

Effect of Beam Size on the Expected Benefit of Customized Laser Refractive Surgery

Antonio Guirao, PhD; David R. Williams, PhD; Scott M. MacRae, MD

ABSTRACT

PURPOSE: Customized laser surgery attempts to correct higher order aberrations, as well as defocus and astigmatism. The success of such a procedure depends on using a laser beam that is small enough to produce fine ablation profiles needed to correct higher order aberrations.

METHODS: Wave aberrations were obtained from a population of 109 normal eyes and 4 keratoconic eyes using a Shack-Hartmann wavefront sensor. We considered a theoretical customized ablation in each eye, performed with beams of 0.5, 1.0, 1.5, and 2.0 mm in diameter. We then calculated the residual aberrations remaining in the eye for the different beam sizes. Retinal image quality was estimated by means of the modulation transfer function (MTF), computed from the residual aberrations. Fourier analysis was used to study spatial filtering of each beam size.

RESULTS: The laser beam acts like a spatial filter, smoothing the finest features in the ablation profile. The quality of the correction declines steadily when the beam size increases. A beam of 2 mm is capable of correcting defocus and astigmatism. Beam diameters of 1 mm or less may effectively correct aberrations up to fifth order.

CONCLUSION: Large diameter laser beams decrease the ability to correct higher order aberrations. A top-hat laser beam of 1 mm (Gaussian with FWHM of 0.76 mm) is small enough to produce a customized ablation for typical human eyes. [*J Refract Surg* 2003;19:15-23]

Customized corneal ablation may improve outcomes of laser refractive surgery.^{1,2} The eye suffers from defects—other than defocus and astigmatism—called higher order aberrations.^{3,4}

Theoretical estimates as well as experimental findings⁵⁻⁸ have shown that correction of these higher order aberrations may improve visual performance compared with a conventional correction. Ocular aberrations can now be measured rapidly and accurately. Thus, data from wavefront sensors may be used to guide customized ablations. Instead of subtracting a rotationally symmetric profile of tissue, as in the case of defocus correction, or removing a cylindrical profile for astigmatism correction, customized correction requires more complex ablation profiles, which correspond to the pattern of higher order aberrations to be corrected. Customized correction places higher demands on the accuracy with which the cornea is sculpted. Just as an artist requires a finer paintbrush for more detailed work, so the laser refractive surgeon requires a smaller laser beam to achieve a successful customized ablation. Using aberration data in real eyes, we studied the feasibility of customized refractive surgery with laser beam diameters ranging from 0.5 to 2.0 mm to correct higher order aberrations, in order to provide a quantitative estimate of the diameter of the beam required for an effective correction.

SUBJECTS AND METHODS

Subjects and Aberration Data

Aberrations for a 5.7-mm-diameter pupil were obtained in 109 right eyes of 109 normal subjects. Subjects having any kind of pathology or surgery were not included. All subjects had 20/20 visual acuity with correction in spectacles or contact lenses. Refractive errors ranged from -9.00 to +5.50 diopters (D) sphere, and from 0 to 3.00 D cylinder. Mean refractive error (\pm one standard deviation) was -2.00 ± 3.00 D sphere, and 0.50 ± 0.50 D cylinder. No restrictions were placed on corneal curvature requirements. Subjects ranged in age between 21 and 65 years (41 ± 10 yr). The root-mean-square (rms) of the wave aberration for higher orders (without defocus and astigmatism), averaged in our

From the University of Murcia, Murcia, Spain (Guirao) and the University of Rochester, Rochester, NY (Williams, MacRae)

The authors have no proprietary interest in the materials presented herein.

Correspondence: Antonio Guirao, PhD, Departamento de Física, Universidad de Murcia, Laboratorio de Óptica, Campus Espinardo, Edificio C, 30071, Murcia, Spain. Tel: 34.968.228314; Fax: 34.968.363528; E-mail: aguirao@um.es

Received: August 6, 2001

Accepted: August 20, 2002

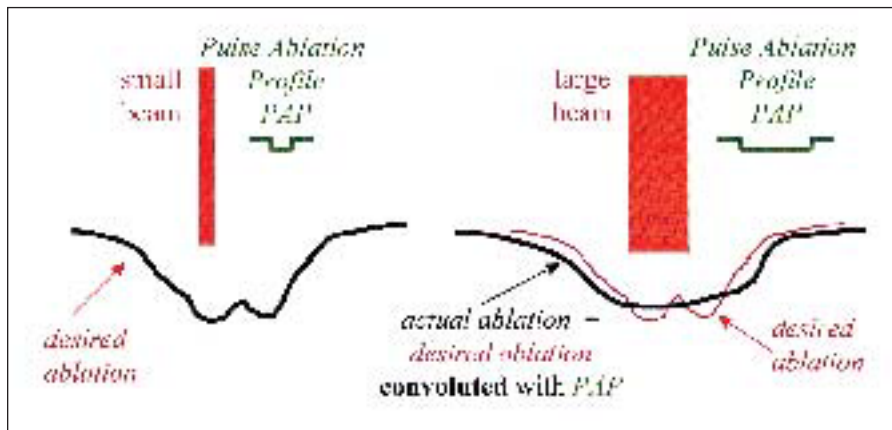


Figure 1. Illustration of corneal customized ablation. The pulse ablation profile (PAP) is the ablation produced by a single pulse (a top-hat profile is illustrated). The superposition of multiple pulses with a very small laser beam diameter (noted on the left) can shape the cornea, creating fine structures to correct the higher order aberrations of the eye. A beam that is too large is incapable of sculpting with the required amplitude. The actual corneal shape is the convolution of the desired shape and the PAP. Note that the larger beam (right) cannot reproduce the two small valleys corresponding to high frequency. The larger beam is only able to reproduce a big valley with attenuated amplitude.

109-eye population, is $0.33 \pm 0.12 \mu\text{m}$ (range 0.13 to $0.65 \mu\text{m}$).^{4,9}

Aberrations in an additional sample of nine eyes of nine normal subjects (21 to 38 years old) measured³ for a large pupil of 7.3 mm were also used to compare the effect of beam size for different pupil sizes. These subjects had normal visual acuity, and correction of defocus and astigmatism of less than 3.00 D. The average root-mean-square of these wave aberrations (truncated to 5.7 mm) was $0.30 \pm 0.14 \mu\text{m}$ (range 0.16 to $0.57 \mu\text{m}$; ie, in the range of the larger population of typical normal eyes, noted previously). Calculated average modulation transfer functions for the population of 109 eyes and for the 9-eye sample were not statistically different.

A third group was considered: aberrations, for a 5.7-mm pupil, in four eyes of four patients with keratoconus. The mean refractive correction for these patients was -2.00 ± 0.75 D sphere (range -1.25 to -2.25 D), and 2.25 ± 2.00 D astigmatism (range 0.25 to 5.00 D). The higher order aberrations in these patients have a root-mean-square range between 0.47 and $2.53 \mu\text{m}$ (mean $1.54 \pm 1.09 \mu\text{m}$). This means that the best keratoconic eye has aberrations similar in magnitude to the most aberrated normal eyes. On average, the higher order aberrations in the keratoconics are about four times larger than in the normal eyes.

Wave Aberration Measurement

The ocular wave aberration is the optical deviation of the actual wavefront formed by the eye from an ideal wavefront at every location in the pupil¹⁰, and is the sum of individual aberrations including defocus and astigmatism, as well as many other higher order aberrations such as coma, spherical aberration, trefoil, etc. To measure the eye's wave aberration, we used a Shack-Hartmann custom non-commercial wavefront sensor. Details of the experimental system have been described.^{3,4} The system

used to measure the aberrations in the 109 normal eyes and the sample of 4 keratoconic eyes had a total of 57 lenslets that sampled the 5.7-mm diameter pupil. Wave aberration measurements were done with natural accommodation and pupil. The subject's eyes were aligned with respect to the center of the natural pupil using a video camera while they fixated to infinity. The wave aberration was calculated for a 5.7-mm-diameter pupil and was expressed as a linear combination of Zernike polynomials^{4,11} up to fifth order (fifth order aberrations). The system to measure the wave aberration in the additional sample of nine eyes had a total of 217 lenslets that sampled a 7.3-mm-diameter pupil.³ In these subjects, the pupil was dilated with tropicamide (1%). The alignment was carried out with respect to the center of the natural pupil by a subjective technique. Wave aberrations were obtained for a pupil of 7.3 mm and a truncated pupil of 5.7 mm, and expressed as a Zernike expansion up to tenth order.

Aberrations were assumed constant with wavelength, except defocus. An extra defocus was added to the wave aberrations to account for the shift corresponding to longitudinal chromatic aberration from 780 nm (or 633 nm) to 555 nm.

Analysis of Beam Size Effect

Due to the finite size of the laser beam, the actual corneal shape after ablation will be the convolution of the pulse ablation profile with the desired corneal shape (Fig 1). The pulse ablation profile (PAP) is closely related to, but distinct from, the irradiance profile of each laser pulse. The pulse ablation profile and irradiance profile would have the same shape if the ablation process were perfectly linear. The operation of convolution may be physically understood as the sum of the ablations produced by all of the pulses. In principle, the number of pulses at the same corneal location is

proportional to the depth of the desired ablation at that location. However, since the beam has a finite diameter, the number of pulses will be proportional to the average depth of the desired ablation within the area covered by the pulse. Thus, the superposition of multiple pulses, with a very small laser beam diameter, can shape the cornea more finely than a large diameter beam.

After ablation, we have a residual wave aberration (WA) that is the difference between the ideal and the actual wave aberration:

$$WA_{res} = WA_{ideal} - WA_{actual} \quad (1)$$

Mathematically, the actual wave aberration is the convolution of the ideal wave aberration with the pulse ablation profile (PAP):

$$WA_{actual} = WA_{ideal} \otimes PAP \quad (2)$$

If the beam is infinitely small (Delta function), the convolution would not change the ideal wave aberration and there would be no residual aberrations.

Taking the Fourier transform on both sides of Eq. (2) and then taking the squared modulus we get

$$|FT(WA_{actual})|^2 = |FT(WA_{ideal})|^2 \cdot |FT(PAP)|^2 \quad (3)$$

Fourier analysis is a useful tool for analyzing the effect of the beam size. The squared modulus of the Fourier transform is called the power spectrum.¹² The power spectrum provides a measure of how rapidly the wavefront changes its height across the pupil. For example, if the wavefront consists of smooth rolling hills, then the power spectrum is confined to low spatial frequencies. However, if the wavefront has jagged hills and valleys spaced closely together, then the power spectrum extends to higher spatial frequencies. In Eq. (3), the power spectrum of the wave aberration indicates the spatial frequency content of the wave aberration. As a general rule, the more the power spectrum extends to higher spatial frequencies, the more the higher order aberrations in the wave aberration.

The pulse ablation profile also has a power spectrum, the shape of which indicates how much the beam will blur or spatially filter the correction. For example, if the pulse ablation profile were infinitely narrow, it would have a flat power spectrum. This would mean that the actual wave aberration is equal to the ideal wave aberration and the customized correction would be perfect. However, for wide beams, the power spectrum of the pulse

ablation profile is a function that falls to zero at lower and lower spatial frequencies as beam size increases. Because the power spectrum of the actual wave aberration is the product of the power spectrum of the ideal wave aberration and the power spectrum of the pulse ablation profile, larger beams smooth away the details (high spatial frequencies) in the actual wave aberration relative to the ideal wave aberration. In other words, the pulse ablation profile acts as a low pass spatial filter.

We applied the above-described mathematical tools to the correction of ocular wave aberrations by customized ablation with the following assumptions. Numerical computation of convolutions was made by using matrices of 128 x 128 pixels for describing the wave aberrations. Implicit in the application of the convolution is the assumption that an adequate number of light pulses have been used to sculpt the cornea. Errors can occur, not only because of too large a beam, but also if the placement of the light pulses is in error, or too few are used. These latter two factors are not treated here. The ablation depth of a single pulse was normalized so the volumes of the pulse ablation profiles were equal to one. We assumed the pulse ablation profile to have a top-hat shape, ie, a circle function with constant intensity within a fixed diameter falling to zero everywhere outside. To determine the spatial frequency content of the eye's aberrations, we calculated the power spectrum of the wave aberration for each eye in the population. We calculated radial profiles for the power spectra by averaging over all directions in the two-dimensional frequency domain.

Retinal Image Quality

For each subject, we considered the ocular wave aberration and calculated the residual wave aberration with equations (1-2) expected after an ablation with laser beams of 0.5, 1.0, 1.5, and 2.0 mm in diameter. From the residual wave aberration, we calculated the modulation transfer function, in monochromatic light, as a measure of the retinal image quality after the theoretical ablation. The generalized pupil function, $P(x,y)$, was first calculated as

$$P(x,y) = p(x,y) \cdot \exp[i(2\pi/\lambda)WA(x,y)] \quad (4)$$

where $p(x,y)$ denotes a pupil circular aperture.¹² The modulation transfer function (MTF) is the modulus of the optical transfer function (OTF) calculated as the complex autocorrelation of $P(x,y)$:

$$OTF(u,v) = \iint P(x,y)P^*(x-u, y-v)dxdy \quad (5)$$

The MTF calculated in this way is a function of a two-dimensional spatial frequency coordinate, (u,v) . A one-dimensional MTF was computed averaging the two-dimensional MTF across all angles. The MTF characterizes the ability of the eye to form a sharp image on the retina. The Y-axis of the MTF is the modulation or contrast transferred by the eye's optics. The X-axis represents sine waves with spatial frequency varying from low (large spacing between adjacent white bars) to high (fine gratings).

To quantify the benefit with a certain beam diameter compared to the benefit with a larger beam, we calculated a parameter, hereafter called "extra visual benefit," as the ratio of the smaller (0.5, 1.0, or 1.5 mm) beam MTF and the larger (1.0, 1.5, or 2.0 mm) beam MTF at the spatial frequency of 16 cycles/deg. This extra visual benefit was calculated for every eye and for the following three cases: extra benefit from 2.0 to 1.5 mm, from 1.5 to 1.0 mm, and from 1.0 to 0.5 mm.

Definition of "Effective Correction"

A problem arises with the concept of "effective correction," since its definition will determine the size of the beam required for customized ablation. We used three different definitions that result in similar conclusions; two are based on the power spectrum of the wave aberration and the filtering made by the pulse, and the other is based on image quality.

The first definition refers to maximum spatial frequencies in the power spectrum of the wave aberration. The correction is effective while spatial frequencies below a "maximum frequency" are filtered less than a threshold of 99%—the maximum frequency for which the power spectrum of the eye's wave aberration remains higher than 0.01. The definition is applied to the population considering the average power spectrum across the 109 eyes.

The second definition states that the correction is effective if the filtering causes at most a 10% loss of area of the power spectrum of the actual wave aberration relative to the area of the power spectrum of the ideal wave aberration. We calculate the loss percentage as

$$\left[1 - \frac{\text{area}(|FT(WA_{actual})|^2)}{\text{area}(|FT(WA_{ideal})|^2)} \cdot 100 \right] \quad (6)$$

where the wave aberrations are averaged in the 109-eye population.

Table 1
Maximum Spatial Frequencies for Each Aberration Order*

Order of Aberration	Maximum Spatial Frequency (cycles/mm)
Second	~0.5
Third	~0.65
Fourth	~0.85
Fifth	~0.85

**Frequencies with power spectral density <0.01*

Finally, a third definition considers that the correction is effective if the extra visual benefit (as defined in the previous section) produced by further reducing the beam diameter is not higher than 1.5 for any eye in the population.

RESULTS

Power Spectrum and Spatial Filtering Produced by the Beam

Figure 2 shows the power spectrum of the wave aberration, averaged in the population of 109 eyes, when the wave aberration includes only second, third, fourth, and fifth order aberrations. The higher the aberration order, the higher the content of high frequencies. Following from the practical definition of maximum spatial frequency given in the Methods section (frequencies for which the power spectrum is below 0.01 are neglected), we obtain the maximum frequencies contained in each order listed in Table 1. The maximum spatial frequency contained in the wave aberration including all orders is about 0.85 cycles/mm in the averaged population.

Figure 3 shows the power spectrum of the top-hat pulse ablation profile for 0.5 mm, 1.0 mm, 1.5 mm, and 2.0 mm beam diameters. The filtering produced by the beam is evident in Figure 3. As the spatial frequency of the structure to be created by ablation increases, the amplitude with which these structures will be produced decreases, falling to zero. The dashed line in the inset in Figure 3 represents a threshold value of 0.01, indicating a 99% filtering. The thresholds are therefore 0.55, 0.65, 1.1, and 2.1 cycles/mm for beams of 2.0, 1.5, 1.0, and 0.5 mm, respectively. Thus, according to the first definition of effective correction and from Table 1, the 2-mm beam could effectively correct the second order aberrations, the 1.5-mm beam could correct up to third order, and beams of 1 mm or less could correct aberrations up to fifth order.

The product of the curves in Figure 3 with the power spectrum of the eye's wave aberration gives

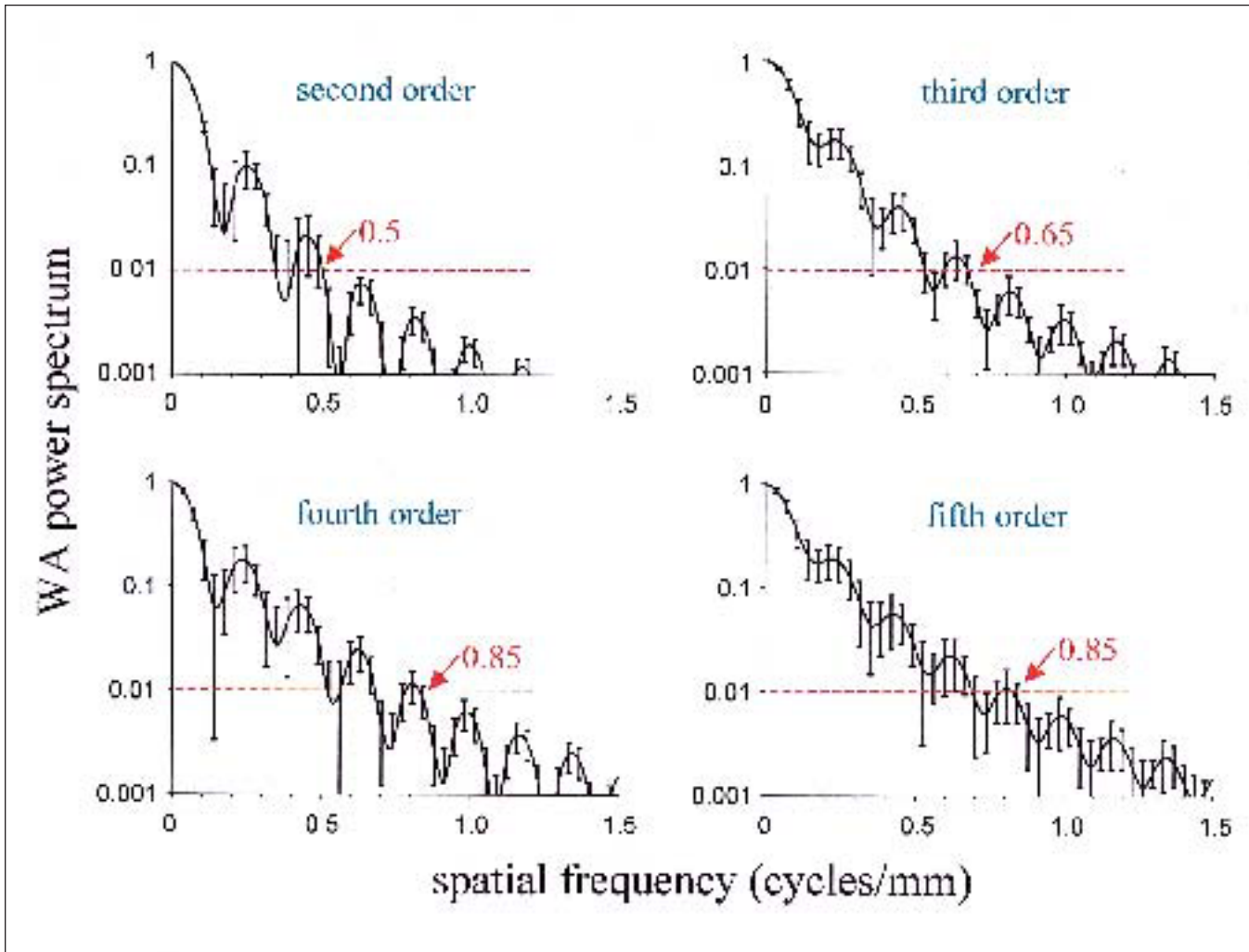


Figure 2. Average power spectrum of the wave aberration in the population \pm one standard deviation, when the wave aberration includes only second, third, fourth, and fifth order aberrations, respectively. The scale on the Y-axis is logarithmic. The dashed red line indicates an arbitrary limit for the power spectrum such that the contribution of spatial frequencies below 0.01 is considered negligible. The arrows and numbers indicate the maximum spatial frequency contained in each aberration order. These maximum frequencies correspond to the interception of the 0.01 horizontal threshold lines with the average power spectra.

the power spectrum of the actual correction. The loss percentages calculated with Eq. (6) are listed in Table 2 for each aberration order. The numbers with an asterisk in Table 2 are the losses lower than 10%. Thus, according to the second definition of effective correction in the Methods, the 2-mm beam can produce an effective correction of second order aberrations; the 1.5-mm beam can create an effective correction up to third order; and beam diameters of 1.0 mm or less can effectively correct up to fifth order.

Table 2
Percent Losses (Eq. 6) in Power Spectrum Produced by Beam Spatial Filtering for Each Aberration Order

Aberration Order	Beam Diameter			
	2.0 mm	1.5 mm	1.0 mm	0.5 mm
Second	10*	8*	4*	1*
Third	17	10*	7*	4*
Fourth	20	15	9*	3*
Fifth	21	15	9*	3*

**Losses of 10% or lower*

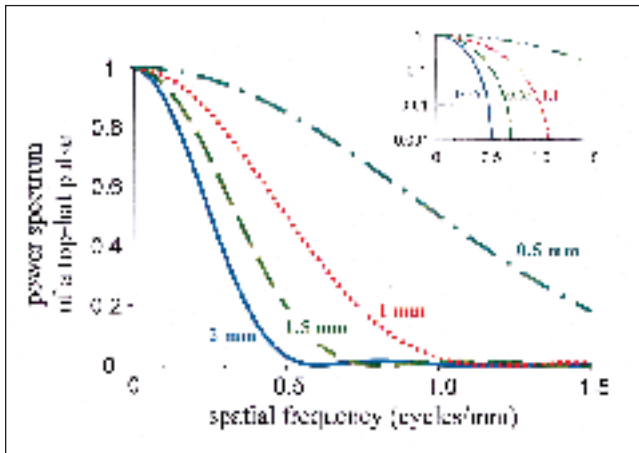


Figure 3. Power spectrum of the pulse ablation profile (top-hat shape) with diameters of 0.5 mm, 1.0 mm, 1.5 mm, and 2.0 mm. The X-axis indicates the spatial frequency of the corneal pattern that the ablation must shape. The Y-axis indicates the fraction of the amplitude that the beam can shape for every spatial frequency. The inset is the same with the Y-axis in logarithm scale. The dashed line in the inset represents a threshold of 99% filtering.

Image Quality and Visual Benefit for Different Beam Diameters

Figure 4 shows the modulation transfer function, averaged across the 109 eyes for the 5.7-mm pupil, and calculated from the residual aberrations remaining after an ablation with the different beam diameters. The MTF, corresponding to a conventional ablation of second order, to correct defocus and astigmatism, is also shown as a baseline for comparison, indicating the best performance achievable with a conventional ablation, assuming the desirable case when the conventional ablation does not introduce higher order aberrations. This second order correction was simulated with either a 2-mm or a 0.5-mm beam. The MTF was the same for both sizes, further indicating that a 2-mm beam is adequate for a conventional correction. The diffraction-limited MTF is also shown, corresponding to the ideal case, when all the aberrations are totally corrected. The MTF for a 0.5-mm beam diameter is practically diffraction limited. The 1-mm beam also yields a good MTF. The performance progressively deteriorates when the beam size increases. Even the customized correction with the 2-mm beam is still better than the non-customized case, however it is not as high as it could be by using a smaller beam.

Table 3 indicates the percentage of eyes with extra visual benefit, larger than 1.3, 1.5 or 2, obtained with the 1.5-mm beam compared to the 2.0-mm beam, with the 1.0-mm compared to the 1.5-mm, and with the 0.5-mm beam compared to the 1.0-mm beam. The number of eyes with an extra

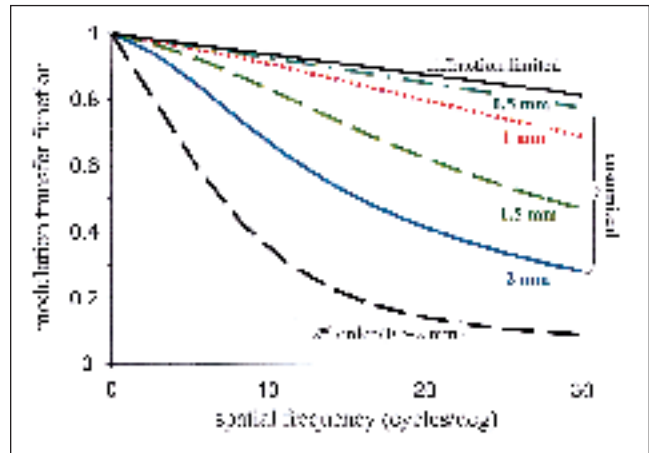


Figure 4. Modulation transfer function (MTF), averaged across the 109 normal eyes, for a 5.7-mm pupil, for the four beam diameters. The MTFs were calculated in monochromatic light (555 nm) from the residual wave aberration after the theoretical customized ablation with the different beam sizes. These MTFs are a measure of the retinal image quality expected after customized ablation. The more blurring in the retinal image, the more the modulation transfer departs from one and approaches zero. Also shown is the MTF corresponding to a conventional ablation that corrects only defocus and astigmatism (with either a 2-mm or a 0.5-mm beam), and the diffraction-limited MTF.

Beam Diameter Reduction	>1.3	>1.5	>2
From 2 to 1.5 mm	65	40	9
From 1.5 to 1 mm	41	23	5
From 1 to 0.5 mm	1	0	0

benefit higher than 1.5 is 40%, 23%, and 0%, respectively. Thus, by placing the limit of significant improvement at 1.5 for all eyes, the progressive reduction of the beam diameter up to 1.0 mm produces a significant improvement. The step from 2.0 mm to 1.5 mm yields a fraction of 9% of the eyes with benefit higher than 2. The step from 1.0 mm to 0.5 mm yielded small benefits, always below 1.3 for 99% of the eyes.

Figure 5 shows the MTFs averaged across the four keratoconic eyes for customized ablation with the different beam diameters (see Fig 4 to compare the same results in normal eyes). Even in these highly aberrated eyes, the 0.5-mm beam gives again a practically diffraction limited MTF. The MTFs for the larger beams are not as good as the MTFs in normal eyes for the same beam sizes. One might say that the 1-mm (1.5-mm) beam performs in the keratoconic eyes like the 1.5-mm beam (2.0-mm) in normal eyes.

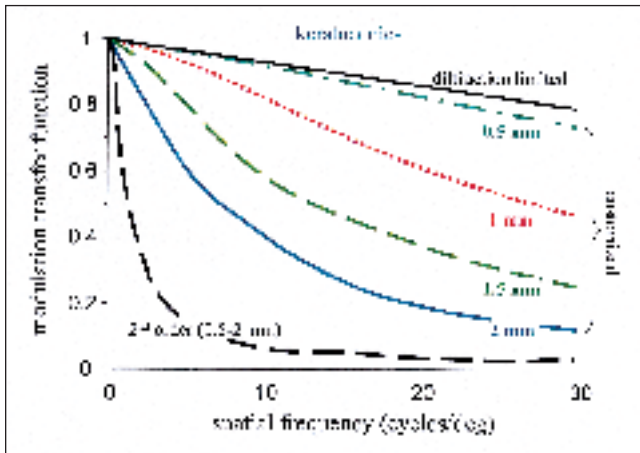


Figure 5. Same as Figure 4 for a sample of four keratoconic eyes, with average higher order aberration approximately four times larger than the normal eyes.

Beam Size Effect for Different Pupil Diameters

The power spectra of the eye’s wave aberration, averaged across nine eyes, were similar for a 7.3-mm and a 5.7-mm pupil, indicating that the content of spatial frequencies in the wave aberration for both pupils is similar. This is because the magnitude of aberrations increases with the pupil, but the spatial frequency of the aberrations decreases as one moves from points at the center of the pupil to points at the edge. To illustrate this, one can consider the example of a wave aberration consisting of a pure trefoil aberration. The structure in the trefoil consists of three cycles (three valleys and three hills). The three cycles are spread along a length of $2\pi r$ (r is the radial position). Thus, the spatial frequency of the trefoil structure is 0.17 cycles/mm at the edge of a 5.7-mm pupil, and 0.13 cycles/mm at the edge of a 7.3-mm pupil. At the edge of large pupils where the amplitude of higher order aberrations is significant, the spatial frequency of the aberration structures is lower. This is an interesting result, which implies that a similar beam diameter could be used for different pupil size ablations. Figure 6 shows the MTFs for a customized ablation with the 1.0-mm beam and the 1.5-mm beam for two pupils. The expected benefit with the 1.0-mm beam is similar for both the smaller and larger pupils.

Top-hat and Gaussian Beams

We used a model with top-hat beam profiles. Gaussian beams are also common. In Figure 7, we compare the power spectrum of both profiles. The top-hat beam is equivalent in its filtering effect to a Gaussian beam with a full-width half-maximum (FWHM) equal to 3/4 of the top-hat beam diameter.

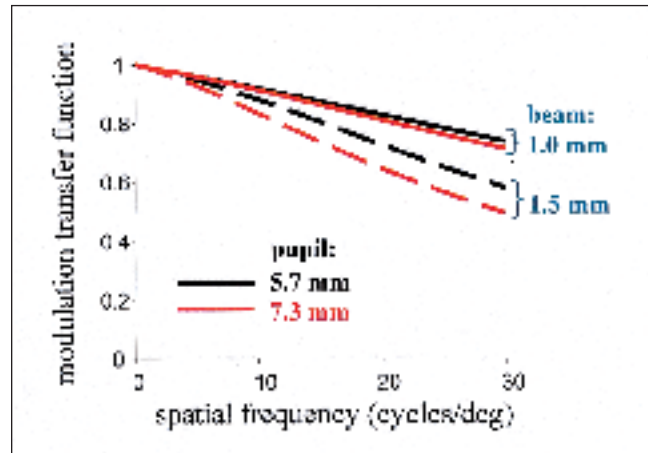


Figure 6. Theoretical modulation transfer functions, averaged across nine normal eyes, for pupils of 7.3 mm and 5.7 mm, for customized ablations with beams of 1.0 and 1.5 mm.

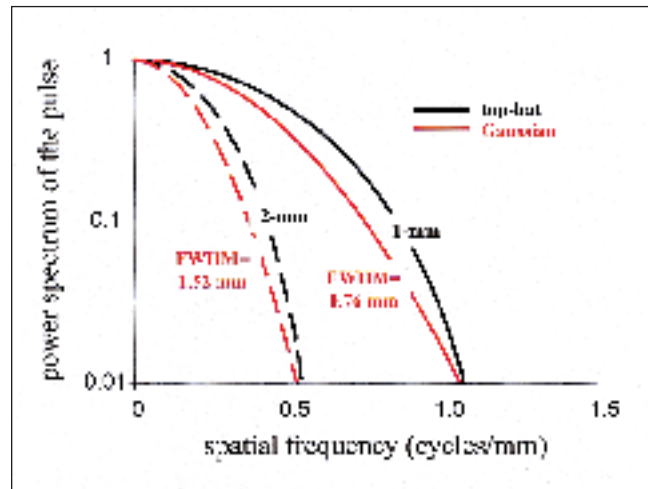


Figure 7. Power spectrum of the pulse ablation profile for a top-hat (black) and a Gaussian profile (red). The beam diameters are 1 and 2 mm for the top-hat beam, and the FWHM (full-width half-maximum) values are 0.76 and 1.52 mm for the Gaussian beam.

Thus, the results for the 1-mm (2-mm) top-hat beam are similar to those for a Gaussian beam that has a FWHM of 0.76 (1.52) mm.

DISCUSSION

We have theoretically analyzed the effect of the beam size on the expected benefit of customized laser refractive surgery by using actual data of wave aberrations from real eyes. The potential benefit of correcting the aberrations is seen in Figure 4 as the increase from the MTF for a conventional ablation to the MTF for a customized ablation using various beam diameters. We used three definitions to consider an “effective correction” if 1) the beam causes a loss lower than 10% in the content of spatial

structures of the wave aberration, 2) if it filters less than 99% the frequencies that fall below a threshold frequency for which the power spectrum is 0.01, or 3) if the further reduction of the beam size does not produce [for any eye] an extra visual benefit (ratio of MTFs at 16 cycles/deg) of more than 1.5 times. The conclusions from our different analyses—based on beam filtering and power spectrum or based on retinal image quality—were similar. The 2-mm beam performs well for a conventional ablation to correct only defocus and astigmatism. The theoretical MTF after correction is actually the same with the 2-mm beam and 0.5-mm beam. Smaller beams are required only for customized ablations, not for conventional ablations. The 1.5-mm beam can also correct the third order aberrations. Beams of 1 mm or less can correct aberrations up to fifth order in a normal eye. The expected MTFs after customized ablation with beams of 1 mm or less are close to the limit imposed by diffraction.

We found that the frequency content in the wave aberration of the eye is similar for different pupil sizes. This means that the same beam size can correct higher order aberrations, regardless of pupil diameter. Our conclusions about optimum beam size are not very different when we consider cases of highly aberrated eyes. Although refractive surgery is not usually an option for keratoconic patients, we used their wave aberrations to have an estimate of the beam effect in extreme cases. The MTF for a customized ablation with a 0.5-mm beam was again practically diffraction limited.

Huang and Arif^{13,14} investigated the diameter of the beam required to have a useful customized ablation. Their study was based on the fraction of root-mean-square of the wave aberration corrected, unlike our study based on Fourier analysis and retinal image quality. Their conclusions were, however, the same as ours. By using a criterion of useful correction defined as “achieved correction at least twice that of induced aberration,” they found that a 2-mm beam is useful only for correction up to second order. A 1-mm beam is useful for correction up to fourth order, according to their criteria. Since ocular aberrations above fourth order are relatively insignificant, their main conclusion is that a beam of 1 mm or less is theoretically capable of eliminating most of the higher order aberrations.

Besides the small beam size, it is also desirable to minimize treatment duration. There are a number of tradeoffs when selecting an optimal beam diameter for customized ablation. For instance, although a 0.5-mm beam can offer almost total correction, a

1-mm beam may be preferred because it can deliver an effective customized ablation in much less time. Another argument in favor of the 1-mm beam is that a small decentration of approximately 0.5 mm will negate the performance of the 0.5-mm small beam. The ablation can be done in two steps: a broad area ablation (2-mm beam) for correcting most of the astigmatism and defocus, followed by a small-beam (1.0 to 0.5-mm) contouring for correcting fine aberrations.

We did not consider the effect of quantization due to the finite amount of tissue that a single pulse removes. The effect of the quantization would be a stromal profile that looks like the steps of a stair. However, it is likely that the flap in laser in situ keratomileusis (LASIK), or the corneal epithelium in photorefractive keratectomy (PRK), may smooth this fine stair step structure. On the other hand, the flap itself could act as an additional spatial filter that reduces the high spatial frequency correction created by customized ablation on the stroma. This would imply that our calculations overestimate the effectiveness of the customized ablation. If flap filtering does affect higher order aberration correction, the benefit of using a smaller beam may be reduced or eliminated. Although we used actual wave aberration data from normal non-treated subjects, our analysis is theoretical. In practice, even conventional laser refractive surgery still warrants additional understanding. We, and others, have noted that standard ablation procedures correct defocus and astigmatism but induce higher order aberrations that were not present in the preoperative eye.¹⁵ Factors such as decentration, healing or biomechanical effects, or long-term stability of the corneal profile could be responsible for these effects. These factors must also be considered in evaluating the outcome of customized refractive surgery. Thus, for example, the healing or biomechanical effect could eliminate subtle structures created by customized ablation. Also, our simulations did not consider the problem of transition zones and the possibility of obtaining better results by means of optimization algorithms that make use of deconvolution methods.

We calculated MTFs in monochromatic light. In everyday conditions, the eye's chromatic aberration further deteriorates the retinal image. Even when we correct higher order aberrations, chromatic aberration is a wavelength-dependent defocus that decreases MTF. The potential visual benefit from the conventional ablation to the customized ablation will be more modest in white light.⁹

We believe that an excimer laser beam size of 1 mm or less (FWHM<0.76 mm for Gaussian beams) is sufficient to treat wavefront aberrations of typical human eyes. Spot sizes larger than this may be useful for treating refractive errors such as sphere and cylinder.

REFERENCES

1. MacRae SM, Schwiegerling J, Snyder R. Customized corneal ablation and super vision. *J Refract Surg* 2000;16(suppl): S230-S235.
2. Mrochen M, Kaemmerer M, Seiler T. Wavefront-guided laser in situ keratomileusis: Early results in three eyes. *J Refract Surg* 2000;16:116-121.
3. Liang J, Williams DR. Aberrations and retinal image quality of the normal human eye. *J Opt Soc Am A* 1997;14: 2873-2883.
4. Porter J, Guirao A, Cox IG, Williams DR. The human eye's monochromatic aberrations in a large population. *J Opt Soc Am A* 2001;18:1793-1803.
5. Liang J, Williams DR, Miller DT. Supernormal vision and high resolution retinal imaging through adaptive optics. *J Opt Soc Am A* 1997;14:2884-2892.
6. Williams DR, Yoon GY, Porter J, Guirao A, Hofer H, Cox IG. Visual benefit of correcting higher order aberrations of the eye. *J Refract Surg* 2000;16:554-559.
7. Yoon GY, Williams DR. Visual performance after correcting the monochromatic and chromatic aberrations of the eye. *J Opt Soc Am A* 2002;19:266-275.
8. Williams DR, Yoon GY, Guirao A, Hofer H, Porter J. How far can we extend the limits of human vision. In: MacRae S, Krueger R, Applegate RA, (eds). *Customized Corneal Ablation: The Quest for SuperVision*. Thorofare, NJ: SLACK Incorporated; 2001:11-32.
9. Guirao A, Porter J, Williams, DR, Cox, IG. Calculated impact of higher order monochromatic aberrations on retinal image quality in a population of human eyes. *J Opt Soc Am A* 2002;19:620-628.
10. Mahajan VM. *Aberrations Theory Made Simple*. Washington, DC: SPIE Press; 1991:3-19.
11. Wang JY, Silva DE. Wave-front interpretation with Zernike polynomials. *Appl Opt* 1980;19:1510-1518.
12. Goodman JW. *Introduction to Fourier Optics*, 2nd edition. New York, NY: McGraw-Hill; 1996.
13. Huang D, Arif M. Spot size and quality of scanning laser correction of higher order wavefront aberrations. *J Refract Surg* 2001;17(suppl):S588-S591.
14. Huang D, Arif M. Spot size and quality of scanning laser correction of higher-order wavefront aberrations. *J Refract Surg* 2002;28:407-416.
15. Applegate RA, Howland HC, Klyce SD. Corneal aberrations and refractive surgery. In: MacRae S, Krueger R, Applegate RA, (eds). *Customized Corneal Ablation: The Quest for SuperVision*. Thorofare, NJ: SLACK Incorporated; 2001:239-246.