Development of an optical probe to measure the flattened area of ocular cornea

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The optical probe in tonometers is a key component in measuring the flattened diameter or area of the ocular cornea. A new kind of optical probe for the direct measurement of the flattened area of the ocular cornea is presented. The optical probe uses the cone prism with a modulating flake of light intensity as its measuring body. The test results on simulated eyeballs with different radii of curvature of the ocular cornea show that there is a linear relation between the flattened area of the ocular cornea and the normalized current. The optical probe, which is more compact compared with existing optical probes and easily configured with its coaxiality of optical parts, may be an excellent probe for constructing a low-cost, miniaturized applanation tonometer.

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Glaucoma is a group of eye diseases characterized by an elevated intraocular pressure (IOP) that damages and destroys the axons of the optic nerve, leading to complete blindness in the absence of continuous medical treatment[1]. For treating glaucoma, ophthalmologists always attempt to normalize the IOP, thereby reducing the danger of progressive visual field loss. The measurement of IOP with so-called tonometers therefore plays an important role in the diagnosis and management of glaucoma. During the past decades, the corresponding inspection technology of IOP measurement has received a great deal of attention. Thus far, some available technologies for the measurement of IOP can be categorized in principle into the following three categories: indention tonometry, applanation tonometry, and non-contact tonometry (pneumatonometry)[2]. Nowadays, many commercial tonometers, such as Schiötz indentation tonometer, Goldmann applanation tonometer (GAT), non-contact tonometer (NCT), and the most recent dynamic contour tonometer (DCT)[3−5], can achieve effective IOP measurements.

The success of IOP measuring technique is based on the development of an optical probe, which allows some tonometers to measure the flattened diameter or area of the ocular cornea. GAT is regarded as the most clinically accurate tonometer, and its optical probe embodies an applanation prism consisting of two semi-cone prisms[3]. A skilled operator can determine the flattened diameter of the ocular cornea (3.06 mm) by means of a slit lamp microscope upon seeing the upper and lower semi-circles on the flattened stamp, separated and tangential to each other horizontally. Perkins presented a hand-held applanation tonometer that does not require a slit lamp microscope, but employs the same principle as that of GAT[6]. In this portable tonometer, the optical probe uses a prism similar to GAT as the measuring body; however, the illumination of the prism obtained from four miniature bulbs is complicated in structure[6]. In general, the above-mentioned tonometers only serve to measure the flattened diameter of the ocular cornea. These techniques have a few difficulties in terms of obtaining the measurement of the flattened diameter of the ocular cornea because the shape of an astigmatic cornea is similar to a tire; consequently, one obtains an elliptic rather than a circular flattened stamp[7]. A right angle prism acting as the measuring body can directly measure the flattened area of the ocular cornea, but the applanation surface is square[8]. So we develop a new kind of optical probe for direct measurement of the flattened area of the ocular cornea.

Using the total internal reflection principle, the cone prism with a taper is designed at a great scale, as shown in Fig. 1. R_1 and R_2 are the radii of the bottom and top surfaces of the cone prism, respectively, and $R_0 = (R_1{+}R_2)/2$. E_0 is the intensity of incident light. θ_1 and θ_2 are incidence angles of light rays at the inclined flank and the bottom surface of the cone prism, respectively. A beam of parallel rays enters the cone prism and is totally reflected at the inclined flank and delivered to the bottom surface of the cone prism. On this surface, it is again totally reflected and arrives at the opposite inclined flank on the top of the cone prism. It should be mentioned that the bottom surface of the cone prism is also called the applanation surface because it is in contact with the ocular cornea during IOP measurement.

In view of the above, the taper of the cone prism is designed. As an estimate, we suppose that the refractive indices of air, glass, and cornea are 1, 1.5, and 1.376, respectively. Using the total reflection critical angle formula, we know that θ_1 should be greater than

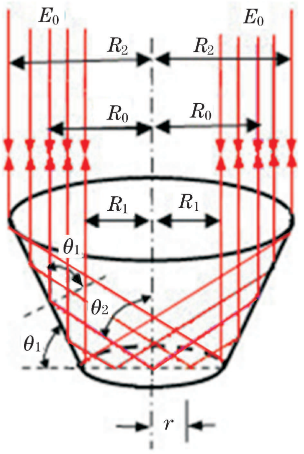

Fig. 1. Schematic diagram of the cone prism and the geometric relationship of optical paths. R_1, R_2, and R_0 are 2.20, 5.20, and 3.70 mm, respectively. The taper of the cone prism θ_1 equals 60°.

41.8° and 41.8°<θ_2<66.5°, where 41.8° and 66.5° are total reflection critical angles of glass versus air and cornea, respectively. In this letter, the taper of the cone prism equals θ_1 and θ_2 =180°−2θ_1.

From Fig. 1, we can obtain the radius of the circumferential ray entering the top surface of the cone prism R by the geometric relationship of rays:

$$R = R_0 - r\cos\theta_2 \quad \left(R_1 \le R \le R_0,\ 0 \le r \le R_1\right) \tag{1}$$

or

$$R = R_0 + r\cos\theta_2 \quad \left(R_0 \le R \le R_2,\ 0 \le r \le R_1\right). \tag{2}$$

The corresponding luminous flux on a circle (with radius r) of the applanation surface is described as

$$\varphi(r) = E_0\pi[(R_0 + r\cos\theta_2)^2 - (R_0 - r\cos\theta_2)^2], \tag{3}$$

where E_0 is the incident light intensity. Then it is reduced to

$$\varphi(r) = 4\pi E_0 R_0 \cos\theta_2 r. \tag{4}$$

Therefore, the light intensity distribution on the applanation surface can be expressed by

$$E(r) = \frac{\mathrm{d}\varphi}{\mathrm{d}s} = \frac{K}{r}, \tag{5}$$

where dφ is the luminous flux on the area element ds = $2\pi r\mathrm{d}r$ of the applanation surface, and the constant $k = 2E_0R_0\cos\theta_2$

The right angle prism is used to measure the flattened area of the ocular cornea based on the fact that the light intensity distribution on the applanation surface is homogeneous. However, from Eq. (5), we know that because of its circular symmetry, the cone prism does not possess the same characteristic as that of the right angle prism. In the current work, a circular modulating flake of light intensity is inserted into the optical path, achieving the homogeneous intensity distribution on the applanation surface of the cone prism, as shown in Fig. 2. The incident light intensity E_0 is properly modulated spatially to $E_0'(R)$.

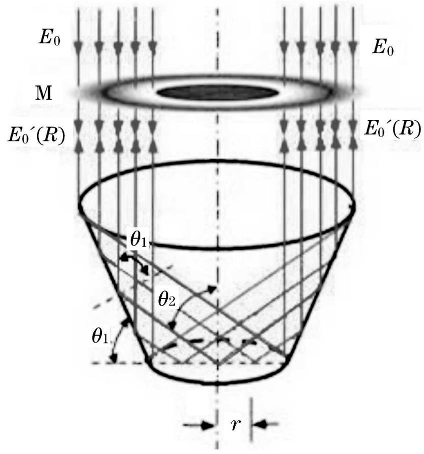

Fig. 2. Modulating flake of light intensity (M) turning E_0 into $E_0'(R)$.

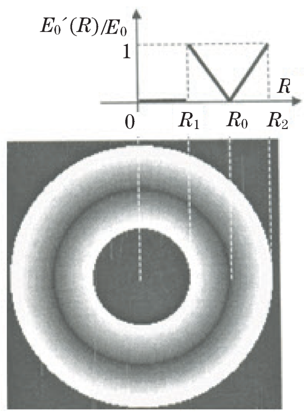

Fig. 3. Modulating flake of light intensity and its transmittivity function curve.

$E_0'(R)/E_0$ is the transmittivity function of the modulating flake of light intensity, and it is defined as

$$\frac{E_0'(R)}{E_0} = -\frac{2}{R_2 - R_1}(R - R_0), \quad R_1 \le R \le R_0; \tag{6}$$

$$\frac{E_0'(R)}{E_0} = -\frac{2}{R_2 - R_1}(R - R_0), \quad R_0 \le R \le R_2; \tag{7}$$

$$\frac{E_0'(R)}{E_0} = 0, \quad 0 \le R \le R_1. \tag{8}$$

Equation (8) means obstructing the light beam that arrives on the bottom surface directly from the top surface of the cone prism, that is to say, the light beam within the range of $0 \le R \le R_1$ is not employed in this study. The transmittivity function curve is shown in Fig. 3.

Referring back to Fig. 2, we again analyze the luminous flux dφ on the area element ds = $2\pi r\mathrm{d}r$ of the applanation surface. It can be written as

$$\mathrm{d}\varphi = 2\frac{R_1 + R_2}{R_2 - R_1}E_0\cos^2\theta_2\mathrm{d}s. \tag{9}$$

Obviously, the light intensity distribution on the applanation surface is a constant, that is,

$$E(r) = \frac{\mathrm{d}\varphi}{\mathrm{d}s} = k' = 2\frac{R_1 + R_2}{R_2 - R_1}E_0\cos^2\theta_2. \tag{10}$$

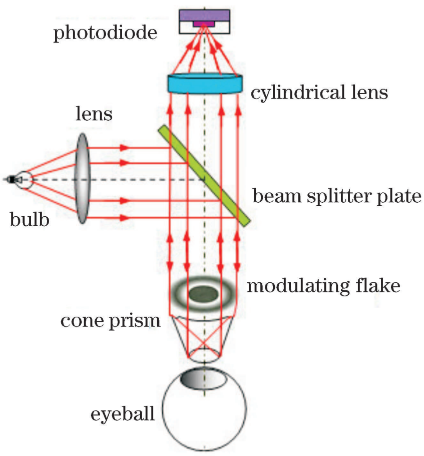

Fig. 4. Optical probe configuration.

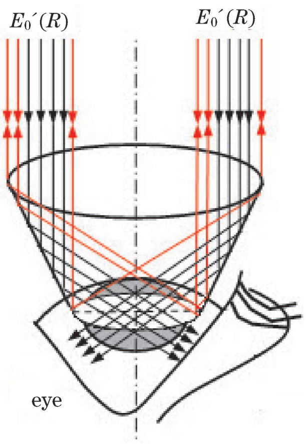

Fig. 5. Cone prism in contact with the eye (not to scale).

This homogeneous distribution of the light intensity on the applanation surface indicates that the cone prism with this modulating flake of light intensity can employ the same principle as that of the right angle prism to act as a measuring body, which makes direct measurement of the flattened area of ocular cornea possible.

In the current research, an optical probe is developed. The simplified configuration is given in Fig. 4. An incandescent light bulb is placed at the focal plane of the converging lens. The light emerging parallel from the lens passes through a beam splitter plate and then reaches the measurement body in which the modulating flake is put close to the top surface of the cone prism. By utilizing a cylindrical lens, the reflected beam returning from the measurement body is imaged onto a photodiode and the photodiode outputs an electrical signal corresponding to the luminous flux returning from the cone prism.

When the cone prism is not in contact with the eye, the luminous flux (φ') returning from the cone prism is equal to the total luminous flux (ϕ) entering the cone prism. However, when the cone prism is in contact with the eye, such as shown in Fig. 5 when it is on the flattened portion of the eye, there is either no reflection or only a weak one because an important part of the light enters into the eye. As shown in Fig. 5, the returning luminous flux φ' is either $\varphi' = \phi - \varphi + R_n\varphi$ or $\varphi' = \phi - R_t\varphi$, where φ is the luminous flux on the flattened portion, R_n and R_t are the reflection and refraction coefficients of the cone prism versus the cornea, respectively, and $R_n + R_t = 1$ (medium absorption is not considered). Thus, the diminished quantity of luminous flux returning from the cone prism can be expressed by

$$\Delta\varphi' = \phi - \varphi' = R_t\varphi. \tag{11}$$

From Eq. (10), we know that the luminous flux (φ) on the flattened portion is directly proportional to the flattened area (s) of the ocular cornea, that is,

$$\varphi = k's. \tag{12}$$

Thus,

$$s = \frac{\Delta\varphi'}{k'R_t}. \tag{13}$$

In this research, the corresponding luminous flux provides a current signal proportional to the flattened area of the ocular cornea. The current will vary according to the flattened area. This is why the optical probe can measure the flattened area of the ocular cornea.

The experiment was carried out on a simulated eyeball with an IOP of 16 mmHg, securely mounted on a straight guideway in which an optical probe was also firmly fixed. The cornea was coated with a film of mineral oil found to be effective in simulating the wettish tear film. A central alignment between the cornea and the cone prism was initially achieved. The simulated eyeball could be driven along the guideway toward the cone prism by a motorized linear displacement actuator such that the flattened area of the ocular cornea can be shaped as shown in Fig. 6. The displacement of the actuator (i.e., applanation height h) was detected by a laser interferometer (HP 5529). The corresponding flattened area can be obtained by the expression $s = 2\pi R'h$, where R' is the radius of curvation of the ocular cornea. Some simulated eyeballs with different radii of curvature of the ocular cornea, such as 6.8, 7.0, 7.4, 7.6, and 7.8 mm, were alternately employed. The data related to the flattened height and area were recorded for later analysis. In this experiment, a spray was used to minimize drying of the cornea. A small water tray was placed beside the guideway to increase the local humidity. All experiments were carried out at room temperature.

In the present experiment, we are mainly interested in the characteristics of the output current with respect to the flattened area of the ocular cornea. A normalized current is defined here as

$$\frac{I - I'}{I} = \frac{\Delta I}{I}, \tag{14}$$

where I is the maximum value of the output current of the photodiode when the cone prism is not in contact with the ocular cornea, I' is the value of the output current when the cone prism is flattening the ocular cornea, and ΔI represents the diminished quantity of the current value. The normalized current is plotted against the flattened area of the ocular cornea, as shown in Fig. 7. The least square linear regression yields an offset of -0.00834 mm^2 with a standard error of 0.01191 mm^2 and a slope of 14.82072 mm^2 with a standard error of 0.02064 mm^2. The statistical results of 70 runs also show that the correlation coefficient is 0.99993 and the standard deviation is

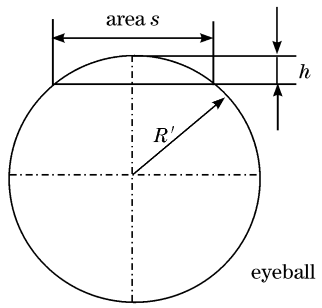

Fig. 6. Schematic diagram of the flattened cornea.

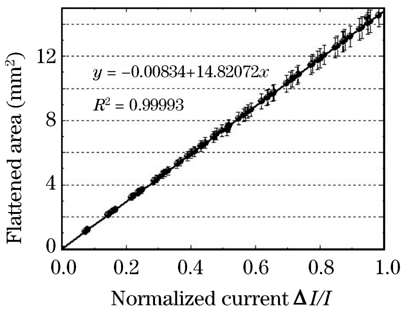

Fig. 7. Flattened area of the ocular cornea versus normalized current (5 different eyeballs measured alternately, $n = 70$). The solid line represents the regression line.

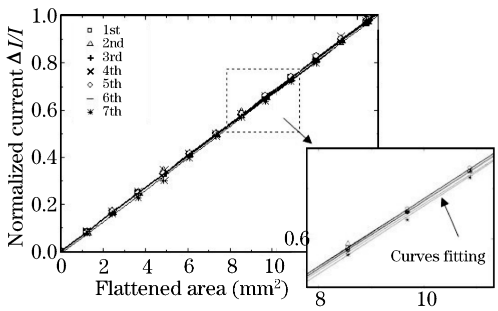

Fig. 8. Reproducibility test of normalized current curve for an eyeball with $R' = 7.8$ mm.

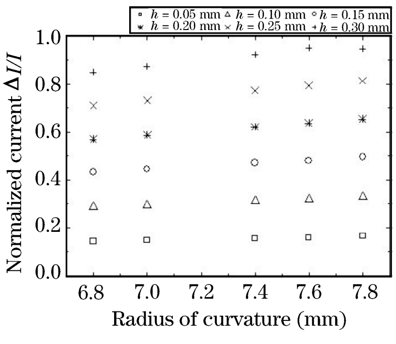

Fig. 9. Effect of the radius of curvature of the ocular cornea on normalized current measurements in the case of the same flattened heights.

0.05182. Such a smaller offset of -0.00834 ± 0.01191 mm^2 attributed to the effect of medium absorption around the cone prism is regarded as the acceptable range. This result indicates that there is a good linear relation between the flattened area of the ocular cornea and the normalized current. In Fig. 7, for example, when the normalized current is reduced by 50%, the corresponding flattened area value of 7.40 mm^2 can be predicted.

The reproducibility of a normalized current curve for a particular eyeball was investigated. Seven consecutive normalized current measurements were taken from each flattened area. Using these readings, the mean and standard deviation were computed. A typical example for the eyeball with $R = 7.8$ mm is given in Fig. 8. The repeatability error of 0.03241 ($\pm3\sigma$) overall is very satisfactory.

The effect of the radius of curvature of the ocular cornea on the normalized current was also investigated. Figure 9 shows that the same flattened heights, owing to the different radii of curvature of the ocular cornea, may produce different normalized currents. Such a phenomenon is reasonable in that the larger the radius of curvature, the larger the flattened area, hence, the larger the normalized current.

Compared with the existing optical probes, the present optical probe is more compact and easily configured with its coaxiality of optical parts. It can successfully achieve the linear measurement of the flattened area of the ocular cornea. Recently, some experimental results associated with IOP monitoring have revealed that there are diurnal variations in IOP, which are risks for glaucoma progression[9]. Owing to the high prices of NCT and DCT, or the difficulty in operating the GAT and Schiötz tonometers, such tonometers can only be employed by ophthalmologists in hospitals. New trends reveal that the IOP should be checked with regularity and relatively often to achieve a better reliable prognosis and management by the clinician, and to facilitate the self-operation of a tonometer (such as the hand-held Perkins tonometer and the portable Tono-pen) by patients at home. The high price and inaccuracy of measurements have, however, limited the spread of their use in medical practice[10,11]. In the future, we will try to develop a low-cost, miniaturized, accurate, and easy-to-use tonometer in which the optical probe is fully utilized.

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