

# Modeling Fluid Structure-Interactions for Biomechanical Analysis of the Human Eye

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## ABSTRACT

A method is presented for studying and quantifying the biomechanical behavior of the human eye under elevated fluid and structural pressure loading. The approach taken involves a coupled synergistic computer modeling and experimental effort in which numerical models are used to predict the biomechanical response of eye under pressure, and physical cadaveric eye models along with appropriate imaging modalities are used to validate the theoretical predictions of tissue deformation within the eye. The numerical models are based on computational fluid dynamic (CFD) analysis combined with a fully-coupled fluid-structure interaction (FSI) capability. Preliminary numerical modeling results of pressure-induced tissue deformation are demonstrated via application to an idealized eye geometry. A discussion is given of challenges and opportunities of using this approach to advance understanding of biomechanical behavior of the human eye.

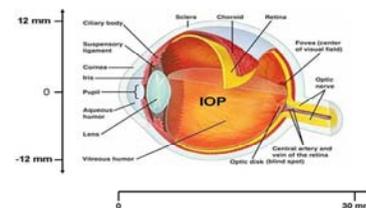
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## 1 INTRODUCTION

The human eye is a highly specialized organ consisting of various tissues with different physical properties and functionality and pressurized fluid-filled compartments in balance with an intricate neurovasculature system. Owing to this complexity, there are many aspects of the biomechanical behavior of the eye that are not fully understood, and yet critical to proper eye function. Among these are the relationships between intraocular (IOP) and intracranial (ICP) pressures and the structurally sensitive constituents such as the ocular globe, retina and optic nerve. These pressures can have significant and detrimental impacts on vision, both short and long term. For example, elevated IOP is known to be a major risk factor for glaucoma, the second leading cause of blindness worldwide. More specifically, an opportunity has arisen that could lead to new breakthroughs regarding pressure

related ocular effects have drawn international attention recently due to newly reported observations of significant ocular abnormalities in astronauts that are subjected to both short- and long-term cumulative exposure to microgravity in space. The anatomical changes and subsequent visual sequelae include choroidal folds, retinal nerve fiber layer thickening, cotton wool spots, optic disc edema, optic nerve protrusion, hyperopic shifts and visual deficits among others [1, 2]. The etiology of these effects is not well understood and is hypothesized to be primarily related to the observed cephaladic fluid shift observed during space flight, but they need to be resolved to enable longer term space exploration.

To date, various mathematical models have been developed in an attempt to elucidate the impact of pressure on the eye and its constituent tissues. However, these models tend to be limited in scope to biomechanical deformation of portions of the eye under specific mechanical loadings. Moreover, the majority of these



(a)



(i)

(ii)

(iii)

(b)

Figure 1. The human eye: (a) interior cross-sectional view showing constituent elements; (b) basic eye geometry, (i) CAD geometry, (ii) 3D STL-based computational geometry, and (iii) sectional view showing computational mesh for structural analysis.

models have been purely structural in nature, i.e. neglecting fluidic analysis. In this presentation we will demonstrate the use of computational fluid dynamics (CFD) combined with fully-coupled fluid-structure interaction (FSI) to study pressure induced tissue deformation (stress and strain) in the eye. We demonstrate the analysis via application to an idealized model of the eye that includes its basic constituents: iris, lens, corneo-sclera shell and a portion of the optic nerve structure as shown in Fig.1a. We used the model to compute the level of stress and strain throughout the eye as shown in Figs. 3 and 4. We discuss the application of the model and the challenges and opportunities of using CFD-FSI to advance understanding of biomechanics of the human eye.

## 2 THEORY AND MODELING

The vast majority of biomechanical models used to study the human eye have been purely structural in nature (i.e. no fluidic analysis) [3-8] Many of these have been developed to study the behavior of the cornea. Similar models have been developed to study IOP-induced stress at the optical nerve head (ONH). Fewer models have been developed to study the entire corneo-scleral shell to analyze its response to IOP or to indentation or applanation tonometry. Very few models have been reported that take into account coupled fluid-structure interactions in the eye as presented here, and these tend to be focused on understanding vibration in the eye. The material properties incorporated into these models varied in complexity, with most being isotropic and a few being anisotropic. Some of the models also included nonlinear tissue behavior as shown in Fig. 3 [6]. However, most researchers have considered small deformations and have made the simplifying assumption that the tissues elastic behavior is linear.

The outer (corneo-sclera) shell of the eye is surrounded by fatty tissue and encased by the bony orbit. There are also six extraocular muscles attached to the eye. The bony orbit, muscles and fatty tissue can be included in the model

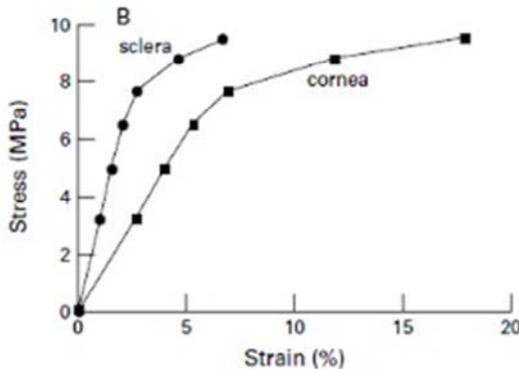


Figure 2. Stress-strain curves ( $\sigma$  vs.  $\epsilon$ ) of the sclera and cornea (Uchio *et al.*, 2009).

explicitly, which adds considerable complexity, or as an

approximation, they can be accounted for in terms of boundary conditions, either via pressure prescribed on the outer boundary of the corneo-sclera shell, or physical constraints at points on its exterior surface.

In our initial work we have studied an idealized model of the eye as shown in Fig. 1. A stereolithography (STL) file was produced from CAD generated eye geometry. This file was then imported into the computational domain of a numerical model. Fig. 1b shows the progression from the CAD geometry to its computational mesh. We have used a CFD-based program FLOW3D ([www.flow3d.com](http://www.flow3d.com)) for the analysis. This program takes into account fully-coupled fluid-structure interaction. Different equation sets and numerical schemes are used for the fluid and stress analysis, respectively. The fluid is modeled using the continuity and Navier-Stokes equations for incompressible Newtonian flow,

$$\nabla \cdot \mathbf{v} = 0 \quad (1)$$

$$\rho \left( \frac{\partial \mathbf{v}}{\partial t} + \mathbf{v} \nabla \mathbf{v} \right) = -\nabla p + \nabla \cdot (\eta \nabla \mathbf{v}) \quad (2)$$

where  $\mathbf{v}$ ,  $p$  and  $\eta$  are the velocity, pressure and viscosity of the fluid. The fluid equations are solved using the Volume-of-Fluid (VOF) method, which is implemented using a finite-differencing numerical scheme that employs a structured finite-difference mesh throughout the fluid regions. Tissue deformation is modeled using the Galerkin finite element method (FEM). The equation governing tissue deformation is

$$\rho \frac{d^2 \mathbf{x}}{dt^2} = \nabla \cdot \boldsymbol{\sigma} + \rho \mathbf{b} \quad (3)$$

where  $\rho$  is the tissue density,  $\mathbf{x}$  is the coordinate of a point in the tissue,  $\boldsymbol{\sigma}$  is the Cauchy stress tensor and  $\mathbf{b}$  is the body force vector. The Cauchy stress tensor is a measure of the stress in the tissue and is computed using

$$\boldsymbol{\sigma}^{n+1} = \boldsymbol{\sigma}^n + \left( K + \frac{2}{3} G \right) tr(\mathbf{E}') + 2G\mathbf{E}' \quad (4)$$

where  $n$  and  $n+1$  refer to successive time-steps and  $K$  and  $G$  are the bulk and shear moduli of the tissue, respectively.

The term  $tr(\mathbf{E}')$  is the trace of the incremental strain tensor,

$$\mathbf{E}' = \frac{1}{2} \left[ (\nabla \mathbf{u})^T + \nabla \mathbf{u} \right] \quad (5)$$

where  $\mathbf{u} = \mathbf{x}^{n+1} - \mathbf{x}^n$  is the incremental displacement and  $\mathbf{x}^n$  and  $\mathbf{x}^{n+1}$  are the positions of a point in the tissue at time steps  $n$ , and  $n+1$ , respectively. To track fluid-structure interaction, the fluid pressure is applied to the boundary faces of the finite element mesh during each time step of the computation. Pressure is extracted for each boundary face of the FE mesh from the fluid mesh by interpolation. The changing conformation of the tissue affects the fluid

flow, i.e. as the tissue surface moves inward or outward, it causes a corresponding inward or outward motion of the tissue layers.

For our initial preliminary analysis, we assume that the interior of the eye is filled with a liquid having the same properties as H<sub>2</sub>O. Standard atmospheric pressure is applied outside the eye. In order to pressurize the interior of the eye in a self-consistent way, fluid is injected into its interior

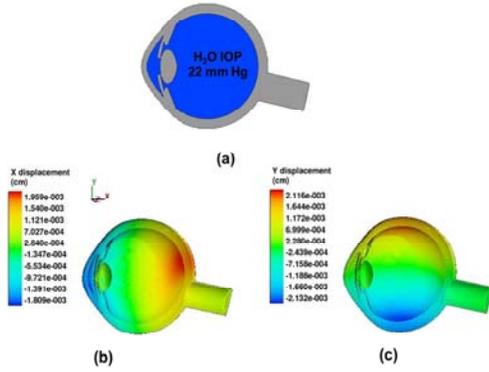


Figure 3. CFD-FSI strain analysis: (a) applied IOP, (b) x-displacement, (c) y-displacement.

compartments until a nominal IOP of 22 mmHg is achieved, as indicated in Fig. 8a. As the incompressible fluid flows into the interior of the eye, the surrounding tissues deforms to accommodate the additional volume of fluid. The principle strain (deformation) profiles in the x and y directions throughout the eye tissue at the target pressure are shown in Figs 3-b,c. The modeling indicates that the maximum x-directed strain occurs along the back edge of the corneo-sclera shell, just above the region at which the optic nerve attaches to the posterior eye. The maximum y-directed strain occurs at the top and bottom edges of the corneo-sclera shell, midway along the ocular cavity. The principle stress in the x-direction  $\sigma_x$  occurs at

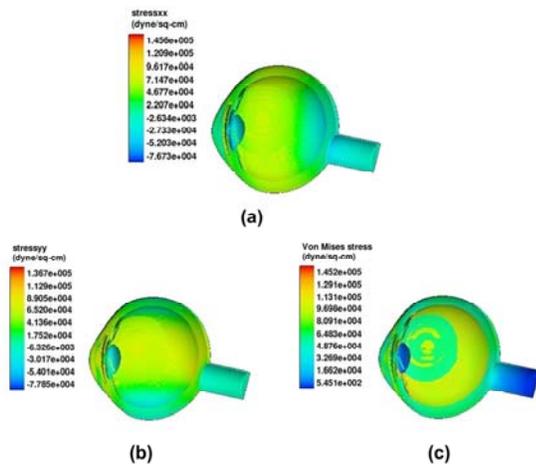


Figure 4. CFD-FSI stress analysis: (a) x-axis stress, (b) y-axis stress, (c) Von Mises stress.

the posterior eye around the region of the optic nerve disk. The principle stress in the y-direction  $\sigma_y$  occurs at the top and bottom edges of the corneo-sclera shell. The Von Mises stress is given by

$$\sigma_{von} = \frac{1}{2} \sqrt{(\sigma_x - \sigma_y)^2 + (\sigma_y - \sigma_z)^2 + (\sigma_z - \sigma_x)^2}, \quad (6)$$

and its maximum occurs at the lens and along the optic nerve structure as shown in Fig. 4c.

There are many challenges in modeling the biomechanics of the eye. Two of these involve the range of length scales and tissue properties that one encounters. For example, different structures of the eye (e.g. tissue thickness, vasculature etc.) can range in size from 10s of microns to millimeters (as in the analysis above). Tissue parameters such as Young's modulus can range over several orders of magnitude as well, from .01 to 10 MPa. This is especially true for the critical posterior portion of the eye where the optic nerve joins the ocular globe.

We plan to develop small-scale models of this region to shed light on the biomechanical behavior of the optic nerve (OH), optic nerve head (ONH), optic nerve sheath (ONS), lamina cribrosa (LC) and surrounding tissues in response to variations in pressure both IOP, and more importantly ICP, which is hydrostatically translated to this

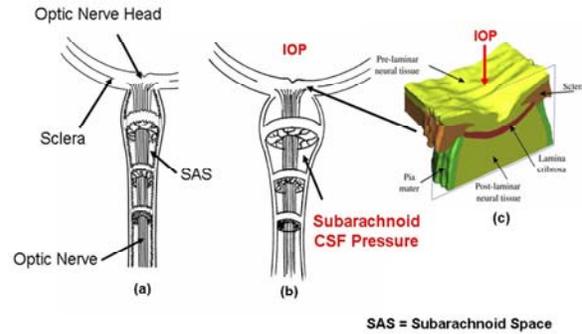


Figure 5. Microscale posterior eye models: Schematic showing the optic nerve, (a) before and (b) after, increased pressure in the subarachnoidal space (SAS), and (c) localized geometry and tissue used for the microscale structural analysis.

region via the subarachnoidal space (SAS) between the ON and ONS. Under normal circumstances, the SAS gap is on the order of 0.1-0.2 mm, whereas the diameter of the optic nerve is on the order of multiple millimeters. However under conditions of elevated ICP, the SAS can become engorged with cerebrospinal fluid (CSF) and balloon substantially. This, in turn, can deform the posterior portion of the ocular shell and induce undesired pressure on the optic nerve and deformation of the posterior orbital layers. These effects can be modeled using the method described above.

In addition to the modeling, we are developing physical models to validate and calibrate the numerical predictions. One such model, shown in Fig. 6, consists of

post-mortem non-fixated human (cadaveric) eyes with isolated human optic nerve preparations obtained from autopsies. This model will reproduce acute changes in ICP and IOP to be correlated with ocular structural changes, which will be evaluated using ultrasound techniques and optical coherence tomography (OCT). For the ICP measurements, two blunt injection cannulae will be inserted into the perineural SAS. The cannulae will be fixated and the SAS sealed using silicone plastic material. One of the cannulae will be connected with a pressure sensor and the other one with a height adjustable water tank. The water

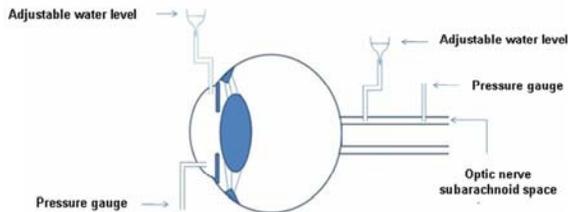


Figure 6. Schematic of experimental eye model. tank will contain isotonic sodium chloride solution and will be height adjustable to apply controlled pressures within the SAS to mimic variations in ICP. The physical models will be used to validate the numerical predictions.

#### 4 CONCLUSIONS

The human eye is a highly complex organ, and much remains to be known about its biomechanical response to elevated intraocular and intracranial pressure (IOP and ICP). The interest in these effects has grown dramatically recently due to new observations of myriad pressure related ocular abnormalities in astronauts exposed to microgravity environments [1-2]. In this presentation we have described a method to advance understanding in this field. Our approach involves a coupled and synergistic modeling and experimental effort in which numerical models are used to predict the biomechanical response of eye under pressure loading, and a physical model combined with appropriate imaging modalities are used to validate the model predictions of tissue deformation within the eye. Specifically, multiscale numerical models involving CFD analysis combined with fully coupled fluid-structure interaction capability are used to predict eye tissue deformation and stress in tandem with cadaveric eye models that closely replicate clinically relevant changes observed over the duration of exposure to the elevated IOP and ICP. We have demonstrated some preliminary CFD-FSI-based calculations of stress and strain in the corneo-sclera shell under nominal IOP loading. The approach presented here could provide a much better understanding of causal effects responsible for the observed changes seen in terrestrial disease states such as Glaucoma and Idiopathic Intracranial Hypertension. The models and experiments may also assist in developing predictive paradigms for

humans working in a microgravity environment for extended periods.

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