be approximately constant between individuals, whereas the local average bone structure varies in each bone and between individuals.

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Appendix

Mechanical properties of tissues

Soft tissues

Table 1 : Mechanical properties of tissues regarding linear elasticity

Author	Substructures[1]	E (kPa)	G (kPa)	K (kPa)	r (kg/m^3)	n	tan (d)
Lee [1987] Lighthall [1989] Ueno [1989,1991]			80.0		1000	0.475 & 0.49	0.2
Lee [1990]	brain (gel)[2]	80.0-121.2				0.49	
Chu [1991]	skull	6.5×10^6			2027	0.2	0.001
	Brain	250.0			1000	0.49	0.001
Ruan [1991a]	Skull	6.5×10^6			1412	0.22	
	Brain	66.7		2.19×10^2 - 2.19×10^6	1040	0.48 – 0.49999492	
	dura, falx, tentorium	31.5×10^3			1133	0.45	
	CSF	66.7			1040	0.499	
Ruan [1991b]	outer table		5.0×10^6	7.3×10^6	3000	0.22	
	Diploe		2.32×10^5	3.4×10^6	1750	0.22	
	inner table		5.0×10^6	7.3×10^6	3000	0.22	
	Brain		1.68×10^3	2.19×10^3	1040	0.4996	
	CSF		500	2.19×10^4	1040	0.489	
Ruan[3] [1992]							
Trosseille [1992]	Brain	240.0			1000	0.49 – 0.499	0.2
	CSF		0.2			0.49999	
Willinger [1992]	Skull	5×10^6				0.2	
	Brain	675.0				0.48	
	Subarachnoid space	5×10^{-2}				0.49	

E : Young's modulus ; G : shear modulus ; K : bulk modulus ; ρ : density ;
ν : Poisson ratio ; tan(δ) : loss tangent

[1] skull is not mentioned if rigid
[2] in 2-D and 3-D full-cylinder model
[3] no values given, probably identical to Ruan [1991b]

Table 2 : Mechanical properties of tissues regarding linear viscoelasticity

Author	Subtructures	E (kPa)	G_∞ (kPa)	G_0 (kPa)	b (s^{-1})	r (kg/m^3)	n	K (MPa)
Galbraith [1998] Tong [1989]	brain (gel)		5.512	11.02	200		0.4995	
Cheng [1990]	brain (gel)		16.2	49.0	145		≈ 0.5	
Lee [1990]	brain (gel) [1]		2.87 - 18	26.9 - 110	50	950		1.25 – 5.44
	CSF/ventricles[2]		3 - 6	9 - 24	50			0.445
DiMasi [1991a, 1991b]	Skull	2.4×10^6						
	Brain		17.225	34.45	100			0.0689
	Dura	6890						

E : Young's modulus ; G_∞ : static shear modulus ; G_0 : dynamic shear modulus ;
β : decay constant ; ρ : density ; ν : Poisson ratio ; K : bulk modulus

[1] used in half-cylinder, full-cylinder and skull physical models of Margulies [1987]
[2] used in skull physical models of Margulies [1987]

Cartilage

Table 3 : Mechanical properties of articular cartilage

Author	**Articular cartilage**	
	Poisson Ratio ν	Young's Modulus E (MPa)
Schreppers & Sauren [1990]	0.4	10
Lengsfeld [1993]	0.35	100
Bendjaballah et al. [1995]	0.45	12
Bendjaballah et al. [1997]	0.45	12

R = structure modelled as rigid

Table 4 : Mechanical properties of meniscus

Author	**Isotropic Meniscus**			**Meniscus Matrix**		**Meniscal Fibres**		**Transverse isotropic meniscus**	
	Poisson Ratio ν	Young's Modulus E (MPa)		Poisson Ratio ν	Young's Modulus E (MPa)	Poisson Ratio ν	Young's Modulus E (MPa)	Poisson Ratio ν	Young's Modulus E (MPa)
Sauren et al. [1984]	0.3	20/200							
Hefzy et al. [1987]	0.3	20/200							
Fithian et al. [1990] *		Parallel	90°						
		60	60						
		198	2.8						
		138	4.6						
Schreppers & Sauren [1990]	0.3	20						$\nu_{12} = 5.0$ $\nu_{23} = 0.3$	$E_1 = 50 / 20$ $E_2 = 10 / 5$
Spilker et al. [1992]								$\nu_{12} = 2.0$ $\nu_{23} = 0.05$	$E_1 = 200$ $E_2 = 0.055$
Lengsfeld [1993]	0.35	50/5							
Bendjaballah et al. [1995]				0.45	8		170 (deep) 60 (superficial)-		
Bendjaballah et al. [1997]				0.45	8		170 (deep) 60 (superficial)-		

Hard tissues

Table 5 : Mechanical properties of cortical and cancellous bone

Author	**Cortical bone**		**Cancellous bone**	
	Poisson Ratio ν	Young's Modulus E (MPa)	Poisson Ratio ν	Young's Modulus E (MPa)
Sauren et al. [1984]	0.2	500 / 5000	NG	NG
Hefzy et al. [1987]	0.2	500 / 5000	NG	NG
Schreppers & Sauren [1990]	0.2	500	NG	NG
Lengsfeld [1993]	0.35	10000	0.35	1000
Bendjaballah et al. [1995]	R	R	R	R
Bendjaballah et al. [1997]	R	R	R	R

R = structure modelled as rigid
NG = information not given

Electrical properties of tissues

Soft tissues

Brain matter, spinal cord, and cerebellum

Table 6 : Resistivity of brain white matter

Author	Animal	Frequency (Hz)	Pulse (ms)	Resistivity (Ω.cm)	
Crile et al. [1922]	Rabbit	1000		746	
Van Harreveld et al. [1963]	Rabbit	1000		957	
Nicholson [1965]	Cat	20		Normal	850
				Parallel	89
Freygang and Landau [1955]	Cat		0.3 → 0.7	333 (isotropic resistivity)	

Table 7 : Resistivity of brain grey matter

Author	Animal	Frequency (Hz)	Pulse (ms)	Resistivity (Ω.cm)
Freygang and Landau (1955)	Cat		0.3 → 0.7	222
Van Harreveld et al. (1963)	Rabbit	1000		208
Ranck (1963)	Rabbit	5		321
Crile et al. (1922)	Rabbit	1000		438

Table 8 : Resistivity of spinal cord

Author	Animal	Frequency (Hz)	Pulse (ms)	Resistivity (Ω.cm)	
Crile et al. [1922]	Rabbit	1000		576	
Ranck [1963]	Cat	5-10		Normal	1211
				Parallel	175

Table 9 : Resistivity of cerebellum

Author	Animal	Frequency (Hz)	Pulse (ms)	Resistivity (Ω.cm)
Crile et al. [1922]	Rabbit	1000		730

Muscle

Table 10 : Resistivity of muscle

Author	Specimen	Frequency (Hz)	Pulse (ms)	Resistivity (Ω.cm)	
Burger and van Dongen	Rabbit muscle			Longitudinal	125
				Transverse	1800
Rush et al. [1963]	Human arm muscle		100	Longitudinal	150
				Transverse	3000
Geddes and Baker [1967]					
Polk and Postow [1986]					

Fat

Table 11 : Resistivity of fat

Author	Animal	Frequency (Hz)	Pulse (ms)	Resistivity (W.cm)
Rush et al. [1963]			100 *	2500
Schwan and Kay [1956]	Dog	1000		1500 → 5000

* = d.c.

Scalp

Table 12 : Resistivity of scalp

Author	Specimen	Frequency (Hz)	Pulse (ms)	Resistivity (W.cm)
Geddes and Baker [1967]	Human scalp		*	230

* = d.c.

Bone

Table 13 : Resistivity of bone

Paper	Specimen	Fréquency (Hz)	Pulse (ms)	Resistivity (W.cm)	
Geddes and Baker [1967]	Thorax bone	low		16000	
Reddy and Saha [1984]	Wet bovine bone	10000		Axial	16600
				Circumfe-rential	36000
				Radial	54000

Fluids

Blood

Table 14 : Resistivity of blood

Author	Specimen	Fréquency (Hz)	Pulse (ms)	Resistivity (W.cm)
Rush et al. [1963]		low		162

Cerebrospinal fluid

Table 15 : Resistivity of cerebrospinal fluid

Author	Subject	Frequency (Hz)	Pulse (ms)	Resistivity (W.cm)
Geddes and Baker [1967]	Human	1000 – 30 000		65

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