Influence of Corneal Curvature on Calculation of Ablation Patterns Used in Photorefractive Laser Surgery

Michael Mrochen, PhD; Theo Seiler, MD, PhD

ABSTRACT

PURPOSE: The aim of this work was to clarify the influence of the effective illumination area and possible reflection losses that occur during laser-tissue interaction on the modeling of profiles for customized corneal ablation, such as wavefront-guided treatments.

METHODS: The changes of the ablation depth per laser pulse due to the projection of a laser spot onto the corneal front surface and reflection losses at the air-tissue interface were calculated.

RESULTS: Moving with a scanning-spot from the center of the cornea toward the limbus resulted in an increase of the effective illumination area and reflection losses, which led to a decrease in the ablation depth per laser pulse. The decrease of the ablation depth was strongly related to the initial radiant exposure and the corneal curvature radius.

CONCLUSIONS: The corneal front surface must be taken into consideration for ablation profile calculations, especially in customized treatments, due to the strong dependence of the ablation depth on the corneal curvature. [*J Refract Surg* 2001;17: S584-S587]

The major advantage of ArF-excimer lasers for photorefractive surgery compared with other techniques for tissue removal is the submicron precision obtained by photoablation. A common method of studying photoablation is to irradiate a sample with a series of laser pulses and then to measure the resultant etch crater depth. To the best of our knowledge, no assumption about changes in etch crater depth (ablation depth) due to different incidence angles of the excimer laser light can been found in the literature. Laser systems used for corneal refractive surgery, however, are designed to illuminate the corneal surface with a laser beam that propagates coaxial or at least parallel through the optical axis of the human eye, as shown in Figure 1. Thus, one might expect a dependence of ablation depth on the position of the laser pulse and a dependence of corneal curvature on the changes in the effective illumination area and an increase in reflection losses. The aim of this work was to clarify these dependencies.

METHODS

Basically, photoablation of corneal tissue with an ArF-excimer laser is a threshold process. The ablation threshold, the minimum radiant exposure (energy per illuminated area) needed for tissue removal, was reported to be approximately 50 mJ/cm^2.[1] Above this threshold, the etch depth per pulse, called ablation depth, increases in a logarithmic fashion with the radiant exposure and might be approximated in first order by the function

$$d = m \cdot \ln\left(\frac{F}{F_{th}}\right), \tag{1}$$

with F the radiant exposure and F_{th} the ablation threshold. The factor m represents the ablation depth at a radiant exposure $F = e^*F_{th}$ ($F \sim 2.7\ F_{th}$) and was found experimentally to be in the order of 0.3 µm per laser pulse.[1] Since currently used clinical excimer lasers work with a mean radiant exposure between 120 and 250 mJ/cm^2 (peak radiant exposure can be more than 400 mJ/cm^2), the resultant ablation depth range from 0.2 to 0.5 µm per pulse.

Refractive Index of the Cornea at 193 nm

The only complex refractive index value for the cornea at 193 nm found in the literature[2] was

From the University of Zurich, Dept. of Ophthalmology, Zurich, Switzerland.

Dr. Mrochen is a scientific consultant of WaveLight Laser Technology AG, Erlangen, Germany.

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Correspondence: Michael Mrochen, PhD, University of Zurich, Dept. of Ophthalmology, Frauenklinik Str. 24, CH - 8091 Zurich, Switzerland. Tel: 41.1.255 4993; Fax: 41.1.255 4472; E-mail: Michael.Mrochen@aug.usz.ch

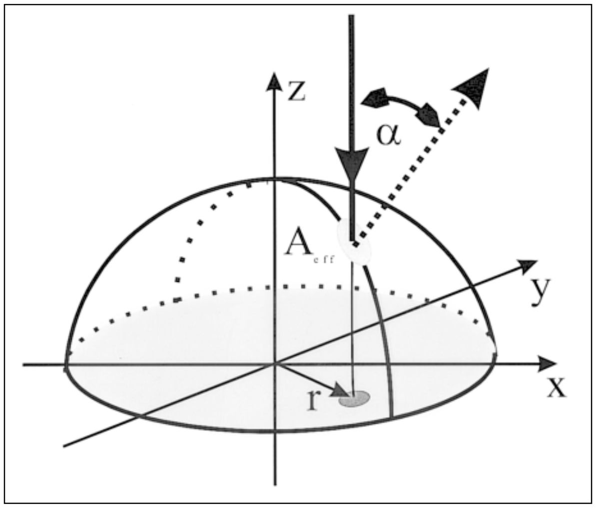

Figure 1. Schematic of light incidence onto the corneal surface. With the effective illuminated corneal area A_{eff}, the radius r in the corneal x-y-treatment plane and the angle of light incidence α.

measured to be $\tilde{n} = n + i\kappa = 1.52 + i \cdot 0.04$. Here, the imaginary component was somewhat smaller compared to the real component ($\kappa^2 << n^2$) and, thus, the imaginary part can be neglected for calculating the reflection coefficient, as determined by the Fresnel's laws.[3] Nevertheless, the refractive index n = 1.52 at 193 nm is significantly beyond the refractive index for visible light, 1.376.[4]

Dependence of Ablation Depth on Effective Illumination Area

As mentioned, laser systems used for refractive surgery are designed to illuminate the corneal surface (curvature radius R) with a laser beam that propagates parallel through the z-axis, as shown in Figure 1. Thus, the effective beam cross section of an illuminated area (A_{eff}) on the cornea increases with an increasing radius r in the x-y treatment plane. The increase of the effective illumination area A_{eff} can be derived mathematically from the projection of a circular laser spot (radius rs) in the x-y plane onto the curved corneal surface, described by the equation

$$z = f(x,y) = f(r) = \sqrt{R^2 - x^2 - y^2} = \sqrt{R^2 - r^2}\ . \tag{2}$$

The asphericity of the corneal curvature is neglected for simplification. According to reference number 5, the effective illumination area A_{eff} of a laser spot (radius r_s) at the position $r^2 = x^2 + y^2$ on a corneal surface with the curvature radius R can be calculated by solving the integrals

$$A_{eff}(r) = \int_{-rs}^{rs} \int_{-\sqrt{rs^2-x^2}+r}^{\sqrt{rs^2-x^2}+r} \sqrt{1 + \left(\frac{d}{dx} f(x,y)\right)^2 + \left(\frac{d}{dy} f(x,y)\right)^2}\, dx\, dy\ . \tag{3}$$

To determine the relative change of the illumination area, this equation can be normalized by the illumination area $A_0 = \pi r_s^2$ of the laser spot in case of a perpendicular incidence of the laser light onto the corneal surface.

$$k1(r) = \frac{A_{eff}(r)}{A_0} = \frac{A_{eff}(r)}{\pi \cdot r_s^2} \tag{4}$$

The illumination area A_0 can be easily derived from equation (3) by assuming x=0, y=0 and a radius of the laser spot r_s smaller as the corneal curvature R ($r_s << R$).

The radiant exposure F = E/A is defined by the quotient between the single laser pulse energy E and the illuminated area A. The increase of the illuminated area, however, leads to a decrease of the applied radiant exposure and, in addition, it changes the shape of the attempted ablation profile.

Dependence of Ablation Depth on the Surface Reflection

According to the differences in the refractive index at the air-tissue boundary presented above, it is obvious that the radiant exposure used for corneal tissue ablation is reduced. As a consequence, one might expect a decrease in the ablation depth in dependence on the radius r in the x-y corneal plane due to a decrease of the absorbed fraction of the initial radiant exposure, according to the Fresnel's laws for reflection.

As the shape of the corneal surface is determined by equation (2), the angle of light incidence a is determined by the behavior

$$\alpha(r) = \frac{\pi}{2} - \arctan\left(\frac{-r}{\sqrt{R^2 - r^2}}\right) \qquad \text{with} \quad 0 \le r^2 < R^2. \tag{5}$$

The relevant mathematical expression for the Fresnel equations were taken from reference number 3 as follows:

$$q_{\perp}(\alpha) = \frac{\sqrt{n^2 - \sin^2(\alpha)} - \cos(\alpha)}{1 - n^2} \tag{6}$$

$$q_{\parallel}(\alpha) = \frac{n^2 \cos(\alpha) - \sqrt{n^2 - \sin^2(\alpha)}}{n^2 \cos(\alpha) + \sqrt{n^2 - \sin^2(\alpha)}} \tag{7}$$

In case of non-polarized light, the reflectance k2(r) at the air-tissue boundary is determined by

$$k2(r) = \frac{q_{\perp}^{2}(r) + q_{\parallel}^{2}(r)}{2}, \tag{8}$$

with $q_{\perp}$ and $q_{\parallel}$ for perpendicular and parallel polarized light, respectively.

Expression (8) results in a decrease of the radiant exposure used for corneal ablation due to optical reflection at the air-tissue boundary. Thus, the relative ablation depth decreases with increasing corneal curvature (increase of the refractive power) and increasing radius at the corneal surface r (ie, increase of the ablation zone).

RESULTS AND DISCUSSION

From the combination of equations (1), (4) and (8) follows:

$$kor(r) = \ln\left(\frac{F}{F_{th}} \cdot \left(\frac{1 - k2(r)}{k1(r)} - 1\right)\right). \tag{9}$$

Where k1(r) stands for the decrease of the radiant exposure according to the changes of the effective illumination area and 1-k2(r) represents the reflection losses.

Expression (9) determines the decrease of the relative ablation depth at the position (r) on the corneal surface (curvature radius R) due to an increase of the effective illumination area k1(r) and an increase in the reflectance k2(r). In addition, there is a strong dependence on the F/Fth quotient due to the logarithmic fashion of the equation (9).

Figure 2 represents the effective ablation depth kor(r) for various corneal curvature radii (R = 6.5 mm; 7.0 mm; 7.83 mm; 8.1 mm) and a radiant exposure of F = 150 mJ/cm². The following basic conclusions can be drawn from expression (9), and the results are presented in Figure 2. First, the effective ablation depth decreases with an increasing radius r at the corneal surface. Second, increasing the radiant exposure leads to a decrease of the ablation depth versus corneal radius dependence. In addition, a decrease of the corneal curvature results in an increase of ablation depth versus corneal radius dependence. As an example, the increase of the effective illumination area as determined by the equation (9) results in a decrease of the effective ablation depth up to 20% at the border of an ablation zone with a radius of 4 mm. In contrast, increasing the radiant exposure F reduces the influence of the corneal curvature on the ablation depth significantly.

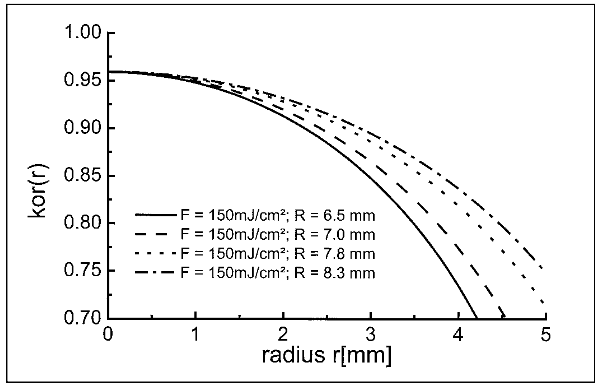

Figure 2. Effective ablation depth kor(r) for a radiant exposure of 150 mJ/cm² and different corneal curvatures.

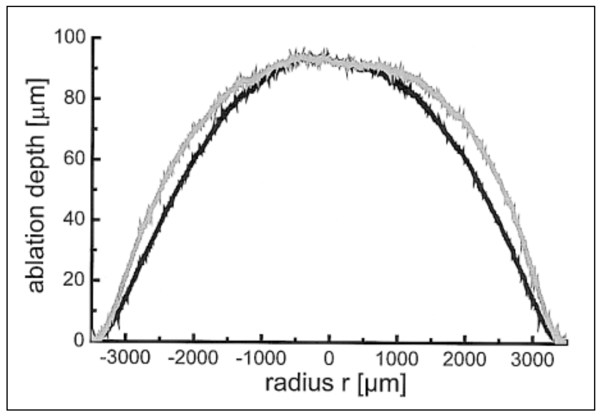

Figure 3. Ablation profile for -6.00 D (optical zone 7.0 mm) with (gray line) and without (black line) taking the reflection losses into consideration. Laser ablation was performed with a modified scanning spot laser (WaveLight Laser Technologie AG, Erlangen, Germany) on fresh enucleated pig eyes taken from the slaughter house. The single spot diameter was 1.0 mm and the repetition rate was 200 Hz. The ablation profiles were measured by means of a surface profiling system.

This result, however, might lead to a positive spherical aberration of higher order after photorefractive laser surgery and might be represented as a central overcorrection and a peripheral undercorrection, as shown in Figure 3. Clinically, this undercorrection was recognized by Seiler and coworkers in 1993.[7] Unfortunately, no assumptions were made about the reasons for the increase of spherical aberration after corneal laser surgery. In the case of a customized ablation treatment, the reduction of the ablation depth affects the planned shape of an individual ablation profile based on wavefront or corneal topography measurements (Fig 4).

The angle of light incident to the corneal curvature increases by moving a laser spot from the center of the cornea toward the limbus. This increases

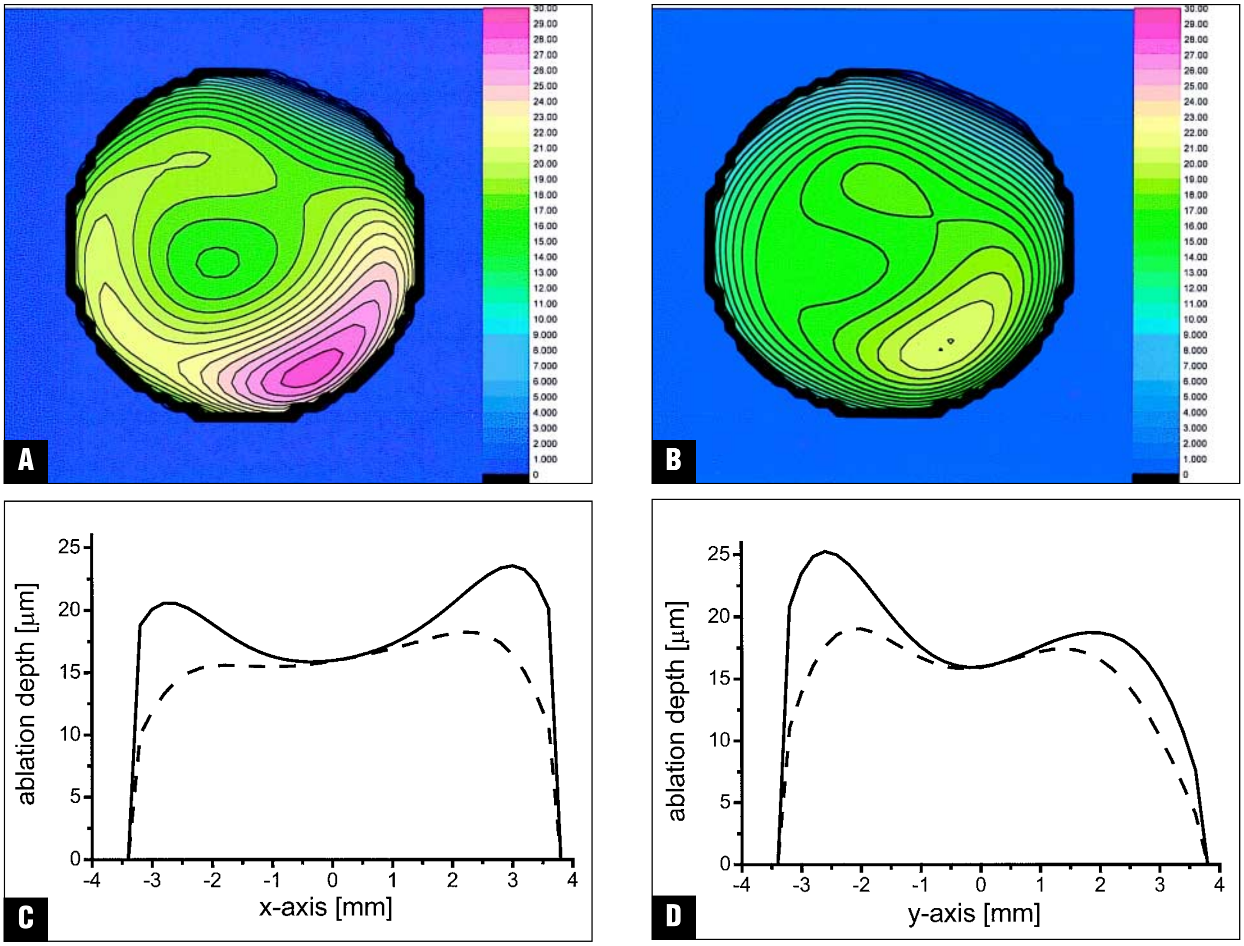

Figure 4. Ablation profile of a wavefront-guided LASIK treatment used clinically for correcting a previously decentered myopic ablation pattern.[6] Taking the losses represented by equation (9) into consideration, the shape of the ablation profile is completely changed (B) compared to the planned profile (A). (C) and (D) are the cross sections of the ablation profiles. The solid lines represent the planned correction (A) and the dashed lines the affected ablation profile (B).

the effective illumination area and the Fresnel reflection losses during photorefractive laser surgery. The decrease of the ablation depth due to these losses is dependent on the corneal curvature at the illuminated position. Thus, the information about corneal front surface must be taken into consideration for the calculation of ablation patterns used in photorefractive laser surgery for vision correction. Nevertheless, further research work is needed to predict other factors that might play a role in ablation profile calculations, such as biomechanics of the cornea, corneal hydration, laser-induced melting processes, epithelial hyperplasia, and the optical role of the flap.

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