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A new method to calculate corneal ablation depth based on optical individual eye model

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Abstract

The corneal shape, axial lengths and the wavefront aberrations of the eye were all considered to calculate the ablation depth based on the individual eye model. The optimization method for the curves surface replaced the direct calculation from only the optical path difference method (OPD). We analyzed the eye's optical system on its exit pupil and offered the optimum corneal anterior surface. And the ablation depth was the difference between the pre- and post-optimization along the optical axis of the eye. In our experiment, the maximum ablation depth decreases by 8.5% and the mean ablation depth decreases by 8.2% compared with the OPD method.
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Keywords: Ablation depth; Eye model; Wavefront aberrations; Zemax

1. Introduction

Wavefront-guided corneal refractive surgery, intraocular lenses, spectacles and contact lenses are the potential means for correcting monochromatic higher-order aberrations in a clinical environment now [1,2]. And the customized corneal ablation had been studied well and carried into action [3–6].

Scott talked about some potential problems about the customized corneal ablation from the point of view of a doctor [7]. On the optics, the main mission is to calculate accurately the ablation depth along the optical axis of the eye. And it will offer great convenience if the ablation result could be estimated at once. Klein had researched optimal corneal ablation with arbitrary Hartmann–Shack (H–S) aberrations with vectorial geometry ways and he claimed that the wavefront aberrations alone were sufficient to specify the optimal corneal ablation accurately [8]. His calculation was based on ray tracing method and the structure of the Hartman–Shack sensors must be taken into account in his work. And the ablation depth he got was along the ray path from the pre-ablation cornea to the post-ablation cornea, not parallel with the optical axis of the eye.

Liang et al. firstly measured the wavefront aberrations of human eye with the H–S sensor and his experimental setup showed that the measurement position was the entrance pupil plane of the eye. This is equivalent to measuring the wavefront error of the eye at the exit pupil in reference to a perfect reference sphere [9,10]. In fact, the position of the entrance pupil is very different for individual eye because different people have different

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anterior chamber distance and corneal surfaces. So it is difficult to locate accurately the measurement plane and the systematic error will occur.

One solution proposed and actualized now takes the plane passing through the corneal apex and perpendicular to the axis as the measurement plane. The ablation depth is given by the division of the optical path difference (OPD) from the retina to the measured plane by the refraction index difference between the cornea and the air [11,12]. The formula is also based on the ray optics. As a matter of fact, the measurement result on the plane of the corneal apex is not equivalent any more to the wavefront error at the exit pupil in reference to a perfect sphere. The exit pupil of the optical system determines the whole optical quality of the system. The cornea has curved surface but the wavefront aberration data is just measured on the plane in front of cornea. The OPD method just assumes that the two surfaces have the one to one correspondence coordinate value. And this will cause error, especially for the aberrations that the peak to valley value is high.

In this paper, we calculated the ablation depth along the optical axis of the eye and our work was based on the individual eye model constructed with the optical designing software Zemax. The measurement plane of our experiment was put in front of the cornea. The measured wavefront aberrations were input into the model with the phase plate in front of the cornea too. Then surfaces of the crystalline lens were determined by optimizing the model to eliminate the aberrations. The aberrations on the exit pupil could be recalculated if the phase plate was removed. And then the optimization on the anterior corneal surface could give the ablation depth which was the difference between the pre-optimization and post-optimization. That was to say, to get the ablation depth, we optimized the curved corneal surface, which had the aberrations coefficients on the exit pupil as merit function, and solved the difference.

2. Measurement and model of wavefront aberrations

Fig. 1 shows the apparatus we built for measuring the eye's wavefront aberrations. The retina was conjugate with the diaphragm S3 and the receiving surface of the CCD. The lenslet array, which was responsible for sampling the wavefront and focused it onto the CCD, was conjugate with the plane in front of the cornea apex. So, the wavefront aberrations of that plane were measured. Orthogonal Zernike polynomials were used to decompose the wavefront aberrations by Gauss' least-squares method. And the measurement results would be a series of Zernike coefficients.

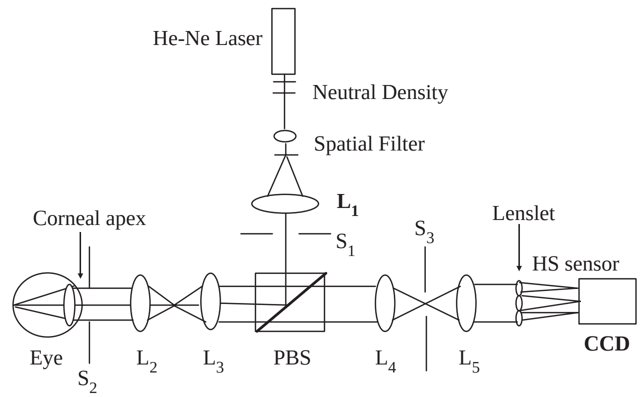

Fig. 1. Hartmann–Shack wavefront sensor for the eye. The measured plane is in front of the corneal which is conjugate with the lenslet of the H–S sensor.

The phase plate can produce required phase corresponding to the aberrations described by Zernike coefficients. The phase added to the wavefront is given by

$$W(x,y) = \sum_{i=1}^{N} C_i Z_i(x,y)\ (x^2+y^2 \leqslant 1), \tag{1}$$

where N is the number of Zernike coefficients in the series, C_i is the coefficient on the ith Zernike fringe polynomial. In Zemax, the Zernike fringe phase surface can be used to model the measured wavefront aberrations. And the Zernike coefficients all have units of waves in this software. So if the aberration coefficients measured were assigned to the corresponding Zernike phase items, the equal wavefront aberrations would be represented well by the phase plate.

3. Individual eye model

Zernike fringe sagittal surface in the software Zemax is defined by the even aspheric surface plus additional aspheric terms defined by the Zernike fringe coefficients. In our research, the even aspheric surface is not used and the surface sag is of the form:

$$z = \frac{x^2+y^2}{R+\sqrt{R^2-(1+c)(x^2+y^2)}} + \sum_{i=1}^{N} A_i Z_i(x,y)\ (x^2+y^2 \leqslant 1), \tag{2}$$

where R and c are the curvature radius and conic constant, respectively. N is the number of Zernike coefficients in the series, A_i is the coefficient on the ith Zernike fringe polynomial. The axis z represents the optical axis of the optical system.

As a kind of high-order aspheric surface, Zernike fringe surface can fit the corneal surface very well. And

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