Influence of Ablation Parameters on Refractive Changes After Phototherapeutic Keratectomy

Harilaos S. Ginis, BSc; Vikentia J. Katsanevaki, MD; Ioannis G. Pallikaris, PhD, MD

ABSTRACT

PURPOSE: The aim of the current study was to control the hypothetical effects of decreased laser energy delivered to the peripheral cornea during phototherapeutic keratectomy (PTK) and provide quantitative calculation of induced low and high order aberrations.

METHODS: We employed a model eye to simulate the refractive effect of homogeneous laser corneal irradiation, as in PTK, for different laser fluences (range 125 to 225 mJ/cm^2) and treatment depths up to 200 μm.

RESULTS: The hyperopic shift induced by the relatively lower energy delivered at the peripheral ablation zone during PTK was proportional to the treatment depth and inversely proportional to the energy fluence. The hyperopic shift calculated using the above ablation parameters was lower compared to previously reported clinical results. Higher order wavefront aberration (total root mean square) changes were of minimal significance for treatment depths up to 200 μm.

CONCLUSIONS: After PTK, a hyperopic shift cannot be attributed to the energy delivery method alone. Modification of laser energy delivery algorithms may only minimize PTK-induced hyperopia. [*J Refract Surg* 2003;19:xxx-xxx]

The 193-nm excimer laser has been used to perform anterior keratectomy on patients with superficial corneal opacities. This procedure has been termed phototherapeutic keratectomy (PTK). The excimer laser can remove precise amounts of superficial corneal tissue and provides an optically smooth surface that is conducive to stable epithelial recovery and basement membrane complex re-formation.

Despite the lack of controlled clinical trials comparing PTK with other treatment modalities, such as superficial keratectomy or lamellar keratoplasty, the US Food and Drug Administration (FDA) identified PTK as an alternative therapeutic approach for the treatment of superficial corneal and epithelial membrane dystrophies, irregular corneal surfaces, and corneal scars and opacities.

The main advantage of the excimer laser lies in its ability to remove deeper corneal opacities that are not readily accessible by conventional superficial keratectomy. Reported complications include induced refractive errors, most commonly hyperopia, corneal scarring, and glare (Summit Technology, Inc. Summary of safety and receptiveness data, excised UV200LA or SVVS Apex [formerly the OmniMed] Excimer Laser System for phototherapeutic keratectomy [PTK]. Waltman, MA: Summit Technology, Inc. 1995).[1] Early studies identified induced hyperopia as the principal disadvantage of PTK.[2-5]

One of the proposed reasons for this complication is the obliquity of incident radiation falling on peripheral cornea, supposed to result in a relative decrease in energy density at the periphery of the ablation zone. According to this assumption, this effect is expected to be more intense for larger irradiation deliveries, ie, deeper ablations.

This is the first study to control the hypothetic effect of diminished energy delivery to the peripheral part of the cornea during PTK and provide numerical data of the expected hyperopic shift, resulting from the oblique angle of incidence of peripheral laser pulses.

Given the known association of laser fluence to the corneal ablation rate[6] we calculated mathematically the photoablation profile of PTK, employing varying fluences and ablation depths on a model eye. We concurrently estimated the expected
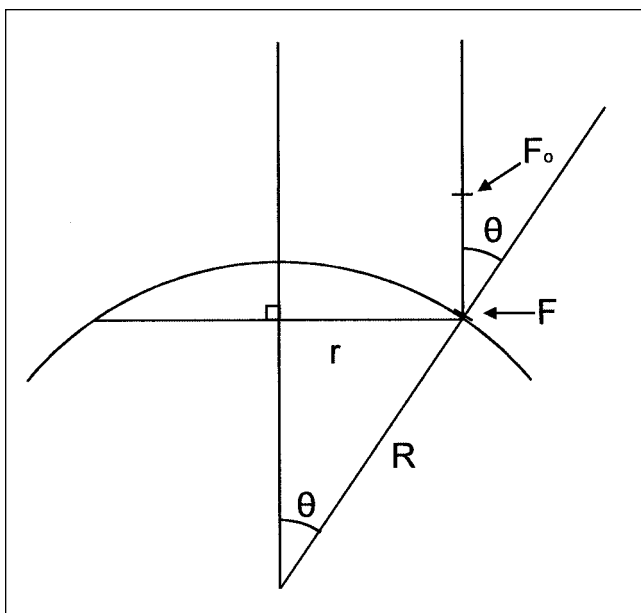
Figure 1. Schematic representation of a laser beam at oblique incidence on the surface of the cornea (see text for notations).

alterations of low and high order aberrations for treatments at an 11-mm ablation zone.

MATERIALS AND METHODS

A recent study by Mrochen and Seiler[7] provided an analytical method for calculation of ablation depth error due to corneal curvature. In this study, the authors took into account the decrease of effective radiant exposure (fluence) due to the inclination of the corneal surface in respect to the angle of incidence of the laser beam at the periphery and the reflection losses at the air-cornea interface according to Fresnel's law. In the current study, we applied this methodology to calculate the ablation profile of uniform laser irradiation of the cornea.

Calculation of Ablation Profile

The angle Θ of the laser beam incidence at a distance r from the apex having radius of curvature R, would be of

$$\sin(\Theta) = r/R \text{ (Fig 1)}$$

given a fluence value at the apex F_o, the fluence $F(r)$ delivered to a point at a radial distance r is described by the equation:

$$F(r) = F_o \cdot \cos(\theta) \Leftrightarrow F(r) = F_o \cdot \sqrt{1 - \sin^2\theta} \Leftrightarrow F(r) = F_o \cdot \sqrt{1 - \frac{r^2}{R^2}} \quad (1)$$

Photoablation of the corneal tissue is a threshold process that requires a minimum radiant exposure for tissue removal. Above this threshold the ablation depth is increased with the radiant exposure in a logarithmic fashion[6,7] as described by the function

$$A = m \cdot \ln(\frac{F}{F_{trh}}) \quad (2)$$

In the above equation, A represents the ablation depth per pulse in microns, F the fluence in mJ/cm², and F_{thr} the fluence threshold required for ablation. The factor m represents the ablation rate at a fluence of $F=e\,F_{thr}$ (F~135 mJ/cm²). Parameters m and F_{thr} were determined experimentally to be of 0.3 μm/pulse and 50 mJ/cm², respectively.[6] Combining the above equations the ablation depth $D(r)$ at any point at a distance r from the apex is determined by the function:

$$D(r) = D_o \cdot m \cdot \ln(\frac{F(r)}{F_{thr}}) \Rightarrow D(r) = D_o \cdot m \cdot \ln(\frac{F_o}{F_{trh}}\sqrt{1 - \frac{r}{R^2}}) \quad (3)$$

where D_o is the ablation depth at the corneal apex (normal incidence of the laser beam).

Moreover, the amount of fluence available for ablation at the corneal periphery is additionally diminished due to increased reflectivity of the cornea at oblique incidence of the leaser beam. Taking this fact into account, equation (3) becomes
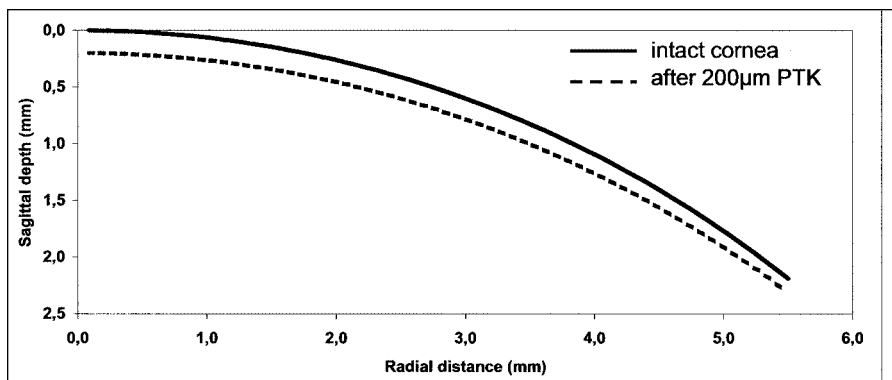
$$D(r) = D_o \cdot m \cdot \ln(\frac{F_o}{F_{thr}}\sqrt{1 - \frac{r}{R^2}} \cdot (1 - refl(r))), \quad (4)$$

where

$$refl(r) = \frac{q_s^2 + q_p^2}{2},$$

is the reflectivity of the air-cornea interface. In the above equation, we calculated the sagittal (q_s) and parallel (q_p) polarization amplitude coefficients of reflection and applied Fresnel's law using a value of 1.52 for the corneal refractive index at 193 nm.[7]

Currently used clinical excimer lasers operate with fluences ranging from 125 to 225 mJ/cm². With the use of equation (4) we calculated the ablation profiles of phototherapeutic keratectomies for this range of fluences and treatment depths of up to 200 μm on an eye model with a corneal curvature of 7.72 mm (Fig 2).

Optical Modeling

Employment of several eye models described in the literature served for simulation of human optical performance. This approach provided the theoretical platform to evaluate efficacy of optical instruments and refractive procedures.[8-10]

In this study we employed a schematic eye originally presented by Sanz and Navarro for modeling the off-axis aberrations of the human eye.[10]

The table summarizes the optical features of this eye model, which incorporates a prolate ellipsoid as anterior corneal surface, a hyperbolic anterior lens surface, and a parabolic posterior lens surface. The posterior corneal surface and retina are modeled as spherical surfaces. The refractive indices of its optical media are defined at four different wavelengths throughout the visible spectrum.

We modeled the anterior corneal surface using the extended cubic spline surface of ZEMAX-EE software (Focus Software Inc, Tucson, AZ). A set of 64 sag points evenly spaced along the radius of a rotationally symmetric shape were calculated according to the prolate ellipsoid formula of the original model. Subtraction of the corresponding ablation depth from each point provided a resulting set of sag points that simulate the postoperative anterior corneal surface. The model's corneal thickness was decreased accordingly to the treatment depth.

The resulting set of sag points was imported to ZEMAX-EE for evaluation. The software automatically calculated the appropriate refractive power of an ideal (aberration-free) thin lens situated at a distance 12 mm from the corneal apex to achieve retinal focus. Focusing was performed on the basis of minimization of root mean square (RMS) wavefront error with respect to the diffraction centroid. The calculated refractive power of this lens provided the measure of the induced refractive shift on the model eye (Fig 3, top).

To evaluate the higher order aberrations induced

Optical Surface	Curvature (mm)	Thickness* (µm)	Semi-diameter (mm)	Conic Constant
Anterior cornea	7.72	0.550	5.5	-0.26
Posterior cornea	6.50	3.05	5	0
Pupil	Infinity	0	5	0
Anterior lens	10.2	4	4.75	-3.13160
Posterior lens	-6	16.3203	4.75	-1
Retina	12		8	0

**Table**
Optical Parameters of the Employed Eye Model

*Distance between consecutive optical surfaces

by the treatment, we used an alternative method for focusing the eye model. In this case the ideal lens was introduced into the vitreous (Fig 3, bottom). With this modality we ensured that both treated and intact anterior corneal surfaces refracted parallel incoming rays. Moreover, the insertion of the lens at this position did not interrupt the known cornea-crystalline lens aberration matching.[11]

RESULTS

Hyperopic shift was nearly inversely proportional to the laser fluence. Figure 4 shows the estimated hyperopic shift per micron of ablation, with laser fluences from 100 to 225 mJ/cm^2. The calculated difference of hyperopic shift within the range of fluences currently used in clinical practice (125 to 225 mJ/cm^2) was less than 0.50 diopters (D) for treatment depths that did not exceed 150 µm, increasing to about 1.00 D after 200-µm treatments.

The hyperopic shift was also related to the treatment depth. The deeper the attempted treatment, the higher the hyperopic shift induced by the decreased laser energy delivered to the outer parts of the treatment zone. Figure 5 shows changes of the
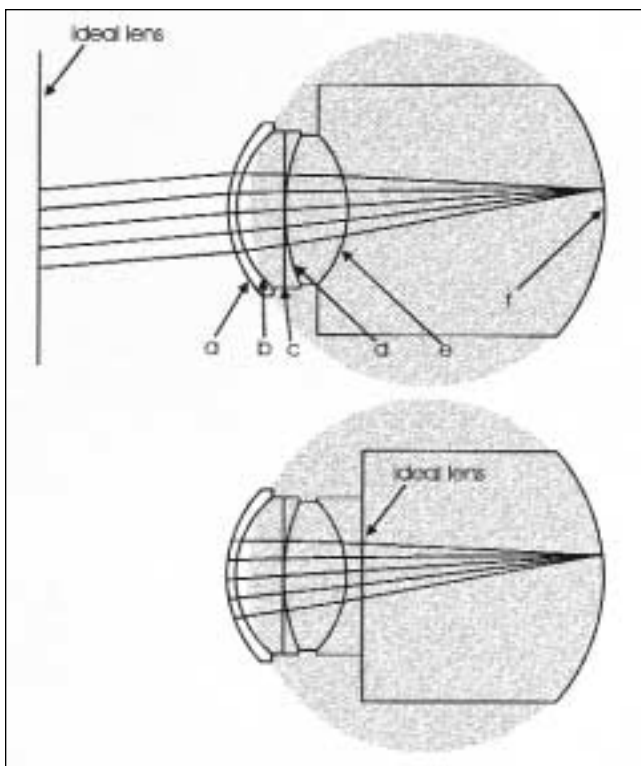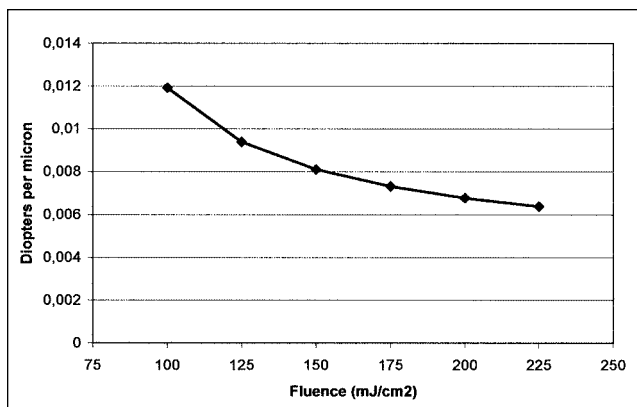
Figure 3. Schematic representation of the eye models utilized in this study (a,b: cornea, c: pupil, d,e: crystalline lens, f: retina). Location of ideal lens for the calculation of: **Top)** Hyperopic shift; **Bottom)** Wavefront aberration and MTF.

Figure 4. Calculated induced hyperopic shift per micron of PTK depth for various laser fluences.

Figure 5. Induced hyperopic shift for various depths of ablation at the apex, calculated for different laser fluences.

calculated hyperopic shift for treatment depths up to 200 µm at different laser fluences.
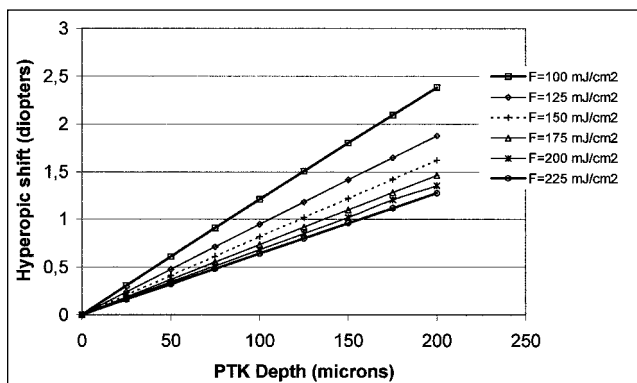
Higher order wavefront aberrations (3rd order and above) were minimally changed upon treatments, representing subtle differences of the wavefront map after the treatment (Fig 6). Similarly, modulation transfer function (MTF) curves were practically unaffected after the treatment (Fig 7).

DISCUSSION

Phototherapeutic keratectomy is an established method for treatment of superficial corneal pathology. One of the principal drawbacks of the technique is the induced postoperative hyperopia due to central corneal flattening that occurs after stroma removal. Although this effect was identified early, it was highly unpredictable among different clinical reports. The individual refractive response is not yet fully understood. Several potential mechanisms that may account for this hyperopic shift include different ablation rates of underlying pathology, centrifugal differential contractive effect of the removal of central corneal lamellae portions, intraoperative shielding by debris toward the edge of treatment,

the increased obliquity of incident radiation toward the peripheral part of the cornea, and corneal healing.[12,13]

In a recent study, Dupps and Roberts[14] reported flattening of donor corneas and thickening of the unablated peripheral stroma following in vitro phototherapeutic keratectomy. They associated their findings to the biomechanical response of the cornea secondary to ablation-induced lamellar relaxation. A complete model for the calculation of acute changes of corneal curvature following PTK could incorporate both the biomechanical factor and the "oblique incidence" effect analyzed in the present study.

Several measures proposed to prevent significant postoperative PTK refractive changes include rotational eye movements during treatment[13], large ablation diameters incorporating transition zones[15,16], shallow treatments[17-19], and the use of masking agents.[20]

Our purpose was to mathematically calculate the isolated effect of the ablation profile and laser fluence on the postoperative refractive result of PTK in
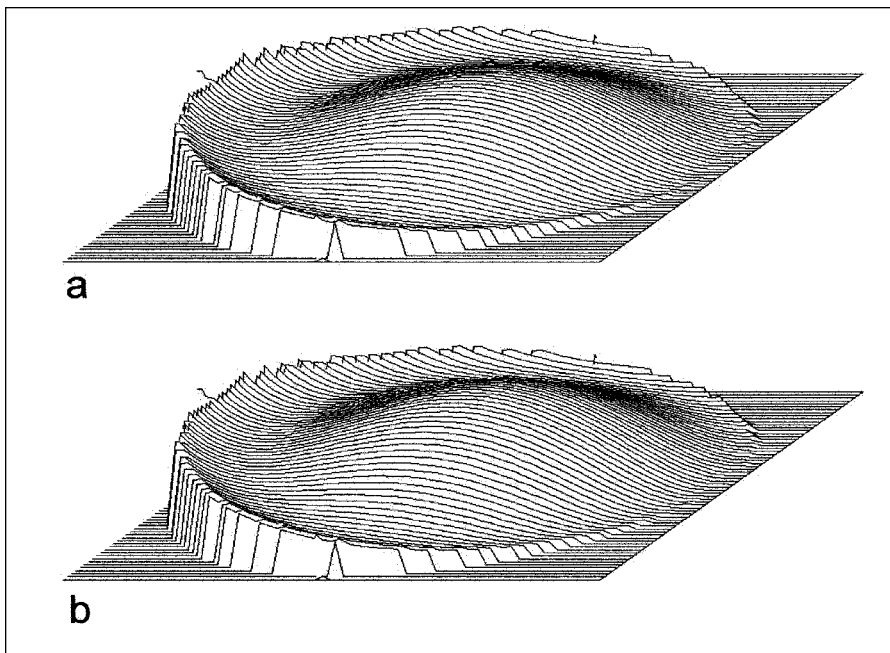
Figure 6. Simulated wavefront aberration maps at the retinal plane. **a)** Before treatment (RMS=1.305243 waves), and **b)** after 200-μm-deep PTK (RMS=1.275031 waves) with laser fluence of 125 mJ/cm². Calculations were performed for a pupil diameter of 5 mm.
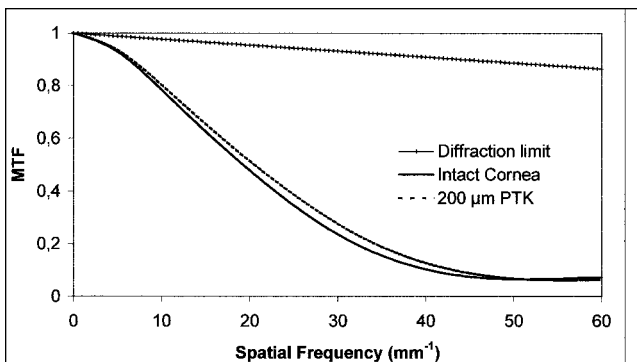
Figure 7. Simulated modulation transfer functions (MTF) of the eye model before and after treatment (D_0=200 μm, F_0= 125 mJ/cm², pupil apperture of 5 mm).

order to provide a rational approach for algorithm design of this modality. We simulated PTK treatment of different depths and laser fluences on an eye model to demonstrate ideal visual performance and calculated the induced wavefront aberrations. The postoperative corneal surface profiles in this study were calculated for an ablation zone diameter of 11 mm. However, refraction and optical performance were calculated by taking into account only the central 5.5 mm of the postoperative corneal surface. It must be noted that according to equation (4), the shape of the central cornea is not dependent on the total diameter of the ablation zone. We utilized an ablation zone of 11 mm (wide in relation to clinical PTK zones) to ensure that there would be no interference of the peripheral unablated cornea with the entrance pupil of our model.

Our results suggest that the postoperative PTK hyperopic shift is nearly inversely proportional to the laser fluence. The higher the fluence the less the hyperopic shift induced by oblique incidence of peripheral laser rays. Most popular laser systems currently operate with fluences ranging from 125 to 225 mJ/cm². According to our calculations, the hyperopic shift within this range of fluences (for a treatment of 150 μm) would be less than 0.50 D, which we consider of minimal clinical significance.

As mentioned before, numerous previous investigators have suggested limitation of the ablation depth as a measure to prevent induced hyperopia after PTK.[17-19] Despite the definite correlation of the intended ablation depth to postoperative refractive status, this relation remains rather unpredictable.

According to our model, the oblique angle of incidence of the laser beam at the peripheral parts of the cornea can substantially modify the refractive status of the eye. Simulation of PTK treatment depths up to 200 μm confirmed previous clinical experience that PTK-induced hyperopia is proportional to the treatment depth. However, compared to clinical results, the estimated hyperopic effect is relatively low. Our results suggest a hyperopic shift of almost 2.50 D after a 200-μm-deep PTK ablation with laser fluence of 125 mJ/cm² (Fig 5). The estimated hyperopic shift is even smaller for higher laser fluences (1.27 D for laser fluence of 225 mJ/cm²). Campos and colleagues[3] reported a mean hyperopic shift of 6.40 D in 56% of 18 eyes treated to depths from 80 to 200 μm. Faschinger

reported a hyperopic shift of +5.00 D after a 25-µm ablation[21]; Ördahl and Fagerholm reported hyperopic shifts up to +6.00 D for intended treatment depths of less than 100 µm[22] and highly unpredictable refractive outcome for treatments deeper than 20 µm. Jain and colleagues reported induced hyperopic shifts of more than +2.00 D in 25% of their series treated with 30-µm-deep stromal PTK for recurrent corneal erosions.[23]

Compared to these reports, our results, although they confirm the implication of the treatment depth on postoperative PTK refractive effect, also indicate that this effect cannot be explained solely by the peripheral under-ablation of the corneal stroma. It seems that other mechanisms play an important role in postoperative refractive results of PTK. Given the unpredictable refractive outcome among the different studies, we conclude that this effect is probably related to the individual healing response of each subject to the treatment. The clinical evidence of favorable results of treatments including transition zones and utilization of masking agents[20] as well as the reported improvement of the postoperative refractive result over time[3] support the assumption that healing response plays a major role in the postoperative refractive status of treated eyes. Under these considerations, proper modifications of PTK nomograms are not expected to eliminate postoperative induced hyperopia. Still, they could serve to minimize this complication.

REFERENCES

1. Maloney RK, Thompson V, Ghiselli G, Durrie D, Waring GO 3rd, O'Connell M. A prospective multicenter trial of excimer laser photherapeutic keratectomy for corneal vision loss. Am J Ophthalmol 1996;122:149-160.
2. Gartry D, Kerr Muir M, Marshall J. Excimer laser treatment of corneal surface pathology a laboratory and clinical study. Br J Ophthalmol 1991;75:258-269.
3. Campos M, Neilson S, Szerenyi K, Garbus JJ, McDonnell PJ. Clinical follow-up of photherapeutic keratectomy for treatment of corneal opacities. Am J Ophthalmol 1993;115:433-440.
4. Fagerholm P, Fitzimmons TD, Orndhal M, Ohman L, Tengroth B. Photherapeutic keratectomy: long term results in 166 eyes. Refract Corneal Surg 1993;9(suppl):S76-S81.
5. Hersh PS, Spinak A, Garrana R, Mayer M. Photherapeutic keratectomy: Strategies and results in 12 eyes. Refract Corneal Surg 1993;9:S90-S95.
6. Seiler T, McDonnell PJ. Excimer laser photorefractive keratectomy. Surv Ophthalmol 1995;40:89-118.
7. Mrochen M, Seiler T. Influence of corneal curvature on calculation of ablation patterns used in photorefractive laser surgery. J Refract Surg 2001;17:S584-S587.
8. Liou HL, Brennan NA. Anatomically accurate, finite model eye for optical modelling. J Opt Soc Am A 1997;14:1684-1695.
9. Thibos LN, Ye M, Zhang X, Bradley A. The chromatic eye: a new reduced-eye model of ocular chromatic aberration in humans. Appl Opt 1992;31:3594-3600.
10. Sanz IE, Navarro R. Off axis aberrations of a wide-angle schematic eye model. J Opt Soc Am A 1999;16:1881-1891.
11. Artal P, Guirao P. Contribution of the cornea and the lens to the aberrations of the human eye. Opt Lett 1998;23:1713-1715.
12. Gartry DS. Early photherapeutic keratectomy studies in the United Kingdom. In: McGhee CNJ, Taylor HR, Gartry DS, Trokel SL, eds. Excimer Lasers in Ophthalmology. London, United Kingdom: Martin Dunitz Ltd; 1997;21:339-356.
13. Helena MC, Talamo JH. PTK complications. In: Azar DT, Steinert RF, Stark WJ, eds. Excimer Laser Photherapeutic Keratectomy. Baltimore, MD: Wiliams & Wilkins; 1997;143-153.
14. Dupps WJ, Roberts S. Effect of acute biomechanical changes on corneal curvature after photokeratectomy. J Refract Surg 2001;17:658-669.
15. Heitzman J, Binder PS, Kassar BS, Nordan ST. The correction of high myopia with the excimer laser. Arch Ophthalmol 1993;111:1627-1634.
16. Gauthier CA, Holden BA, Epstein D, Tengroth B, Fagerholm P, Hamberg-Nystrom H. Role of epithelial hyperplasia in regression following photorefractive keratectomy. Br J Ophthalmol 1996;15:545-548.
17. Starr M, Donnenfeld E, Newton M, Tostanoski J, Muller J, Odrich M. Excimer laser photherapeutic keratectomy. Cornea 1996;15:557-565.
18. Forster W, Atzler U, Ratkay I, Busse H. Therapeutic use of the 193-nm excimer laser in corneal pathologies. Graefes Arch Clin Exp Ophthalmol 1997;235:296-305.
19. Rapuano CJ. Excimer laser photherapeutic keratectomy: long-term results and practical considerations. Cornea 1997;16:151-157.
20. Dogru M, Katakami C, Yamanka A. Refractive changes after excimer laser photherapeutic keratectomy. J Cataract Refract Surg 2001;27:686-692.
21. Faschinger CW. Photherapeutic keratectomy of a corneal scar due to presumed infection after phtorefractive keratectomy. J Cataract Refract Surg 2000;26:296-300.
22. Ördahl MJ, Fagerholm PP. Treatment of corneal dystrophies with photherapeutic keratectomy. J Refract Surg 1998;14:129-135.
23. Jain S, Austin DJ. Photherapeutic keratectomy for treatment of recurrent corneal erosion. J Cataract Refract Surg 1999;25:1610-1614.