Investigation of the biomechanical environment within the optic nerve head by finite element modelling

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Abstract

Aims. To evaluate the role of mechanical stresses in the development of glaucomatous optic neuropathy. Methods. A simplified "generic" model of the optic nerve head was created, consisting of sclera, pre-laminar neural tissue, lamina cribrosa and post-laminar neural tissue. The biomechanical environment was simulated using the finite element method, and results were compared with experimental measurements of pressure-induced deformation of optic nerve head tissues. Results. The computed pattern of pressure-induced deformation of the anterior retinal surface is similar to that measured experimentally. Changes in optic nerve head topography as a function of pressure are similar to differences in topography between parameterized normal and glaucomatous eyes. Conclusions. Qualitative correspondence between results from the simplified model, experimental topographic change and parameterized topographic change are consistent with the assumption that pressure-induced mechanical stress is related to the development of glaucomatous optic neuropathy.

Keywords: Glaucoma; Lamina cribrosa; Optic nerve head; Biomechanics; Human eyes; Finite element modelling

1. Introduction

Glaucoma, a leading cause of blindness, is an ocular disease affecting approximately 50 million people world-wide. Vision loss in glaucoma is due to slow, irreversible, apoptotic loss of retinal ganglion cells, thought to result from a response to insult to their axons as they pass through the lamina cribrosa (LC), a specialized connective tissue at the posterior pole of the eye (Fig. 1). Elevated intraocular pressure (IOP) is associated with the development of glaucomatous optic neuropathy, but the pathogenesis of this damage remains unclear and is controversial. It has been postulated that it involves mechanical stresses on retinal ganglion cell axons, local ischemia or both. Our goal is to evaluate the possible role of mechanical stresses in the development of glaucomatous optic neuropathy, which in turn requires a characterization of the biomechanical environment within the optic nerve head (ONH). In this work we present the results of finite element modelling in which the eye was modelled as a pressure vessel to determine the IOP-related stresses and strains present in load-bearing tissues within and around the ONH. We also compare computational results with experimental measurements of pressure-induced deformation of ONH tissues.

2. Materials and methods

2.1. Finite element modelling

ONH morphology shows considerable variation from person to person. In this initial work we have used a simplified "generic" axisymmetric model incorporating the essential elements of the ONH. It consists of four regions: the sclera/cornea, the pre-laminar neural tissue (including retina), the LC, and the post-laminar neural tissue (Fig. 1). Away from the ONH region the sclera/cornea forms a 1 mm thick spherical shell with 12 mm internal radius. In the ONH region the sclera incorporates an anatomically realistic scleral canal. The pre-laminar neural tissue includes an optic cup and peri-papillary rim in the ONH region. Further from the ONH, the retina has constant thickness before finally tapering away to zero thickness. The LC is modelled as a section of spherical shell 0.3 mm thick. The

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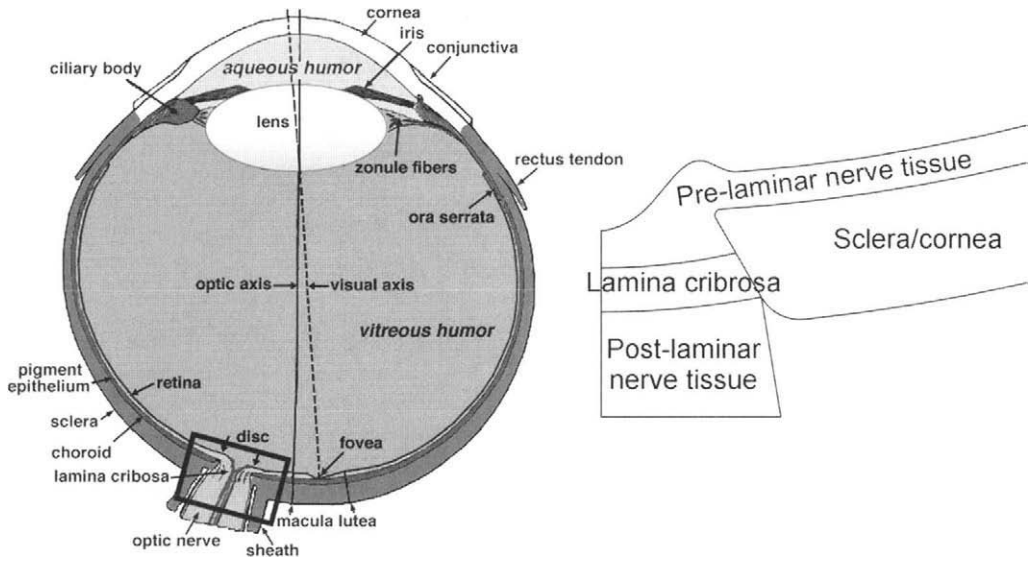

Fig. 1. (Left) Schematic cross-section of the eye with terminology (from http://webvision.med.utah.edu/). (Right) A zoomed-in view of the ONH region of the simplified "generic" axisymmetric model used for modelling, corresponding to the boxed region in the left panel.

post-laminar optic nerve extends posteriorly from the LC and fills most of the scleral canal.

One problem in modelling is that mechanical properties of constituent tissues are incompletely known. For this study the materials were assumed to be linearly elastic, isotropic, and effectively incompressible ($\nu = 0.49$). The sclera was assigned a modulus of 3 MPa (Friberg and Lace [1]). All neural tissue (including retina) was assumed to have stiffness similar to brain, and was thus assigned an elastic modulus of 30 kPa (Miller [2]). LC mechanical properties are unknown; however, the LC is expected to be stiffer than neural tissue but more compliant than sclera. Thus, the LC solid elements were assigned a Young's modulus of 1 MPa. However, the LC is a porous structure with an average solid fraction of 30%, and was thus modelled as a homogenous incompressible material ($\nu = 0.49$) with an effective modulus of 0.3 MPa.

The following boundary conditions were imposed: the nodes on the ONH axis of symmetry were constrained to move radially. The nodes on the axis of symmetry on the cornea were fixed. A fixed IOP was applied to all interior surfaces (anterior retina and sclera/cornea). The model was created using Pro/Engineer 2000i (PTC, Needham, MS, USA), then fine-tuned, meshed and solved using Ansys v5.4 (ANSYS INC, Canonsburg, PA, USA). Two-dimensional, eight-node elements (Plane82) were used for meshing.

2.2. Experimental techniques

Ostensibly normal, fresh, post-mortem, human eye-bank eyes were surgically prepared so that the posterior pole topography could be measured with a scanning laser tomographer (Heidelberg Retinal Tomograph (HRT), Version 2.01a, Heidelberg Engineering, Germany) while controlling IOP at 37°C. (Yan [3])

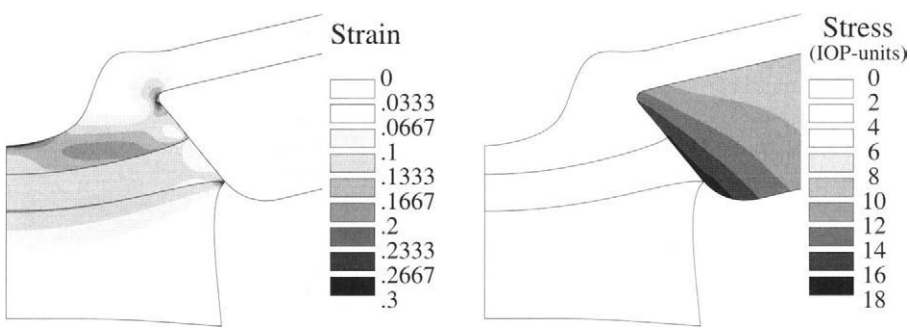

Fig. 2. Computed distributions of von Mises strain (left) and stress (right) in the ONH region for an IOP of 50 mmHg. This region corresponds to right panel of Fig. 1.

3. Results

Fig. 2 shows the computed distributions of von Mises stress and strain in the ONH region for an IOP of 50 mmHg. A region of high strain (maximum strain = 0.4) is visible in the prelaminar neural tissue next to the central anterior border of the sclera. Large stresses (maximum stress = 18 × IOP) are observed in the relatively stiff sclera as compared to the relatively soft retina and optic nerve. When compared to the undeformed model (IOP of 0 mmHg, Fig. 1) a lateral enlargement of the cup is notable.

Comparison of anterior retinal surface topographies at IOPs of 50 and 15 mmHg shows the surprising result that the maximum IOP-induced displacement does not occur in the centre of the optic cup (Fig. 3). Experimentally, a similar pattern is seen in most eyes (Fig. 3), although there is significant inter-eye variability. Generally speaking, the magnitudes of the computed and measured deformations are comparable (Fig. 3).

It is of interest to compare the deformation of our generic ONH region with patient-averaged topographies measured for large populations of normal and glaucomatous eyes (Swindale et al. [4]). These measured topographies are not axisymmetric, and we have therefore used Swindale et al.'s data to extract cross-sectional contours along orthogonal axes passing through the centre of the ONH. The differences between Swindale's normal and glaucomatous topographies (representing "average" ONH topographic changes in glaucoma) are compared with our computed changes in ONH topography in Fig. 4. In spite of the differences in scale it is interesting to note the close similarity in the modelled topography. As previously shown in Fig. 3, maximum deformation does not occur at the centre of the ONH.

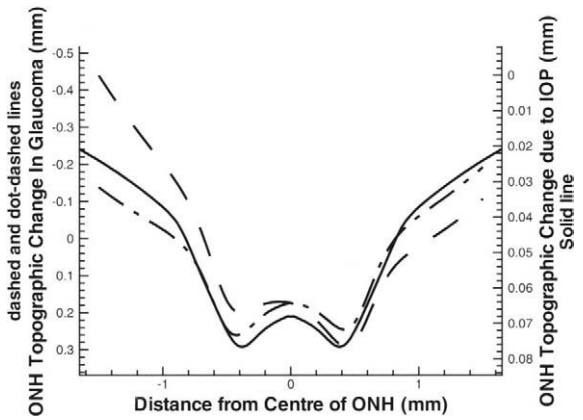

Fig. 4. Plot of changes in ONH topography vs. distance from ONH centre. Solid line: computed topographic change due to IOP increasing from 15 to 50 mmHg [right scale]. Dashed line: experimentally measured topographic change (Swindale et al. [4]) between average normal and average glaucomatous topography along a horizontal line passing through the ONH (left scale). Dot–dashed line: similar to dashed line, but along a vertical line passing through the ONH.

4. Conclusions

Our computed pattern of IOP-induced deformation of the anterior retinal surface is similar to that measured experimentally, with maximal deflection of the prelaminar neural tissues occurring in a ring at the bottom of the optic cup. However, because of significant inter-individual differences in experimental measurements, it is clear that further verification is required.

There was an interesting similarity between FEM-computed changes in topography (induced by changing

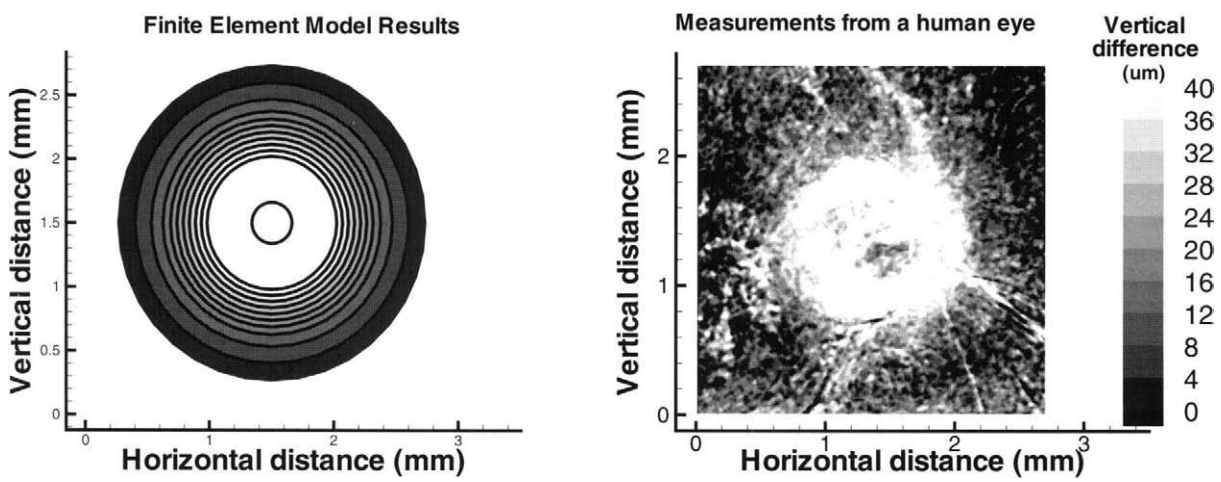

Fig. 3. Topographic difference contour plots computed by finite element model (left) and measured in a single human eye (right). The plotted quantity is the net posterior displacement of the anterior retinal surface between IOPs of 15 and 50 mmHg. Note that maximal displacement does not occur in the centre of the optic cup.

IOP) and differences in topography between parameterized normal and glaucomatous eyes, as reported by Swindale et al. This is consistent with the assumption that changes in ONH topography in glaucomatous eyes are related to mechanical effects of IOP. However, the larger magnitude of the topographic changes seen in glaucoma indicate that secondary remodelling processes must play a large role in determining ONH shape change.

The model presented here is obviously simplified. Ongoing work seeks to incorporate effects of blood vessels, anisotropy and non-linearity in material properties, large deformation effects and patient-specific anatomical features.

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