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Determination of the stability of the cornea for describing the structural behaviour with a FEM-model

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Abstract

With refractive surgical interventions ametropia like myopia or hyperopia can be corrected. For getting a good vision of the eye optimal co-operation of the components is needed. The eye components are complex biological tissues and for describing the behaviour of the eye after surgical interventions it is necessary to have knowledge about the mechanical and optical behaviour of the specific components.
Refractive surgical interventions are operational techniques concerning the tissue of the cornea. Incisions or ablation of corneal tissue are done to get a defined modification of the corneal shape and therefore an improvement of the vision. Simulations of the structural behaviour of the cornea in combination with optical simulations of the eye will be used to describe postoperative results. The final aim should be an information and recommendation for the surgeons about the optimal surgery parameters.

The cornea is a heterogeneous tissue with complex material behaviour (anisotropic behaviour) and furthermore different from patient to patient. A FEM model will be used to describe the response of the cornea to refractive surgery. Basic research will be done concerning the structural mechanics and experimental techniques to get fundamental results in the field of biomechanical tissue behaviour of cornea to include the behaviour description into the FEM-model.

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Introduction

The biomechanical system of the human eye consists of the cornea at the front, the anterior chamber, a lens system, the vitreous and the retina with „fovea centralis" before signal processing in the brain (Fig. 1). The cornea is the most responsible part for the refractive behaviour (43 dpt). The so called "Gullstrand Eye" is used to consider these very simplified biological components.

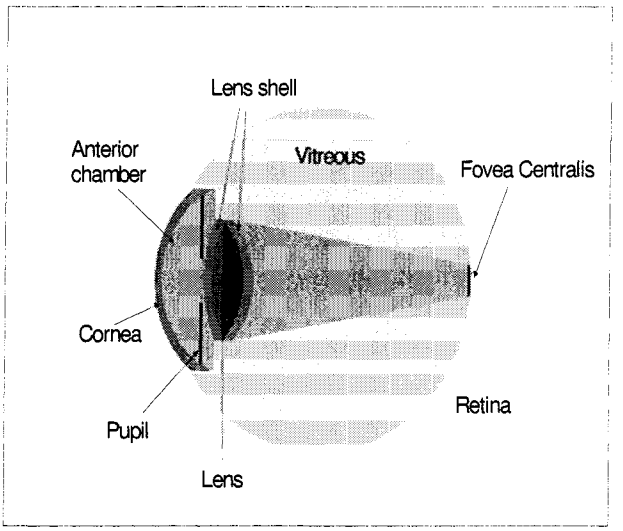

Fig 1: The "Gullstrand Eye" model

The most important intention by surgical interventions in the cornea is to manipulate the curvature of the anterior cornea by structural mechanical acts. The structural mechanical behaviour is influenced by different surgery type, i.e. by the laser parameters, incision parameters and geometric parameters, regarding different constraints and it depends on biological tissue in a very closed manner.

Before parameter optimisation there exists need of basic research concerning material dependent behaviour of the biological corneal tissue. To get more information about the complex material behaviour like stress-strain functions for the biological viscoelastic material and the stress of rupture point of corneal tissue, different measurements at porcine and human corneae would be necessary. Since the stress is direction dependent, we have to cut corneal tissue not only in different parallel layers but also out of different regions (central or near the limbus).

Because the employment of human corneae for experimental research is very critical and difficult first experimental series at porcine corneae are planned to minimise the number of experiments at human tissue. Before stretching the corneal lentils in a tension test machine, they have to be dissected and fixed on a special designed metallic foil. Experiments with porcine corneae should display

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the dependency on experimental parameters to draw conclusions to real physical values. The so determined material parameters must be integrated into the FEM models to predict the corneal response to refractive surgery.

Biological tissue of the cornea

The simulations relate to the biomechanical behaviour of the very complex corneal tissue and also on the attached optical effects based on the refractive behaviour at the deformed cornea surface.

To get more information about the complex material behaviour like stress strain functions for the biological viscoelastic material, different measurements at human cornea would be necessary. The problem is the availability of measured parameters only at post mortem corneae. The stability of the corneal components (epithelium, Bowman lamella, stroma, Descemet membrane and endothelium) indicate the curvature. Removal of tissue by PRK or LASIK disturbs the stability and changes the curvature. To compute the deformation at the corneal surface, finite element methods (FEM) are used.

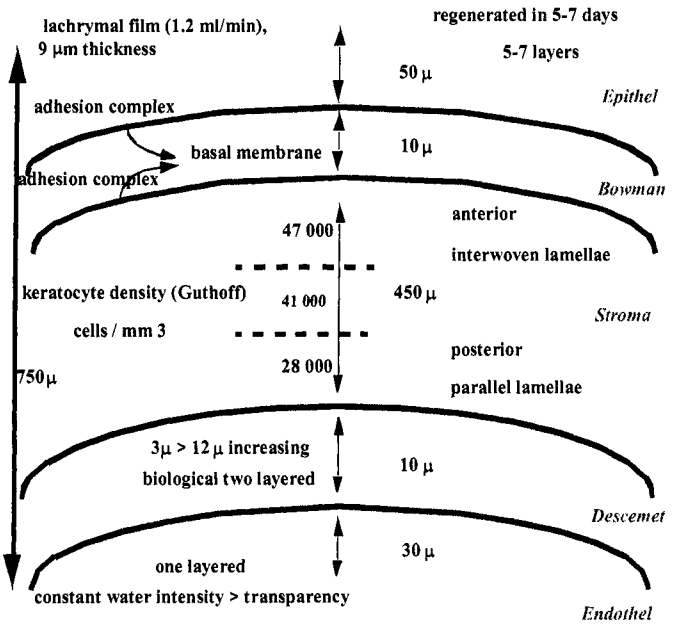

Fig 2: Biological layer architecture of a human eye

Normally the material behaviour of the biological tissue depends also on temperature. Experiments [1, 2] have shown, that as well transparency and also biomechanics of the cornea tissue in the range of 25°C up to 60°C are

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temperature independent. Between 60°C and 90°C there could be recorded an increased contraction, but that's outside of any cold light laser equipment for refractive surgical acts.

In the anterior part of the stroma alternating collagen lamellae are interwoven, but in the posterior part they are not. So, the stability effects are layer dependent different. An additional problem is the hydration, which has immediately consequences for the thickness (shrinking and swelling mechanisms). For post mortem analysis of the stroma different treatments (1 day in phosphate buffered saline or in demineralized water) leads to thickness increase up to 0.65 mm or decrease down to 0.4 mm. That is 50 % different and stress-strain experiments relates on forces and thickness parameters.

Fig. 2 shows the five main layers of the human eye with their biological characteristics.

Finite element models in the biomechanics

The real material parameters are very important for simulations of the corneal tissue. So, a material dependent refinement of the cornea is useful and necessary. From biological point of view, there are different biological layers with different geometrical thickness parameters. Furthermore each layer consists of different material behaviour like Young's modulus, Poisson ratio and so on [3]. The consequence is a consideration of these layer related geometrical and material parameters. For FEM-modelling, the net topology must guarantee a continuous transition in stress/strain parameters at the common nodes at material interface. In a first step, we use isotropic elastic material in each layer for simulation.

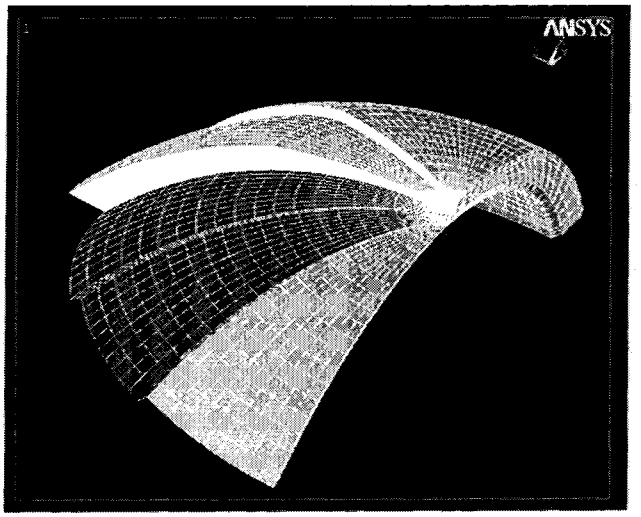

Fig 3: Layer dependent corneal mapped mesh

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Additional to the one layer model, consisting of a unique material solid, a parameterised model of layer related composition is performed [4,5]. For the given geometrical and material parameters, the number n of layers is specified. So, the different biological tissue composites can be regarded in a very fine manner (epithelium, Bowman lamella, stroma, Descemet membrane and endothelium). But also within the stroma the keratocyte densities can be regarded as material with own characteristics.

Concept of material parameterisation

In the literature, the models related to an estimation of curvature change due to artificial surgical intervention and the cornea are mostly considered as an uniforme sphere. Because of thin shell, the stresses are assumed to be constant through the thickness and shear transverse to the wall is negligible.

Under these conditions, the stress terms depend only on intraspheric pressure, radius of curvature and wall thickness (Laplace rule). In contrast to the sclera the cornea is bounded by a relatively inelastic and homogeneous layer on one side (Bowman lamella) and an elastic unhomogeneous layer on the other side (Descemet membrane).

Although here is used a first approximation, biological soft tissues by their nature are viscoelastic and nonisotropic [3]. The stress-strain functions perpendicular to the tissue plane wall are closed different from the in-plane function. A first analytical approach of a thin walled vessel (thickness < 0.1 radius), the IOP related in plane stress is given by the formula

$$\sigma = p \times r / 2\,t$$

with p pressure of the vessel, perpendicular to inner wall, r inner radius, t thickness, σ principal planar stress.

So, for a nonlinear human eye (total bulbus) with r =12 mm, t = 0.6 mm we get

$$\sigma_1, \sigma_2 = p \times 12 / 1.2 = 10 \times p$$

and because of $\sigma^2 = \sigma_1^2 + \sigma_2^2$ the IOP related stress is $\sigma = 14 \times IOP$.

To perform a biomechanical simulation, there are needed exactly three vector sets:

- parameter set of geometry
- parameter set of surgical specifications
- parameter set of materials

After such a simulation process the result vectors are given for each node in the mesh based on the FEM model. The structure of the vector is $(\Delta_x, \Delta_y, \Delta_z)$ for each node. The number of nodes depends on the solid mesh and must be kept variable.

In the first approach we used n=5 for the main biological tissue layers. Each of them is parameterised by the specific inner and outer surface, based on 2D-data and extruded in the 3D space by radial symmetrical rotation around the optical axis.

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The n-layered material vector is characterised by the components n (number of layers), E_1, E_2 ...E_n (elasticity of the layers), μ_1, μ_2 ... μ_n (Poisson ratio of the layers) and G_1, G_2 ...G_n (shear modulus of the layers).

Experimental strategy

In the past time, for simulations of the corneal behaviour mostly homogeneous material and/or tissue with constant thickness was used. But because of the complex interaction of the fibrils within the different layers also in one layer no homogeneity can be expected. Additionally, there must be distinguished between areas near the centre (optical axis) and near the limbus. There is different mechanics of the biological fibrils. Also the stress near limbus is direction dependent' that means, we expect radial stresses in other amount like border parallel stresses orthogonal to the former.

Thus a problem will be to cut corneal tissue in different parallel layers and out of different areas of the cornea (central, limbal) for experimental measurements.

Dissection of corneal tissue

From the ethic point of view obtaining human corneae for experimental research is very critical and difficult. It's easier to cut porcine corneae and to measure stress-strain parameters for material determination. To minimise the experiments at human tissue, first experimental series at porcine corneae are performed.

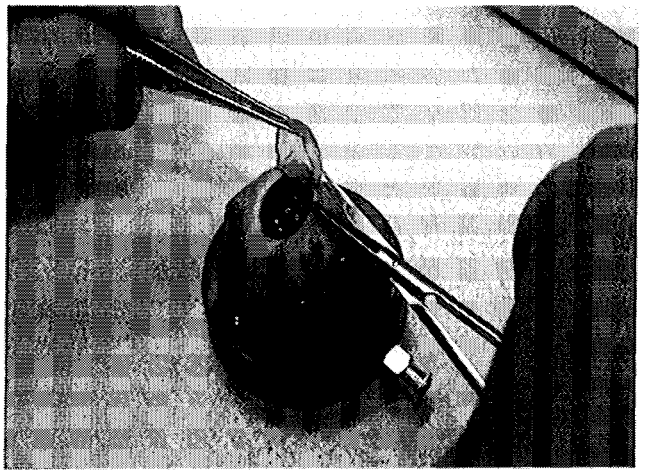

Fig 4: Dissection of cornea from porcine bulbus

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Probably basic results concerning layer dependent stability and structural behaviour may be transposed to human corneal tissue due to similarity, but numerical values have to be determined in special human corneal experiments.

Fig. 4 show bulbus of a pork with separation of the total cornea. This corneal tissue is the starting point for layer separation of different thickness. The difficulty is to separate layers with constant thickness down to 100 μm from the whole cornea of about 1500 μm (in contrast to the human cornea with a total thickness of about 500 μm).

A special equipment is necessary to guarantee such a high precise dissection. After fixation of the cornea at the metal block and under predefined pressure the shape of the cornea is stressed. A rotating cutter performs the separation of the 100 μm thick biological layer of the cornea. The thickness is variable adjustable (Fig. 5).

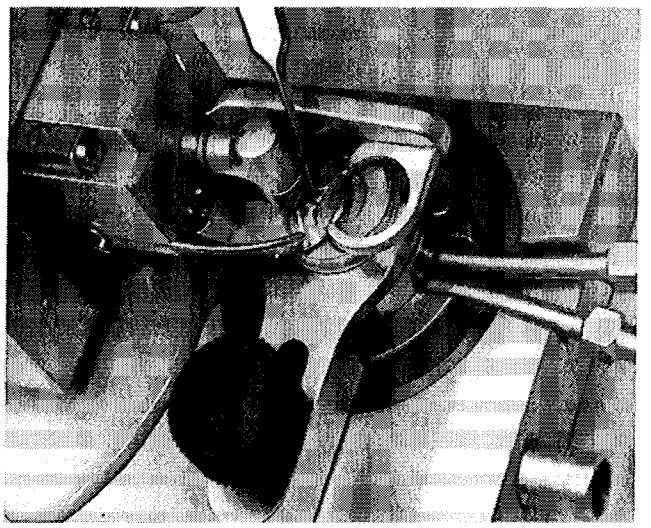

Fig. 5: Layer separation from corneal lentil

Great efforts are performed to determine layer dependent material parameters. After separation of the cornea from the bulbus, layer down to 0.1 mm thickness could be cutted from the total cornea. The very thin lentil are fixed on special designed steel foils produced for stress-strain experiments. The so determined material parameters must be integrated into the FEM models to predict the structure mechanical behaviour before surgical acts.

Before stretching in a tension test machine, the corneal lentils must be fixed. It is a special designed metallic foil of 0.1 mm thickness with holes at well defined positions, where the cornea layer is stitched with normal surgical threads. Fig 6 shows the metallic foil with the separated cornea layer.

The FEM simulations themselves are performed by the commercial tool ANSYS running on a Sun Workstation.

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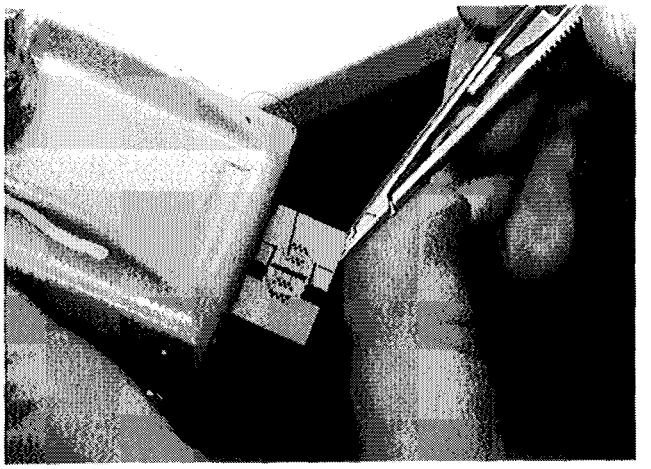

Fig. 6: Metallic foil with stitched cornea lentil

First experimental measurements

First experiments with porcine corneae are realised. Before the sample on the metallic foil will be clamped into the tension test machine and stretched with a constant velocity parallel cuts have been done to get samples with a defined width.

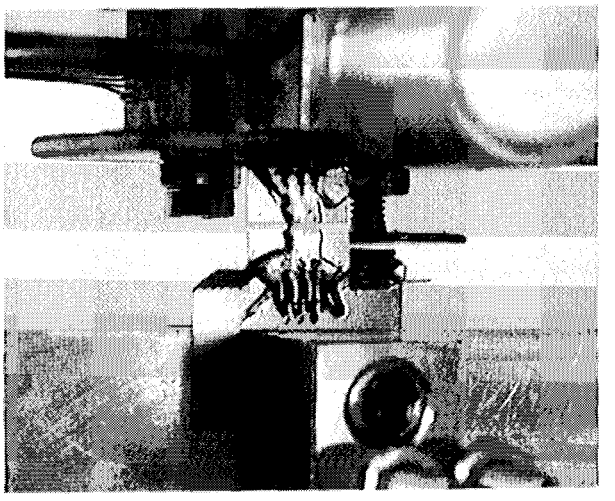

Fig. 7: Clamped sample in the tension test machine

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For first measurements we had used some frozen porcine eyes. After defrosting we dissected them and stitched some cornea lentils on the metallic foils. The samples were clamped into the tension test machine and stretched until they break. Fig. 7 shows such a sample at the beginning of the measurement.

Normally fresh corneae will be used to avoid tissue changes by reason of storage. Fresh corneae means that they are enucleated by the slaughter in the morning and that the dissection and measurement will be done at the same day.

After dissection the stitched samples will be stored in a liquid solution to avoid dehydration. Because clamping and measurement take a few minutes the samples run dry. Using a brush they can be humidified not only before but also during the measurement.

As result of the tension test we get a force-path diagram. Regarding the original length and cross section Young's modulus can be calculated by the use of Hook's law in the range of elastic behaviour. Fig. 8 shows a typical prototypic result of a tension test of a porcine cornea sample.

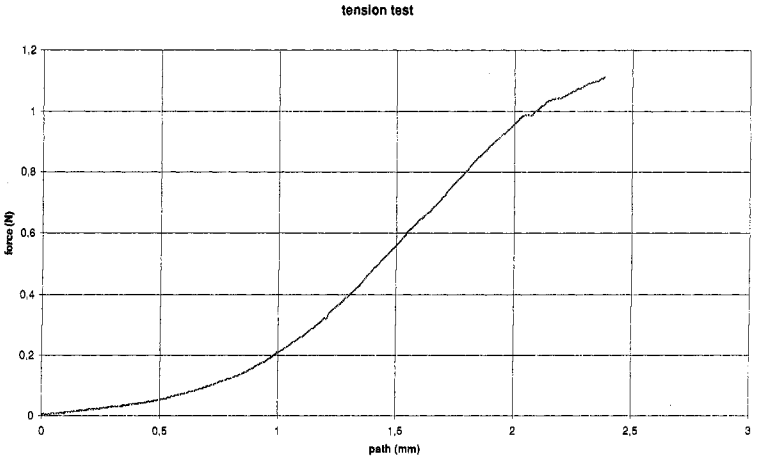

Fig. 8: Prototypic function of a tension test of a porcine cornea lentil

Further measurements will be done with samples dissected out of different corneal depth (different corneal layers). The tension test results will be compared with the anisotropic nature of the cornea to get knowledge about the different material parameters of the layers and the mathematical description of their behaviour. The FEM model consists of multiple layers and the material behaviour will be integrated into the model by attributing the material equations to the different layers.

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Conclusion

The relationship between geometry, material and loaded human cornea data is important for the FEM simulations of the 3-D corneal tissue behaviour. In combination with optical simulations postoperative results concerning visus are expected. Basic results concern the qualitative behaviour of corneal tissue regarding different constraints. For elastic material properties any deformation simulations for different surgical types (PRK, LASIK) could be performed.
The FEM simulations refer to a parameterised solid and net model and regards also an n layered biological tissue. The advantage of simulation and modelling surgical acts are the possibility of reiteration, the parameterisation and the prediction without training at real patients. But the experimental research concerning refinement of material knowledge for the porcine corneae is not finished, and results have to be analysed before experimental treatment of human corneae.

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