Spot Size and Quality of Scanning Laser Correction of Higher Order Wavefront Aberrations

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ABSTRACT

PURPOSE: To investigate the effect of laser spot size on the outcome of aberration correction with scanning laser corneal ablation.

METHODS: Numerical simulation of ablation outcome.

RESULTS: Correction of wavefront aberrations of Zernike modes from second to eighth order were simulated. Gaussian and top-hat beams of 0.6 to 2.0-mm full-width-half-maximum diameters were modeled. The fractional correction and secondary aberration (distortion) were evaluated. Using a distortion/correction ratio of less than 0.5 as a cutoff for adequate performance, we found that a 2 mm or smaller beam is adequate for spherocylindrical correction (Zernike second order), a 1 mm or smaller beam is adequate for correction of up to fourth order Zernike modes, and a 0.6 mm or smaller beam is adequate for correction of up to sixth order Zernike modes.

CONCLUSIONS: Since ocular aberrations above Zernike fourth order are relatively insignificant, current scanning lasers with a beam diameter of 1 mm or less are theoretically capable of eliminating most of the higher order aberrations of the eye. [*J Refract Surg* 2001;17:S588-S591]

One of the exciting frontiers of refractive surgery is customized corneal ablation, which reduces higher order wavefront aberrations of the eye, in addition to eliminating spherocylindrical refractive errors. Reduction of higher order aberrations may allow us to achieve supernormal vision in terms of both acuity and contrast.[1,2] Clinical trials of wavefront-guided corneal ablation were initiated in 1999 by Theo Seiler, MD, using a laser produced by Wavelight, Inc. (Germany) and by Marguerite McDonald, MD, using the LADARVision laser system produced by Alcon Summit Autonomous, Inc. (Orlando, FL).[1] Supernormal visual acuity was reported after customized ablations in a fraction of these eyes. Reduction of higher order aberrations is also useful in restoring vision to a normal level in pathologic conditions such as corneal scar, ectasia, and complicated keratorefractive surgery (decentered ablation, steep central island, and other irregularities).[3-9]

The necessary ingredients of higher order aberration correction include: 1) a method for mapping the aberration through wavefront sensing or corneal topography, 2) a method for generating ablations map from the measurement map (3) a method of laser delivery to produce the planned ablation. This paper addresses the limitations of laser delivery. Laser delivery to the cornea is currently modulated by either beam shaping apertures/masks or beam scanning. The theoretical model we use here applies directly to flying spot scanning lasers that shape the ablation by directing a laser pulse of constant size and fluence profile.

Scanning a small-spot laser beam is the simplest and most flexible method to produce a complex ablation profile. It is obvious that smaller laser spots will be able to produce more accurately ablation patterns with finer depth variation. This study attempts to improve our quantitative understanding of the relationship between laser spot size, beam profile, and aberration correction. The investigation was performed using numerical simulations of ablations with beam diameters and profiles typical of current commercial excimer lasers marketed for laser vision correction.

MATERIALS AND METHODS

Numerical simulations were used to study the outcome of scanning laser ablations to achieve various modes of optical aberration described by Zernike polynomials. The simulations use a digital canvas of 300 x 300 pixels that represents a 12 x 12-mm area.

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Table 1
Beam Characteristics of Current Scanning Spot Excimer Lasers Used in Refractive Surgery*

Laser Name	Minimum Diameter (mm)	Maximum Diameter (mm)	Beam Profile	Peak Fluence (mJ/cm^2)
VISX S3	0.65	6.5	Tophat	160
A.S.A. LADARVision	0.75 FWHM	Same as minimum	Gaussian	400 to 600
Wavelight	0.95 (FW1/*e*)	Same as minimum	Gaussian	400
Schwind	1.0	Same as minimum	Gaussian	
B&L 217	1.0	2.0	Truncated Gaussian	120
Asclepion Meditec	1.8 mm	Same as minimum	Gaussian	250
Nidek EC-5000	<2 mm	10 x 2	Pseudo-Gaussian	250

"Revealing Company Secrets" session chaired by George O. Waring III, MD, at the 2nd International Congress of Wavefront Sensing and Aberration-Free Refractive Correction, Monterey, CA, February 9-10, 2001, supplemented with information from product brochures from laser manufacturers. Diameter definition was not available except for the LADARVision system and was inferred from the peak/average fluence relationship for the Wavelight laser.

The ablation profile was calculated from the beam fluence using a Beer's law approximation of the ablation process.[10] The ablation characteristics of the 193-nm wavelength ArF excimer laser were used. An ablation efficiency of 0.3 µm and a threshold fluence of 60 mJ were adopted from the literature.[11-13]

We simulated corneal ablations to correct aberrations up to the eighth order Zernike mode. Zernike circle polynomials are an orthonormal series used to represent optical aberrations with circular geometry.[14-16] The Zernike series has been adopted as a standard for representing both wavefront aberrations of the eye and corneal topographic aberrations.[17] Because the Zernike series is orthonormal, an aberrated surface can be decomposed into Zernike modes, analyzed mode-by-mode, and then recombined. Each Zernike mode describes a surface and has a radial order n and an azimuthal frequency m. For example, the "defocus" mode is Z_2^0, which describes a parabolic surface and has a radial order of 2 and an azimuthal frequency of 0. The defocus mode is linked directly to the spherical equivalent refraction. Astigmatism is represented by Zernike modes Z_2^2 (cardinal astigmatism) and Z_2^{-2} (oblique astigmatism), which have a radial order of 2 and azimuthal frequencies of 2 and -2, respectively. Higher-order wavefront aberrations can be defined as those described by Zernike third and higher order modes.

For each Zernike mode, we simulated the ablation pattern by convolving the target map with the ablation map for a single pulse. This simulates the smearing effect due to the finite size of the ablation spot. Due to the smearing effect, simulated ablation is not identical to the target Zernike mode correction; the target pattern is only partially corrected. In addition, undesirable distortion is introduced. To evaluate the quality of the correction, we defined a Distortion/Correction Ratio (DCR), which represents the ratio of root mean square (RMS) secondary aberration over the RMS correction inside the optical zone.

RESULTS

We simulated ablations using beam diameters and profiles that are similar to those currently used in commercial excimer lasers for vision correction. These laser systems (Table 1) can be classified as either variable-spot or constant-spot lasers. Although our simulations use constant spot sizes, the results may apply to the smallest spot sizes employed on the variable-spot lasers. The smallest spot size for the lasers range from 0.65 mm to 2.0 mm (Table 1). We conducted simulations of 0.6, 0.9, 1.0, and 2.0-mm-diameter beams. Beam profiles (Table 1) can be either tophat, Gaussian, or in between (truncated Gaussian). To investigate whether the beam profile makes a difference, we simulated both the tophat and Gaussian profiles for the 1-mm-diameter beam.

A complex issue in our simulations involved the several definitions for the diameter of a Gaussian beam. The full-width-half-maximum (FWHM) diameter is measured between the points on a fluence profile where the fluence falls off to one-half of the central maximum. Full-width-1/e (FW1/e) diameter is measured between the points where the fluence falls off to 1/e (e = 2.718 is the base of the natural logarithm) of the maximum. For the Gaussian profile, the FW1/e diameter is 1.2 times larger than the FWHM diameter. The FW1/e^2 definition[18] is also used in wave optics because an optical intensity of 1/e^2 corresponds to an optical amplitude of 1/e. The

FW1/e^2 diameter is 1.7 times larger than the FWHM diameter. Most laser manufacturers do not specify the definition of the beam diameter in their product presentation or brochure and we were unable to get a response from them on this subject. In the following results, we used the FWHM definition. The readers should be aware that the Gaussian beam diameter is significantly larger when the FW1/e or FW1/e^2 diameter definitions are used. For example, the LADARVision beam diameter of 0.75 mm FWHM is essentially the same as the Wavelight beam diameter of 0.95 mm FW1/e (Table 1).

For the Gaussian beam, the peak fluence level affects the diameter and depth profile of the single-shot ablation spot. For current commercial lasers, the peak pulse fluence (Table 1) ranges from 120 mJ/cm^2 for the variable-spot laser with larger beams to 400 to 600 mJ/cm^2 for the smallest beam constant-spot laser. At peak fluence levels of 120, 290, and 500 mJ/cm^2, the corresponding ablation diameters for a 1-mm FWHM Gaussian beam are 1.00, 1.51, and 1.75 mm, respectively. Our simulations used a peak fluence of 290 mJ/cm^2. The reader should be aware that higher fluence lasers produce a larger ablation footprint relative to the beam diameter when interpreting the Gaussian beam results.

We simulated corneal ablations to correct aberrations from Zernike second order modes (sphere and astigmatism) to eighth order modes. The Distortion/Correction Ratio (DCR) was computed for each mode for each beam size and profile. DCR measures how completely the laser beam can correct the target aberration pattern. If the distortion-to-correction ratio is greater than one, then increasing the laser treatment would only lead to a net increase in the ocular wavefront aberration. The lower the DCR is, the more completely one can reduce aberrations. We found that the DCR increases with increasing Zernike order, but less rapidly for smaller beams. If we define a DCR of <0.5 (achieved correction at least twice that of induced aberration) as the criteria of useful correction, then we find that a 2-mm beam is only useful for correction of up to second order Zernike modes (spherocylindrical refractive errors), 0.9 to 1.0-mm beams are useful for correction of up to fourth order modes, and a 0.6-mm beam can be used to correct up to sixth order aberrations (Table 2). There was no significant difference between Gaussian and top-hat beam profiles according to our simulations.

Table 2
Beam Diameter and Correctable Zernike Order

Beam Diameter (mm FWHM*)	Beam Profile	Highest Correctable Zernike Radial Order†
2.0	Flat-top	2
1.0	Flat-top	4
1.0	Gaussian	4
0.9	Gaussian	4
0.6	Gaussian	6

** Full-width half-maximum*
† The highest correctable Zernike radial order for each beam is defined as the highest order for which the Distortion/Correction Ratio (DCR) is less than 0.5 for all azimuthal frequencies associated with the order.

DISCUSSION

To improve quality of vision, it is not necessary to completely remove all of the wavefront aberration from the eye, but merely reduce it. Beyond spherocylindrical error (Zernike second order), most optical aberrations of the eye are contained in the Zernike third and fourth order terms, and nearly all of the remainder resides in fifth to eighth order terms.[2,19] Wavefront measurements in a normal eye have shown that removal of wavefront errors up to the fourth Zernike order is sufficient to achieve diffraction-limited optical performance for a 3.4-mm-diameter pupil.[19] For a 7.3-mm pupil, removal of aberrations up to the eighth Zernike order is sufficient.[19] Adaptive optics correction of most of the eye's monochromatic aberrations up to the fifth Zernike order has allowed imaging of individual photoreceptors[2], suggesting that correction of up to fifth order Zernike terms is sufficient for reaching the neural acuity limit set by photoreceptor spacing. Given that the great majority of higher order aberrations in normal eyes are accounted for by third and fourth order Zernike terms, ≤1-mm-diameter scanning spot lasers are theoretically capable of eliminating at least half of the higher order aberrations, which should significantly increase the quality of vision in terms of both acuity and contrast. Correction of aberrations in some pathologic cases (ie, nodular scarring, steep central island after laser) may require smaller spot sizes.

The simulations presented here have several limitations. The simulations do not account for the microscopic roughness of the ablated surface caused by the imperfect overlap of a finite number of pulses. The Gaussian beam's gradually sloping profile may produce a smoother surface than a top-hat beam. The clinical significance of this is uncertain given the smoothing effect of the epithelium and the laser in situ keratomileusis (LASIK) flap. The simulations did not study the possibility of using transition zones and deconvolution algorithms to improve the ablations results. Our simulations translate directly the target ablation map into the pulse density distribution. Deconvolution algorithms can be used to alter the pulse density distribution both inside and outside the optical zone so that the ablation result more closely follows the intended target. Even with deconvolution, however, the achievable accuracy is still limited by the ablation spot size. The simplicity of our simulations is better suited for an impartial assessment that relates beam size to ablation outcome. The addition of deconvolution and transition zone designs would have involved too many complex and arbitrary choices.

We have not simulated the results of using a variable beam size in an ablation. Such a simulation would be meaningless without specific knowledge of how the beam size is varied. If the ablation characteristics of the various beam sizes are well characterized, a well-planned variable-beam ablation would probably produce results similar to that of using the smallest beam in the variable range. Variable-beam technology allows the use of larger beams for more rapid ablation and thus reduces the variability due to stromal drying. However, systems that utilize broad beams (5 to 6.5 mm) must contend with the central island effect[20-22], which may be difficult to compensate for accurately.

Our study shows that, based on fundamental considerations, the 1 mm and smaller beam sizes utilized on current scanning laser systems should be able to eliminate most higher order optical aberrations from the eye. Thus, we have the laser technology to produce supernormal vision. Apart from laser beam issues, however, there are other difficulties for wavefront-guided ablation. Eye tracking becomes more critical for the smaller laser beams required for higher order aberration correction. Consistent achievement of supernormal vision will also be limited by our ability to predict and compensate for the eye's biological response to laser surgery, such as the draping effect of the lamellar flap, epithelial remodeling, and corneal hydration and structural changes.

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