Transferring Wavefront Measurements Into Corneal Ablations: An Overview of Related Topics

Michael Mrochen, PhD; Michael Bueeler, PhD; Hans Peter Iseli, MD; Farhad Hafezi, MD; Theo Seiler, MD, PhD

ABSTRACT

We give an overview of possible side effects that are specific for, or of particular relevance in, customized treatments. Certain processes involved in customized ablations have the potential to alter the quality of the optical correction. Professionals associated with customized treatment should be informed and trained with respect to possible sources of error. [*J Refract Surg* 2004;20:S550-S554]

The aim of this report is to give an overview of side effects and possible sources of error that might affect predictability, efficacy, or safety when transferring a theoretical ablation profile onto a vital cornea. To demonstrate the complexity of this issue, we are focusing on side effects directly linked to the technology used for pre- and intraoperative diagnostics and treatment (Fig 1). Our intention with this report is to sensitize clinicians and other professionals to the impact of side effects on optical outcomes.

EYES NEED STANDARDIZATION

The influences of a patient's eye condition during corneal topography or wavefront sensing have been studied in detail and reported frequently in the literature. Reasons are manifold and range from tear film conditions to patients' ability to fixate, status of accommodation, or head tilts that are usually difficult to control. A problem of clinical relevance is the influence of the tear film during wavefront sensing. Usually, single spots are not well detected in Shack-Hartmann or Tscherning images and, thus, the wavefront is plagued with a larger error. The question is: Should the investigator apply artificial tears to achieve a more precise detection of spots with the risk of altering the individual wavefront by the introduction of this additional factor? This specific example might show the importance of standard procedures in clinical routine. Further clinical research should focus on such standards for wavefront sensing and corneal topography.

One should keep in mind that optical aberrations are not stable and vary over time and age.[1,2] In addition, fluctuations of optical aberrations with accommodation or the pulse heart rate have been shown.[3] Consequently, standards for wavefront sensing should address such factors.

FIXATION DURING MEASUREMENT AND TREATMENT

The report of optical errors such as wavefront aberrations with respect to the line of sight is accepted in ophthalmology. However, to determine the line of sight, precise measurements of the pupil location are not the only prerequisite. Centration also requires the patient's ability to fixate on a fixation light that is coaxially aligned to the optical axis of the measuring device. Creating a flap or removing the epithelium during treatment reduces the image quality of the fixation light. Thus, the patient may lose the capability of accurate fixation. Similarly, precision of detection of the entrance pupil is reduced due to light scattered by a rough cornea.

Generally, centration is a task of 6 degrees of freedom. The eye is able to perform horizontal and vertical shifts; it is able to rotate around its longitudinal, horizontal, and vertical axes; and it can move back and forth. Most commercially available
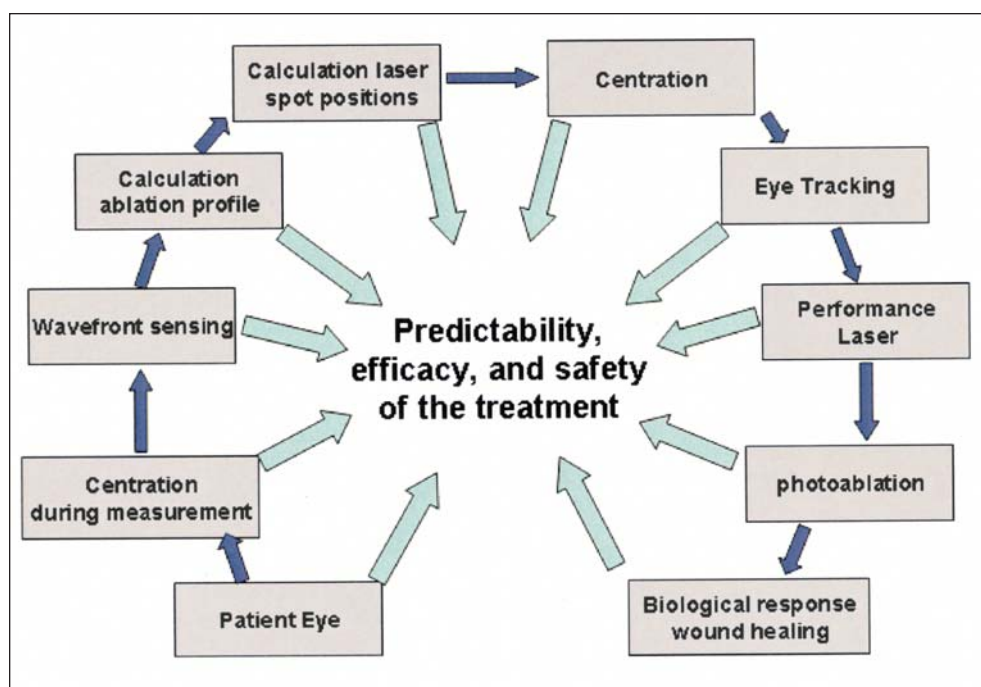
Figure 1. Overview of the processes involved in customized ablation. Each process has specific sources of error that may affect the outcome of a refractive procedure.

eye-tracking systems measure and compensate lateral eye movements along the horizontal and vertical axis of the eye and some eye-tracking systems are capable of compensating for rotations around the longitudinal axis (cyclotorsion). Nevertheless, all eye-tracking devices (centration and registration) need the cooperation of the patient.

Eye-tracking systems are built to compensate for eye movement and to center the measurement and the treatment with respect to the line of sight. Besides the benefits, the use of an eye-tracking system bares the risk of new possible sources of error such as inaccurate calibration between the detection of the eye's position (eg, video camera) and the optical path of the laser beam (scanning mirrors). Such an error may result in under- or overcorrections, systematic decentrations, or large spherical aberrations.

WAVEFRONT SENSING AND CORNEAL TOPOGRAPHY

Calibration errors or misalignment of the optical scheme are a source of error because the user might not be able to determine such errors before each measurement. Wavefront and topography data that are used for a customized treatment should be used only after verification of the system calibration. Whereas in optical diagnosis such errors might not play a significant role, they may have a substantial impact when treatments are based on data measured incorrectly. We should keep in mind that an ablation of only 10 μm can cause a change of

several diopters when the treatment is performed within a small zone. Manufacturers provide test eyes and certain verification methods for the calibration of their devices to avoid such calibration errors. Those tests are often time consuming but they should be performed before measurements for customized treatment are initiated.

WAVEFRONT RECONSTRUCT ALGORITHMS

In principle, wavefront sensors determine the first derivative of the wavefront (slope) at a specific point within the pupil. One gets a set of measured wavefront slopes distributed over the entire pupil. Wavefront reconstruction methods fit the slope data to a set of polynomials with a least square technique. Typically, the polynomial set for wavefront fitting has been either Zernike or Taylor polynomials. The least square techniques minimize the absolute error between the measured slopes and the reconstructed wavefront. From the principles of fitting, there is a residual error between the reconstructed wavefront and the measured slopes.[4] The amount of residual error depends on the number of sample points and the number of polynomials used for fitting. As a rule of thumb, the number of sampling points should be three times more than the number of polynomials (modes). Especially in highly aberrated eyes, the fitting errors can cause a significant under- or overestimation of wavefront aberrations and may lead to poor outcomes when used for ablation profile calculations.
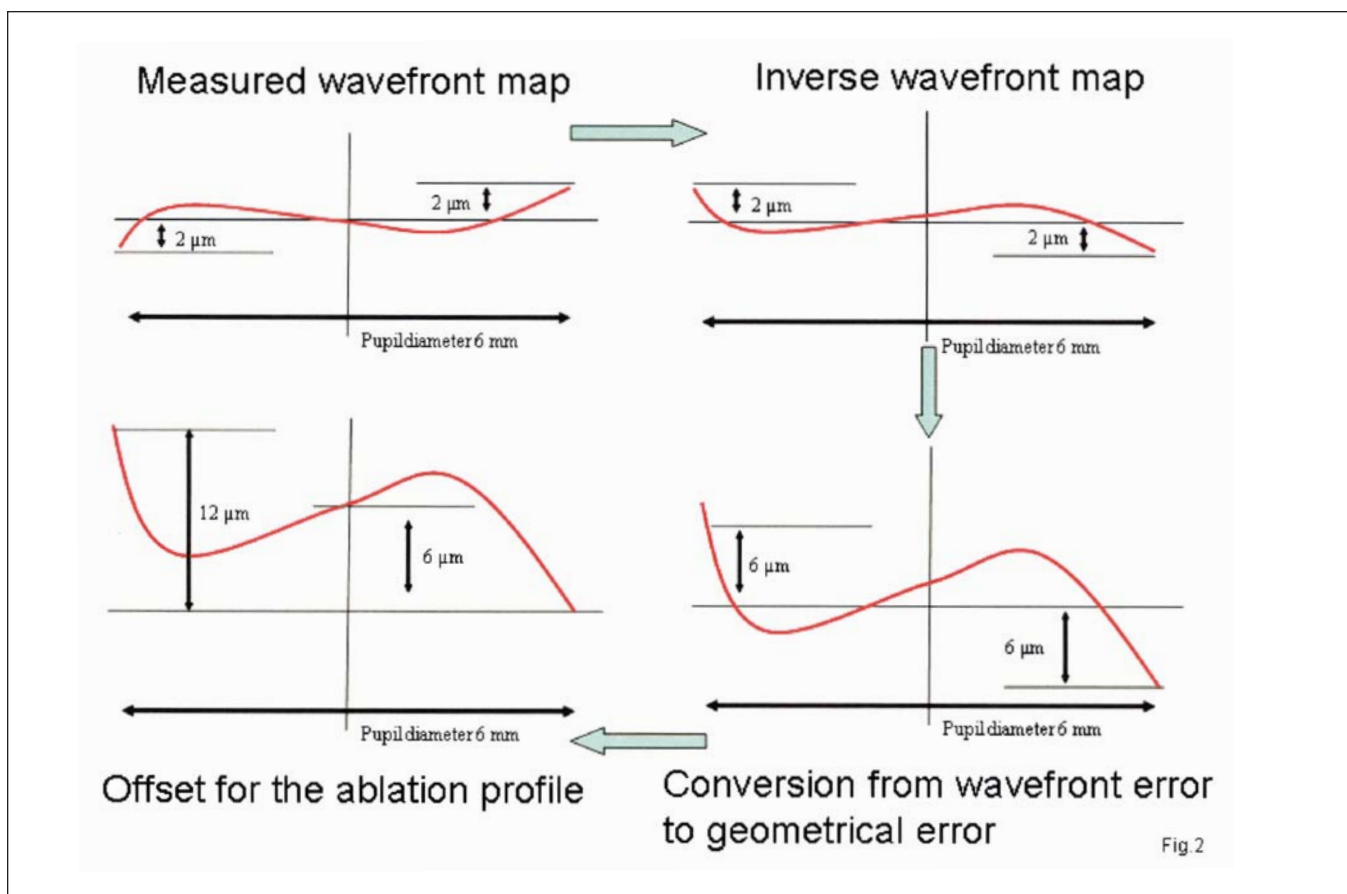
Figure 2. The principle of transferring a wavefront map into an ablation profile. Inversion of the wavefront map, conversion into geometrical shape information and offset for the ablation profile.

ABLATION PROFILE CALCULATION

The design of ablation profiles is usually based on theoretical eye models that should represent the optical behavior of the human eye. The simplest way is to use the refractive power and shape of the anterior front surface of the cornea as a basis to calculate the required tissue removal for a spherocylindrical correction within a given optical zone. This general assumption, originally made by Munnerlyn[5], was used to determine the amount of tissue that must be removed from a spherical cornea to achieve a postoperative spherical cornea with a different radius of curvature by simple geometrical considerations. This derivation assumed both the thin lens theory and paraxial optics. Later, various authors improved the assumptions for pre- and postoperative corneal shape by the introduction of corneal asphericity or methods for precompensating specific higher-order aberrations.

Three general steps can achieve an ablation profile from a wavefront measurement:

1) Inversion of the wavefront map: As the aim is to correct the wavefront, one must reverse signs/orientation of the wavefront. If two or more wavefront aberrations are calculated in the same reference plane, they can be added or subtracted.

2) Conversion of the wavefront map: The wavefront information must be transferred into corresponding geometrical shape information. In a first approximation, this can be done by simply assuming the wavefront is equal to the optical path difference in the eye, which is true for small aberrations usually observed in normal eyes. As the optical path difference is the product of geometrical length times the refractive index, one could easily derive the ablation profile in terms of a height map.

3) Offset for the ablation profile: As refractive surgery lasers are only able to remove tissue, one must consider this fact in ablation profile design. The geometrical information derived from the wavefront must be shifted by the amplitude of the height profile.

Figure 2 gives an example for this transfer of the ablation profile. An initial wavefront with peak to
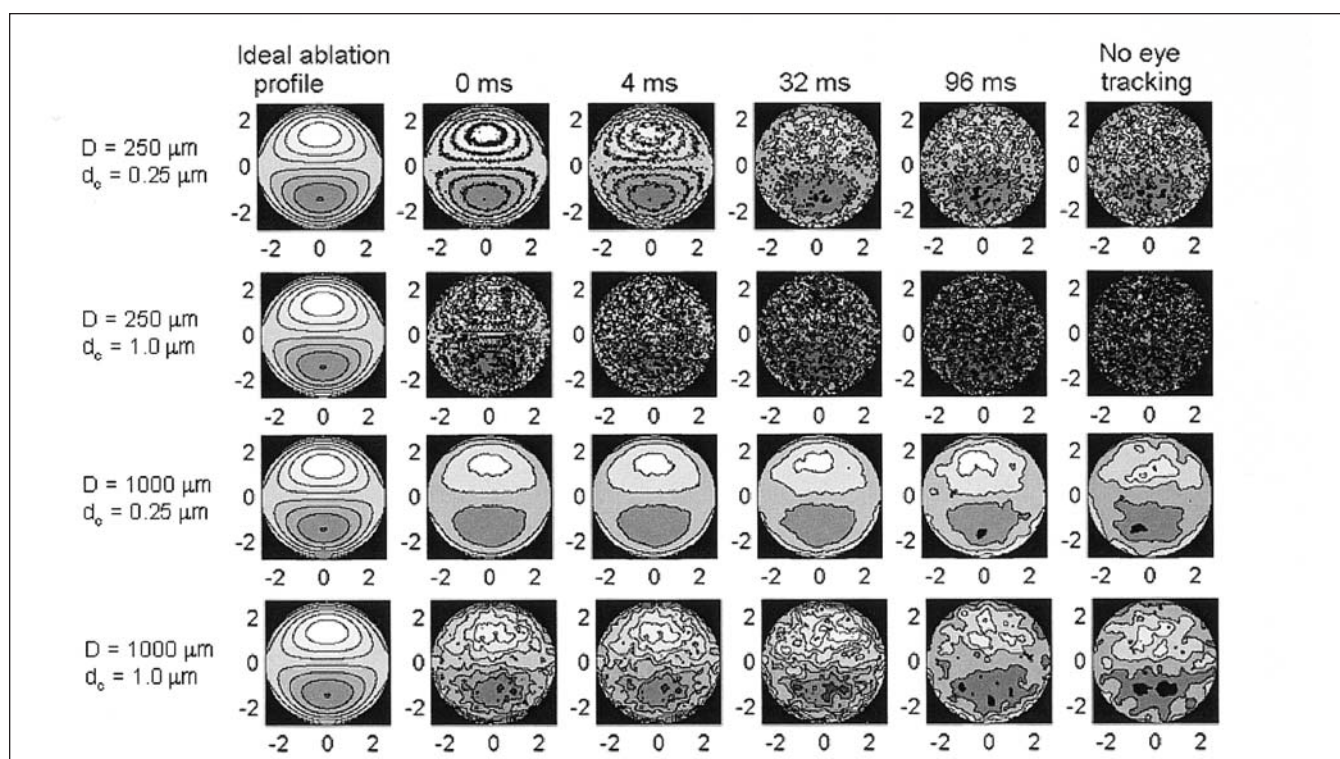
Figure 3. Ablated profiles after simulated scanning spot correction of vertical coma. Each row corresponds to a different combination of spot size D and central ablation depth per pulse d_c. The first column shows the ideal ablation profile. The following columns depict the approximations of the same profile resulting from scanning spot laser ablation performed with increasing eye tracker latency.

peak values of only 4 µm results in an ablation profile with a maximal ablation depth of approximately 12 µm.

Generally, ablation profile design requires an eye model including optical and geometrical assumptions of the individual eye. Including more measured data into such a model will improve the eye model in predicting the optical situation of an individual eye. In contrast, including more measured data for the individual ablation profile calculations also bares the risk of introducing more sources of error in the calculation process. We should remember that all measurements are plagued with a certain error and, if measurements are combined, the errors are usually increased—law of error propagation.

SPOT SIZE, SINGLE SPOT ABLATION DEPTH AND EYE-TRACKING LATENCY

Only limited data on the theoretical impact of intraoperative eye movements on the optical outcome of refractive treatments with a scanning spot laser are published.[6] In scanning spot laser surgery, the ideal corneal surface is approximated by directing and overlapping a finite number of laser pulses. Displacements of single laser shots from their ideal overlap positions may have the potential to significantly decrease the accuracy of the desired correction and to increase the surface roughness after ablation.

When the eye does not move, a small spot is able to correct finer details (higher Zernike modes) than a larger spot.[7] This situation changes when positioning errors of the single laser spots due to incomplete compensation of eye movements are introduced (Fig 3). In this more realistic case, a reduction of the spot diameter reduces the stability of the correction toward spot displacements. So far, combinations of a large spot diameter (typically 0.75 to 1.00 mm) and a small ablation depth per pulse (typically 0.2 to 0.3 µm) yield the best parameters in case of typical eye-tracking latencies (>1 ms).[6]

RADIANT EXPOSURE (FLUENCE)

Changes in the ablation depth for each single laser pulse, when moving the laser beam from the corneal apex toward the limbus, changes the angle of light incidence resulting in significant undercorrection in the periphery and consequently generation of spherical aberrations are known.[8] Energy drifts or fluctuations during a treatment may further influence the optical outcome.

Even more important is the stability of the laser ablation. If the radiant exposure is reduced and reaches values in the order of the ablation threshold, one receives unpredictable ablation depths. The ablation process gets more sensitive regarding environmental conditions such as temperature, individual variations in the ablation rates, humidity, alcohol concentration in the air, or air flow. For example, after the epithelium is removed or the flap is opened, the cornea starts to dehydrate. Decreasing the water content results in higher ablation efficiency for the collagen structure and, thus, in a higher amount of effective ablation depth after the cornea has been rehydrated.[9] Furthermore, the laser radiation might be shielded by a thin water film after the corneal wound bed is cleaned with a liquid before ablation. Even water has been assumed to have a small absorption coefficient at 193 nm radiation under physiological conditions; absorption increases by several orders of magnitude during the photoablation process.[10]

An increase in radiant exposure leads to a more stable and reliable ablation per pulse; however, increasing the radiant exposure will also increase the amplitudes of the acoustic waves that are induced during the ablation process.[11] Those high stress waves may harm the corneal endothelium or other fragile structures in the eye.

Consequently, the range of radiant exposures that can be used in clinical routine is limited to a range of approximately 150 to 600 mJ/cm^2 dependent on the type of beam profile used—Gaussian or top hat.

POSTOPERATIVE VARIABILITY IN WOUND HEALING

The influence of the individual variability in wound healing on optical outcomes in terms of wavefront aberrations is thus far unknown. Future clinical trials should provide data on biomechanical effects, epithelium remodeling, and stromal wound healing in terms of predictability of individual results. Based on such data, one may be able to simulate the expected outcome for a specific eye and to include such factors in the ablation profile design.

In summary, the use of wavefront aberrations and corneal topography as a basis for customized ablations is a complex matter. A perfect wavefront measurement does not necessarily guarantee a perfect result after treatment. Other factors related to the technology and the clinical status of the patient must be considered for predicting the outcomes of a customized ablation. Besides this, the chain of processes involved in customized surgery requires not only well-trained surgeons; all professionals involved in the diagnosis and treatment should be informed and trained with respect to possible sources of error of customized ablations.

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