Air Puff Induced Corneal Vibrations: Theoretical Simulations and Clinical Observations

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ABSTRACT

PURPOSE: To investigate air puff induced corneal vibrations and their relationship to the intraocular pressure (IOP), viscoelasticity, mass, and elasticity of the cornea based on theoretical simulations and preliminary clinical observations.

METHODS: To simulate the corneal movement during air puff deformation, a kinematic viscoelastic corneal model was developed involving the factors of corneal mass, damping coefficient, elasticity, and IOP. Different parameter values were taken to investigate how factors would affect the corneal movements. Two clinical ocular instruments, CorVis ST (Oculus Optikgeräte GmbH, Wetzlar, Germany) and the Ocular Response Analyzer (ORA; Reichert, Inc., Buffalo, NY), were employed to observe the corneal dynamical behaviors.

RESULTS: Numerical results showed that during the air puff deformation, there would be vibrations along with the corneal deformation, and the damping viscoelastic response of the cornea had the potential to reduce the vibration amplitude. With consistent IOP the overall vibration amplitude and inward motion depths were smaller with a stiffer cornea.

CONCLUSIONS: A kinematic viscoelastic model of the cornea is presented to illustrate how the vibrations are associated with factors such as corneal mass, viscoelasticity, and IOP. Also, the predicted corneal vibrations during air puff deformation were confirmed by clinical observation.


The understanding of corneal biomechanical properties is important because they are associated with corneal refractive surgeries and diagnosis and monitoring of some ocular diseases. In recent years, greater attention has been paid to the topic of corneal biomechanics.1–4

In vitro experimental studies such as stress–strain tensile tests on corneal strips and inflation tests on the whole cornea have shown the viscoelastic, hyperelastic, and anisotropic features of both animal and human corneas.5–8 Electromagnetic radiation scanning approaches are able to demonstrate how the collagen fibrils are spatially arranged within the stroma tissue, which have a great impact on the corneal biomechanical performance.9 Meanwhile, the corneal biomechanical properties may vary between humans and animals,7 between people with different ages,7 between different ocular diseases,10 and before and after clinical treatments.5 On the other hand, the numerical simulation method is a powerful methodology to study the mechanical behavior of the cornea. For example, the finite element method has been successfully employed to carry out simulations such as keratoconic corneas and response to ocular surgeries,11–20 showing great potential in the prediction of corneal biomechanical and refractive behaviors.
In addition to studies that have focused on the static state of the cornea as mentioned above, understanding corneal biomechanics through a dynamic approach is also important (e.g., with vibration tonometers). In recent years, the Ocular Response Analyzer (ORA; Reichert, Inc., Buffalo, NY), an ocular instrument that can emit an air pulse directed to the corneal surface, has been used to dynamically measure the intraocular pressure (IOP) and investigate the corneal viscoelastic response. Compared with the number of experimental studies, there is less theoretical work in these aspects. Glass et al. developed a simplified model to evaluate the corneal viscosity and elasticity in response to an air puff, but neither the nonlinearity of the corneal elastic properties nor the effect of corneal mass were taken into consideration. Elsheikh et al. used finite element analysis to simulate the corneal response as a function of thickness, curvature, age, and true IOP. Salimi et al. simulated the dynamic response of the eye considering the fluid–structure interaction effect by using the finite element method.

The current study aims at developing a numerical model to investigate the dynamic responses of the cornea under air pulse pressure. This model includes the factors of corneal mass, viscoelasticity, IOP, and the external air pressure. Clinical observations are also presented to support the numerical findings.

**MATERIALS AND METHODS**

**CLINICAL OBSERVATION**

Two ocular instruments, the ORA and the CorVis ST (Oculus Optikgeräte GmbH, Wetzlar, Germany), a new non-contact tonometer, were employed in the current study.

During measurement with the ORA, an air pulse is delivered to the eye, which is in its initial state. Because the air pressure on the cornea experiences an increasing and then decreasing process, the cornea moves inward and then backward to its initial curvature, passing through the application state two times. The two pressures are recorded by an electro-optical collimation system, which can record the light signals reaching peak values at the corneal applanation times when the cornea reaches the applanation position. The whole process lasts approximately 30 ms for each measurement.

The CorVis ST has the same working principle as the ORA in that the corneal response to an air puff is assessed to determine IOP and corneal biomechanical parameters. However, the CorVis ST permits visualization of the response of the cornea to an air impulse via a high-speed Scheimpflug camera capturing images at approximately 4,300 frames per second during the 30 ms air puff, resulting in approximately 140 images per examination.

**Figure 1.** (A) A illustration of the cornea under intraocular pressure (IOP) and air pressure. (B) A kinematical model of the cornea under air puff deformation. Here, m, k, and c represent the equivalent cornea mass, the corneal elasticity coefficient, and the corneal damping coefficient, respectively. The sum of the external force on cornea F(t) equals the force from IOP, F_{IOP}, minus the force from the CorVis ST (Reichert, Inc., Buffalo, NY) air pulse, F_{air}(t). A simplified kinematical differential equation could be written as follows to describe the corneal movement caused by the air pressure. The nonlinear dynamic model is presented in Figure 1.

\[
\begin{align*}
\frac{d^2 y(t)}{dt^2} + c \frac{dy(t)}{dt} + ky(t) &= F(t) = F_{IOP} - F_{air}(t) \\
y(0) &= F_{IOP}/k \\
y(T) &= F_{IOP}/k
\end{align*}
\]

In this model, m is the equivalent corneal mass involved in corneal movement. For a normal human eye, the whole corneal mass is approximately 0.14 g with a dimension of approximately 11.5 x 11.5 mm. According to the instrument instruction, the spatial radius of the air pulse is approximately 1.5 mm, which may imply that not all the corneal mass should be taken into account. Here, we evaluate several equivalent corneal mass values in the computations as m = 0 g, m = 0.00433 g, m = 0.01 g, and m = 0.02 g, which accounts for different ratios of the whole corneal mass.

The coefficient of corneal elasticity, k, was varied between 38 and 50 to test how it affected corneal deformation and vibrations. The damping coefficient, c, reflects the viscoelastic capacity of the corneal tissue. For comparison purpose, it was set at c = 0 and c = 0.0003 to investigate how the damping coefficient affects corneal motion.

F(t) is the sum of the external force on the cornea model, and it can be divided into two parts: the force from IOP, F_{IOP}, and the force from the air pulse, F_{air}(t).

\[
F(t) = F_{IOP} - F_{air}(t)
\]
where $F_{\text{IOP}} = \text{IOP} \times A$ and $F_{\text{air}}(t) = P_{\text{air}}(t) \times A_{\text{air}}$. During calculations, the “A” represents the effective area for the IOP. The IOP in this formula could be replaced by the ORA reading, IOP measurement with Goldmann applanation tonometer ($\text{IOPg}$), and $A_{\text{air}} = \pi r^2$, $r = 1.5$ mm is the effective radius of the air pulse, $P_{\text{air}}(t)$ is the air pressure, which is recorded in the ORA system. It changes with time and can be fitted by the following function:

$$P_{\text{air}}(t) = a_1 \exp\left[-\left(\frac{t - b_1}{c_1}\right)^2\right] + a_2 \exp\left[-\left(\frac{t - b_2}{c_2}\right)^2\right]$$

(5)

where $a_1$, $a_2$, $b_1$, $b_2$, $c_1$, $c_2$ are coefficients to be determined. Although this formula was the expression chosen, any other mathematical expression capable of having a good fit for the air pulse pressure can also be used instead.

The function of time, $y(t)$, represents the cornea position at time $t$. At $t = 0$, the initial state before each measurement, the cornea is assumed to be in the position $y(0) = F_{\text{IOP}}/k$. And at $t = T$, the end of the ORA measuring process, the cornea returns to its initial position, so it is again set at $y(T) = F_{\text{IOP}}/k$, where $T = 30$ ms.

All of the calculation codes were compiled based on the Fortran 90 platform. The governing equation with boundary conditions was solved by finite difference method by splitting the time interval $[0, T]$ into 400 equal parts. The coefficients in (5) were computed using the curve fitting tool box in Matlab R2007a (Mathwork, Inc., Natick, MA).

RESULTS

AIR PULSE PRESSURE

The air pressure profile used for simulations was obtained from the ORA measurement of a 27-year-old healthy man in China. The test was completed in a normal, experimental room temperature environment.

For the measurement of the ORA, IOPg = 13.8 mm Hg, and corneal hysteresis = 11.5 mm Hg. The entire time process lasted 30 ms, and the two applanation points were found at 7.875 and 18.975 ms. The calculated maximum air pressure was approximately 43.5 mm Hg (Figure 2A).

Using the Curve Fitting Toolbox in the Matlab R2007a software, the appropriate parameters were estimated as $a_1 = 5149$, $b_1 = 0.01752$, $c_1 = 0.00441$, $a_2 = 3398$, $b_2 = 0.01167$, and $c_2 = 0.004311$. The comparison between the measured and calculated curves of air pulse pressure versus time is shown in Figure 2B. Note that the pressure was described using the unit of Pa instead of mm Hg used in the clinics, and time used the unit of “second.”

CORNEAL VIBRATIONS

To investigate the corneal dynamic performances, we obtained from the formulas the numerical movements under a simulated IOP of 13.8 mm Hg using a finite difference numerical approach. The calculated results are shown in Figure 3. In these graphs, several curves of corneal movement under different considerations of $m$, $c$, and $k$ were plotted with the horizontal axis of time $t$.

According to the numerical simulations, the shapes of the corneal motion curves, which changed with time $t$, were similar overall to that of $F_{\text{air}}(t)$ in Figure 2. However, there were some differences when corneal mass $m$, elasticity $k$, and damping coefficient $c$ were changed.

In Figure 3A, IOP = 13.8 mm Hg and $k = 43.724$ (Unit: N/m), also showing no damping effect with $c = 0$, the variation caused by different corneal masses was depicted. When $m = 0$, there was no vibration; the cornea only experienced the inward and outward process. At other corneal mass ratios, vibrations of similar amplitude were found. Meanwhile, different vibrating frequencies were observed in different groups. When IOP and $k$ are fixed, whereas $c = 0$, larger mass tends to have a larger vibrating period and lower frequency.

In Figure 3B, the effect of corneal viscoelasticity was tested when $k$ was fixed as 43.724. At $m = c = 0$, there was no induced vibration. At $m = 0.00433$, but $c = 0$, the vibrations occurred but the amplitude remained stable. When $m = 0.00433$ and $c = 0.0003$, both the corneal mass and
Damping effects were taken into account, and although vibrations were observed, the amplitude became smaller with the passage of time. This can imply that the damping ratio has the potential to reduce the vibrating amplitude by absorbing the energy of the air pulse pressure.

In Figure 3C, IOP = 13.8 mm Hg, m = 0.00433, c = 0, and k was changed as 33.724, 38.724, and 43.724. It is observed that lower k leads to larger inward depth and is associated with larger vibrating amplitude.

**Clinical Observation and Comparisons**

The responses of corneal vibrations predicted from our current model were validated from clinical observations of a human cornea measured by the CorVis ST. Sample images through the course of an air puff are shown in Figure 4. These images demonstrate how the cornea behaves under the air pulse stimulus. It is deformed inward and then outward to its initial position, which was in agreement with the numerical investigations.

The time course of a single surface point through an air puff is given in Figure 5A, which supports the existence of corneal vibration behavior predicted by our simulated results. In Figure 5B, the numerical results of IOPg = 13.8 mm Hg, m = 0.00433, c = 0, and k = 43.724 are presented and compared. These parameters were adjusted based on the clinical data. It was found that the two curves in Figure 5 have many similar aspects: (1) the maximum inward distance in Figure 5A was approximately 0.48 mm, the same order as in Figure 5B; (2) the corneal vibrations were observed, which could support the numerical
oscillation phenomenon in Figures 3 and 5B; and (3) between 15 and 20 ms, there are approximately 2.5 vibrating cycles in both curves. However, the difference lies in that the clinical vibrations did not occur throughout the entire course of deformation as predicted by the model, but were isolated to the time period when the cornea was in a concave state, between applanation points.

**DISCUSSION**

During the air pulse pressure from the ORA or CorVis ST instruments, the cornea passes the applanation position twice, once in the inward direction during loading and once in the outward direction during unloading. Our numerical models additionally showed that there would be vibrations along with the corneal movement. Based on this analysis, it was suggested that the corneal vibrations were mainly affected by corneal mass and elasticity under a dynamic external force, and that the damping ability had the potential to reduce the vibrating amplitude. Due to corneal viscoelasticity (damping coefficient c), the amplitude of the vibrations became smaller with time t. In addition, a stiffer cornea results in reduced inward depth and lower vibrating amplitude, as shown in Figure 3C. The numerical results to some extent agree with the observation by CorVis ST that there are vibrations along the corneal inward and outward movements. However, the time course is different, in that the CorVis ST shows the vibrations occurring while the cornea is in a concave state, and not throughout the entire course of the air puff.

As reflected by both the clinical observation in Figure 5A and the numerical results in Figures 3C and 5A, the corneal performance is associated with two aspects: the overall inward-outward movement and the vibrations. The overall movement is mainly caused by the air pressure applied to the cornea. The vibrations may reflect the characteristics of the natural frequency of the cornea. As shown in Figure 5, there are approximately 2.5 vibrating cycles between 15 and 20 ms (5 ms). So, it is estimated that the corneal natural frequency is in the order of approximately \[ \omega = 2\pi / (0.005 / 2.5) = 1,000 \pi \]. However, it should be recognized that the numerical model developed in this study is only a simplified one and, therefore, is not able to distinguish the convex and concave states of corneal shape. Therefore, the model has some limitations that should be discussed.

First, the corneal dynamic model is constructed in one dimension. For the real cornea, three dimensional models are required to simulate the convex and concave states, with their effect on vibrations. However, describing the three-dimensional air pulse, the corneal dynamic responses, and the IOP and aqueous humor changes is a complicated process. The parameters are difficult to determine and have to be estimated; the fluid–structure interaction algorithm is needed and the pre-stress effect within the corneal tissue caused by the IOP should be considered. Meanwhile, a large number of computations is required. Despite these difficulties, this issue is still regarded as valuable and worthy of further investigations.

Second, the corneal mass, m, and the viscoelasticity coefficient, c, were adjusted and changed to test how they would affect the results. They can only show some overall characteristics of the cornea. Third, it is assumed in the expressions of \( P_{\text{IOP}} = \text{IOP} \times A \) and \( P_{\text{air}}(t) = \text{P}_{\text{air}}(t) \times A_{\text{air}} \) that the IOP, A, and \( A_{\text{air}} \) are unchanged during the whole measuring process. Nevertheless, due to the complicated interaction effects between the air pulse and the cornea tissue, and between the cornea tissue and the aqueous humour within as short as 30 ms, it is still unclear whether these parameters remain unchanged. Fourth, comparing the vibrating cycles in Figure 5, it is found that the corneal vibrating frequencies between the numerical and the measured data have a good match. But the parameters are adjusted through trial and error processes. As we understand, this vibrating frequency is also the interacting results of the air, aqueous humour, IOP, and corneal tissue. It should be stated that the current model is unable to give a precise prediction of such a complicated coupled system. A more detailed mathematical analysis is required in the further investigation.

A nonlinear, viscoelastic dynamic corneal model was presented that predicts corneal vibrations during air puff...
deformation, which is a novel prediction with supporting clinical observations. The corneal mass m, the damping coefficient c, the coefficient of elasticity k, intraocular force $F_{\text{IOP}}$ and the air force $F_{\text{air}} (t)$ were involved in our multi-parameter model, where k and $F_{\text{IOP}}$ were related to IOP. According to our results, the corneal vibrations in the deformation were mainly caused by the corneal mass m and elasticity k under a dynamic external force. The damping viscoelastic coefficient c had the potential to reduce the amplitude of the vibrations. Clinical observations by the CorVis ST support the cornea vibrating performance found in our numerical models. Our results would be helpful to understand the viscoelastic and dynamical characteristics of the corneal mechanics during an ORA measurement, as well as with the CorVis ST.

AUTHOR CONTRIBUTIONS

Study concept and design (CT, QR, ZH); data collection (CT, YS, ZH); analysis and interpretation of data (CZ, CJR, DZ, ZH); drafting the manuscript (CT, ZH); critical revision of the manuscript (CZ, CJR, DZ, QR, YS); obtaining funding (CZ, DZ, QR); administrative, technical, or material support (QR)

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