Modelling the Deformation of the Human Cornea Produced by a Focussed Air Pulse

Nouran Bahr, Noor Ali, Dipika Patel, Charles McGhee, Peter Hunter, and Harvey Ho

1 Introduction

The cornea is a transparent, avascular structure that forms the anterior part of the eye. The normal mean corneal diameter is 11.7 mm horizontally and the cornea is thinnest centrally, gradually increasing in thickness towards the periphery, with mean values of 0.52 mm and 0.67 mm, respectively, [1]. The cornea is subject to a number of forces, including the internal eye pressure, termed intraocular pressure (IOP), and external atmospheric pressure (Fig. 1a). The mechanical properties of the cornea are predominantly defined by the corneal stroma, which contributes to 90 % of the corneal thickness [2].

Accurate measurement of IOP is crucial in the diagnosis and management of glaucoma, a common, potentially blinding eye disease [3]. The thickness and biomechanical properties of the cornea are known to significantly influence IOP measurement [3]. Many models for the cornea have been proposed (for reviews, see [2, 4]), which recruit various constitutive equations to describe the nonlinear anisotropic mechanical properties of the cornea (for a review, see [2]). Their results are often compared and validated using tonometers (devices for measuring IOP in clinical practice).

Recently a new, non-contact tonometer (Corvis ST, Oculus, Wetzlar, Germany) (Fig. 1) has become commercially available. This device uses an ultra-high-speed Scheimpflug camera (4,330 frames/s) to visualize and measure the corneal deformation response to an air pulse [5], which yields the mechanical properties of cornea

N. Bahr • P. Hunter • H. Ho (✉)
Auckland Bioengineering Institute, University of Auckland, Auckland, New Zealand
e-mail: harvey.ho@auckland.ac.nz

N. Ali • D. Patel • C. McGhee
Faculty of Medical and Health Sciences, Department of Ophthalmology, New Zealand National Eye Centre, University of Auckland, Auckland, New Zealand

B. Doyle et al. (eds.), *Computational Biomechanics for Medicine*,
DOI 10.1007/978-3-319-15503-6_9

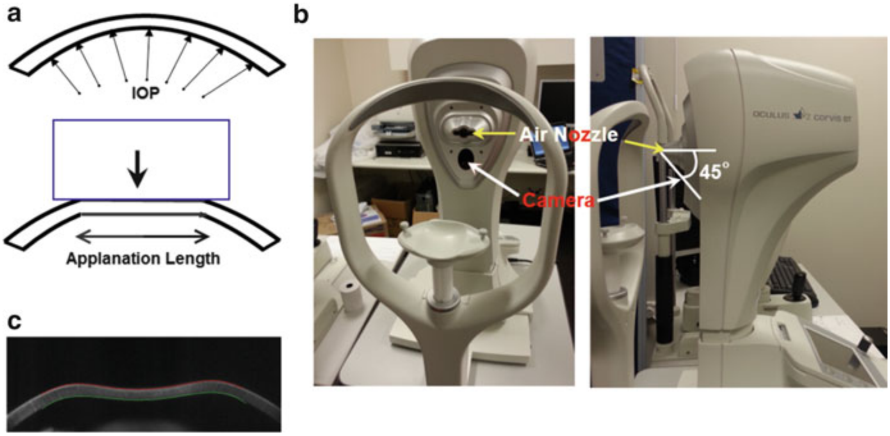

Fig. 1 (**a**) Illustration of the cornea, the IOP and applanation length; (**b**) the Corvis device from two different angles; (**c**) the cornea image it takes

implicitly. This new technique poses challenges to current finite element models (FEMs) which provide alternative means to understand cornea properties, as both the fluid and solid domains need to be solved, with the forces updated across their moving boundaries. The aim of this study was to develop an initial FSI model and to establish some basic parameters for future wide scale studies.

2 Methods

2.1 Medical Imaging

The Corvis ST device was used to measure the corneal deformation in an adult, male volunteer. The fast motion of the cornea under the air puff (duration ∼25 ms) was captured by an ultra-fast camera into a slow motion video. A series of images were selected from the video for further analysis and comparison with our model. The image with the highest concavity is shown in Fig. 1c. Note that the camera was not vertically facing the cornea but placed at a 45° angle towards the cornea [5]. Other relevant data collected included the corneal thickness, IOP, deformation amplitude, applanation length, and corneal velocity.

2.2 Geometry Modelling

The Corvis system was simplified as a working domain consisting of a nozzle, a cornea, and an air space. Their respective geometric modelling is described below.

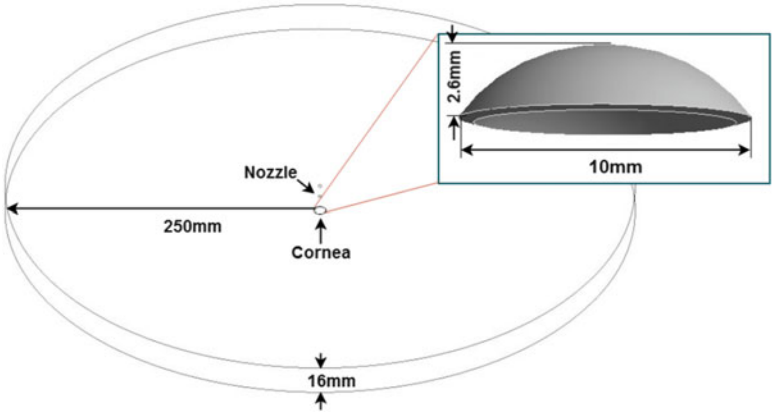

Fig. 2 The entities in the working domain: the open space, the nozzle, and the cornea

2.2.1 Air Space

The air field was designed as an open space consisting of a circular enclosure (500 mm in diameter, 16 mm in height) where the nozzle and the cornea were placed at the centre (Fig. 2). The space was designed to be much larger than the cornea to avoid the influence of rebounding air flow from the wall.

2.2.2 Cornea

Taking the measurements made by the Corvis ST device into account, the cornea was modelled as a dome-shaped elastic solid. We used a constant thickness of 0.52 mm, a horizontal diameter of 10 mm, and a vertical height of 2.6 mm to approximate the cornea in this initial work (Fig. 2).

2.2.3 Air Nozzle

The nozzle was configured as a hollow pipe 3.08 mm in diameter placed 11 mm from the cornea, as per the Corvis manual [5].

Using the grid generator within ANSYS, an adaptive mesh was generated for the air space where the smallest element size was 1 mm. A fine mesh was created at a region under the nozzle and the discharging area. The final fluid mesh contained 4,784,871 tetrahedron elements and 844,235 nodes, the cornea mesh contained 44,409 elements and 70,148 nodes (Fig. 3).

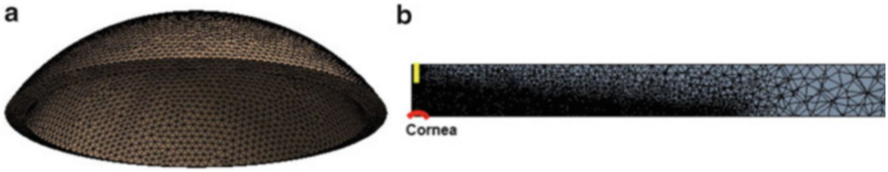

Fig. 3 Computational grid for: (**a**) the cornea; (**b**) the enclosure space (adaptive mesh)

2.3 Biomechanics Modelling

2.3.1 Fluid Solver

With the aim of modelling air flow through a pipe and the resulting effect on the cornea. The ANSYS CFX software was used for air flow simulation. A κ-ω turbulence model was used for flow modelling. The flow was treated as transient and the highest flow velocity from the outlet of the nozzle was set as 80 m/s. Therefore the airflow had a brief and strong pressure impact on the cornea. The wall was treated as outlet of the domain.

The air flow was treated as transient so that the airflow had a brief and strong pressure impact on the cornea. The jet inflow from the nozzle was approximated by a sinusoidal wave (Fig. 4). A κ-ω model was used for air turbulence modelling.

2.3.2 Solid Solver

With different fibre orientations in the corneal centre and periphery, the mechanical properties of the cornea are very complex. In this work we employed a much simpler Neo-Hookean strain energy function:

$$W = \frac{\mu}{2}(\bar{I}_1 - 3) + \frac{1}{d}(J - 1)^2, \tag{1}$$

where μ is the initial shear modulus, d is the incompressibility parameter, $\bar{I}_1$ is the invariant of the Cauchy–Green deformation tensor, J is the determinant of the elastic deformation gradient. The shear modulus (=100 kPa) was computed using the relationship:

$$\mu = \frac{E}{2(1 + \nu)}, \tag{2}$$

where a Young's modulus $E = 225{,}000$ Pa and a Poisson's ratio $\nu = 0.49$ were assumed in all calculations (refer to [2] $E = 0.3$ MPa, $\nu = 0.49$). The ANSYS interface requires the initial shear modulus and the incompressibility parameter to be specified. The remainder are automatically computed and exploited by the software to fit the model to the experimental data.

2.3.3 FSI Framework

The FSI framework within ANSYS is built upon a multi-field analysis (MFX) solver. In brief, within each time step there is a stagger loop whose number of iterations is determined by the convergence of the load transfer between fields or the maximum number of stagger iterations specified [6].

Since the movement of the cornea is insignificant compared with the enclosure size, a one-way fluid–structure analysis was performed, i.e., the transient air force working on the cornea was solved in CFX at first. The computed data are then passed to the structure solver.

2.3.4 Boundary Conditions

The boundary conditions for the fluid and solid solvers are summarized as below:

- A transient outflow profile of 30 m/s was applied at the nozzle (as inflow to the air space);
- The highest flow velocity from the nozzle was set as 80 m/s [5];
- The wall of the enclosure was treated as outlet of the domain;
- A fixed support was configured at the periphery, preventing rotation and translation of the cornea;
- A constant normal pressure of 1 kPa against the cornea wall was also applied to mimic the effect of IOP (normal 1.5 kPa).

Note that the above boundary conditions were set as close to reality as possible. However there were also some simplification treatments, e.g., the fixed periphery condition, to ease the numeric simulations.

3 Results

3.1 Simulation of Air Jet Flow

The transient air flow dynamics was solved within 5.5 h on a desktop computer (Intel Core Quad CPU @ 2.5 GHz) for a simulation of 30 ms. The computer solutions were exported at 20 time steps. Figure 4a shows the jet flow velocity at $t = 15$ ms, where the highest velocity ($\sim$80 m/s) occurred at the air flow centre just below the nozzle (as the boundary condition). After hitting the cornea the air pulse was discharged at a velocity of $\sim$40 m/s. Of particular interest were the pressure forces acting on the cornea, which are shown in Fig. 4b. It can be seen that the highest pressure force acting on the cornea was at its centre at $\sim$2.3 kPa, whereas the pressure was negative at the corneal periphery, due to the IOP (1 kPa) acting from the opposite direction. These raw load data were passed to the solid solver for structural analysis, as described below.

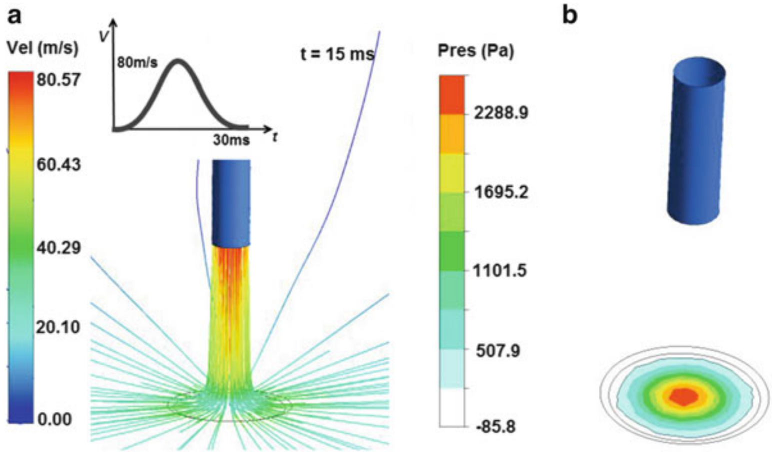

Fig. 4 (**a**) Air flow velocity along the flow streamlines; (**b**) the pressure force acting on the cornea. Note that the pressure at the edge of the cornea (coloured in *white*) was negative due to IOP acting from the opposite direction

3.2 Simulation of Corneal Deformation

Figure 5a provides the in vivo data for the applanation length and the maximum concavity given by the Corvis ST device. In comparison, it can be seen from Fig. 5b that the applanation length yielded from the model was ~2.6 mm, which agreed with that measured by the Corvis ST device (2.5 mm). The curve of the deformation of apex (cornea centre) shows that the largest amplitude was 1.1 mm ($t = 17$ ms) from the Corvis data, whilst our model result suggested a similar maximum amplitude of 0.9 mm ($t = 15$ ms).

A comparison was also made between the image sequence of cornea deflection and that simulated from our model (Fig. 6). The similarity of the concavity of cross section, the thickness of cornea and the applanation length were confirmed in the comparison, yet our model was able to yield more "hidden" information such as the stress and strain that the cornea was subjected to.

4 Discussion

Understanding the biomechanical behaviour of the cornea offers a wide variety of applications including the diagnosis of glaucoma and the detection of eye diseases. Previous works modelled deformation of the cornea subject to a uniformly distributed solid object load, with various material properties proposed. However, a different approach is necessary to estimate the deformation induced by a focussed air pulse. The purpose of this study was to apply an FSI analysis for this problem.

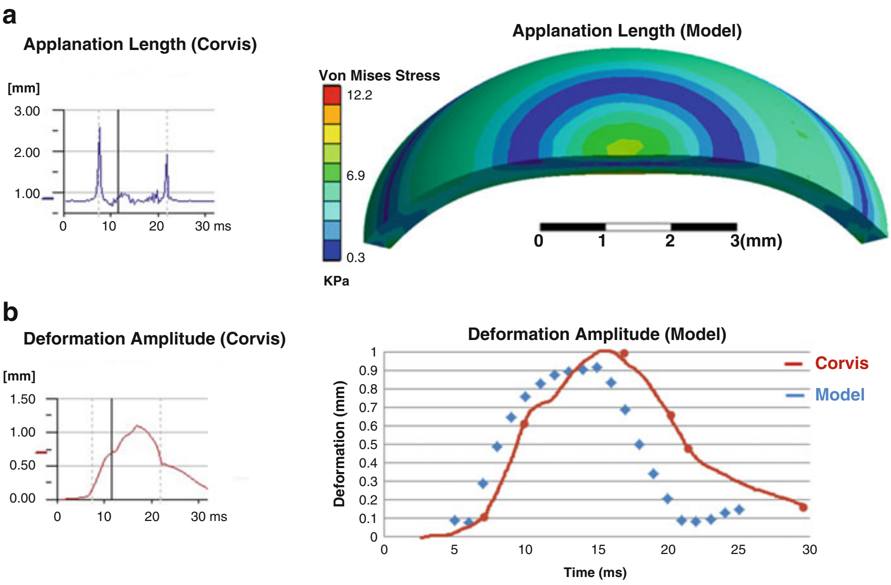

Fig. 5 Comparison between the Corvis measurement and the model: (**a**) applanation length—Corvis: 2.5 mm, model—2.6 mm; (**b**) the deformation amplitude of apex—Corvis: 1.1 mm, model: 0.9 mm

The model was validated against test data obtained in vivo, in particular against the video captured by the device's ultra-fast camera. To our knowledge, this study represents the first FSI model for non-contact corneal deformation.

The complex microstructure of the cornea was simplified to an isotropic material recruiting the nonlinear Neo-Hookean constitutive equation, as focus has not been placed on testing sophisticated material laws at this stage, but rather on establishing the FSI pipeline. Neither did we consider the time-dependent strain for this basic model. Nevertheless the current work could be extended by taking into account important variables, including: varying corneal thickness, non-homogeneity, corneal-scleral connection, and scleral deformation as the cornea deforms.

Further collaboration between ophthalmic clinician-scientists and bioengineers will be beneficial in enhancing the current model. For example, corneal biomechanical properties are known to be significantly altered in keratoconus [7], a non-inflammatory disease in which the cornea assumes a conical shape due to thinning and protrusion resulting in the possibility of corneal transplantation. The diagnosis of keratoconus is usually made on the basis of a combination of clinical and computerized corneal tomographic signs. However, early cases are notoriously difficult to detect. By assessing biomechanical properties in vivo coupled with computer modelling, the Corvis ST is a potentially useful adjunctive tool in the diagnosis of keratoconus.

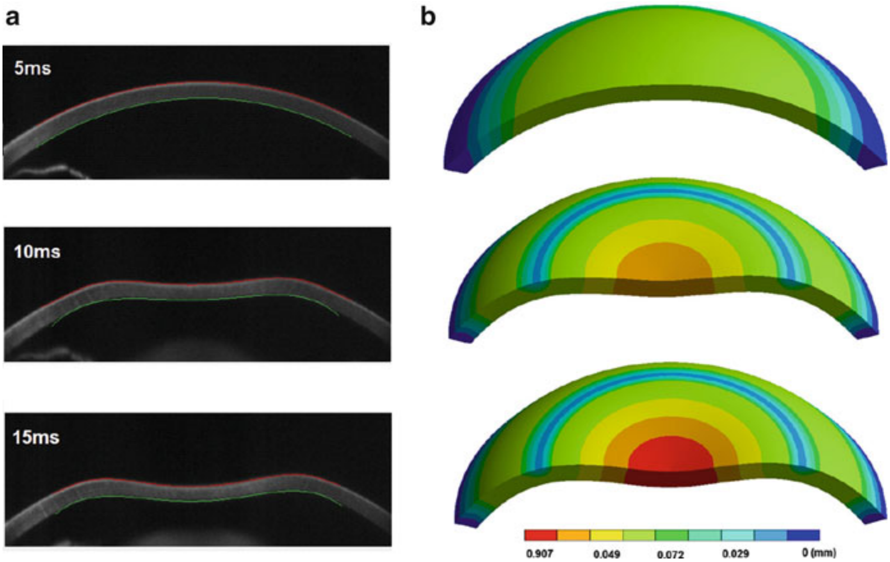

Fig. 6 (**a**) The cross section of the cornea at three time steps (5, 10, and 15 ms, respectively); (**b**) the corresponding corneal deflection simulated from our model

5 Conclusion

This study reports a fluid–structure interaction (FSI) model to describe the deformation of the human cornea under a focussed air pulse. The simulation results were validated against in vivo data derived from the Corvis ST device. The model has a range of potential applications in understanding corneal biomechanical properties in health and disease, and in the diagnosis and management of corneal pathology.

References

1. Bron, A.J., Tripathi, R.C., Tripathi, B.J.: The cornea and sclera. In: Wolff's Anatomy of the Eye and Orbit, 8th edn. Chapman & Hall Medical, London (1997)
2. Pandolfi, A., Manganiello, F.: A model for the human cornea: constitutive formulation and numerical analysis. Biomech. Model. Mechanobiol. **5**(4), 237–246 (2006)
3. Browning, A.C., Bhan, A., Rotchford, A.P., Shah, S., Dua, H.S.: The effect of corneal thickness on intraocular pressure measurement in patients with corneal pathology. Br. J. Ophthalmol. **88**(11), 1395–1399 (2004)
4. Pinsky, P.M., Datye, D.V.: A microstructurally-based finite element model of the incised human cornea. J Biomech. **24**(10), 907–922 (1991)
5. OCULUS: Corvis ST instruction manual (2011)
6. ANSYS: ANSYS CFX-Solver, Release 10.0: Theory. ANSYS Europe Ltd. (2005)
7. Nash, I.S., Greene, P.R., Foster, C.: Comparison of mechanical properties of keratoconus and normal corneas. Exp. Eye Res. **35**(5), 413–424 (1982)