Effect of air-flow on the evaluation of refractive surgery ablation patterns

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Abstract: An Allegretto Eye-Q laser platform (Wavelight GmbH, Erlangen, Germany) was used to study the effect of air-flow speed on the ablation of artificial polymer corneas used for testing refractive surgery patterns. Flat samples of two materials (PMMA and Filofocon A) were ablated at four different air flow conditions. The shape and profile of the ablated surfaces were measured with a precise non-contact optical surface profilometer. Significant asymmetries in the measured profiles were found when the ablation was performed with the clinical air aspiration system, and also without air flow. Increasing air-flow produced deeper ablations, improved symmetry, and increased the repeatability of the ablation pattern. Shielding of the laser pulse by the plume of smoke during the ablation of plastic samples reduced the central ablation depth by more than 40% with no-air flow, 30% with clinical air aspiration, and 5% with 1.15 m/s air flow. A simple model based on non-inertial dragging of the particles by air flow predicts no central shielding with 2.3 m/s air flow, and accurately predicts (within 2 μm) the decrease of central ablation depth by shielding. The shielding effects for PMMA and Filofocon A were similar despite the differences in the ablation properties of the materials and the different full-shielding transmission coefficient, which is related to the number of particles ejected and their associated optical behavior. Air flow is a key factor in the evaluation of ablation patterns in refractive surgery using plastic models, as significant shielding effects are found with typical air-flow levels used under clinical conditions. Shielding effects can be avoided by tuning the air flow to the laser repetition rate.

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OCIS codes: (170.1020) Ablation of tissue; (170.3890) Medical optics instrumentation; (330.5370) Physiological optics; (330.4460) Ophthalmic optics; (220.1000) Aberration compensation.

References and links


10. FDA, “Ophthalmic devices panel 110th meeting,” (Food and Drug Administration, 2008).


Ablating plastic surfaces is a common procedure for the evaluation and calibration of refractive surgery lasers. Recently, experimental models based on the ablation of plastic model corneas have been used as research tools in the exploration of the limits of refractive surgery [1-2]. The study of the ablation on plastic materials allows analysis of the physical processes without the biological variability of tissue. The use of plastic model corneas in refractive surgery has allowed deeper understanding of the procedure, including the origin of the roughness of the ablated surface [3-5], the calibration of the laser fluence and the evaluation of its stability [6], the optimization of ablation patterns [1, 7]—including wavefront guided correction-, the sources of surgically induced aberrations [6], or the compensation of geometrical energy losses [8]. Paradoxically, purely biological effects such as wound healing or biomechanical deformations [9] can also be explored by means of plastic models, through the comparison of pure physical experimental predictions with the clinical outcomes [1].

Plastic models have also been proposed by regulatory agencies [10] as a tool for security and efficacy assessment of new laser platforms or ablation algorithms [11–13], due to their high quality control potential.

The key question approached with physical models is the accuracy of the transfer of an energy pattern generated by the laser to a predictable depth map in the material. The Beer-Lambert Law governs the non-linear relationship between the laser fluence (\(F\)) and the depth of the ablation (\(d\)):

\[
d = \frac{1}{\alpha} \ln\left(\frac{F}{F_0}\right)
\]

(1)

where \(F_0\) is the ablation threshold and \(\alpha\) the absorption coefficient. These two ablation properties depend on the material, and differ across different polymers and the cornea [14].

Unless the ablation properties of the plastic model [15] are identical to that of corneal tissue, the ablation patterns obtained in plastic will differ from those expected in corneas. However, under certain assumptions [1], the results can be extrapolated to the cornea. Other important differences in ablation properties across materials and cornea include a hydration-dependent dynamic absorption coefficient [16–19] or shock wave dynamics on corneal tissue [20-21].

An important effect of the ablation, with potentially important consequences in the effective ablation pattern, is the plume of smoke ejected during the ablation [22], consisting of new chemical products in gas phase, and ablated material fragments, atoms, ions and molecules [23-24]. This plume of smoke can produce shielding (a loss in the effective energy by absorption or ablation, reflection or scattering) in the subsequent pulses [25–27], and has been related to the appearance of central islands [28-29]. This plume has potentially a larger effect on plastic models than on the cornea, as the evolution of the ablation plume over cornea is significantly faster than over PMMA [28].

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Dorronsoro et al. [1] have presented a method to estimate a correction factor for the geometrical laser efficiency losses (lower laser efficiency in the periphery of the cornea), based on the ablation of flat and spherical plastic surfaces. This correction factor can compensate for the discrepancies from the intended corneal ablation pattern and avoid the increased asphericity (and increased spherical aberration) in non-corrected standard patterns. The method was applied to the study of the ablation profiles of different state-of-the art refractive surgery laser platforms on Filofocon A plastic model corneas [8]. In that study, the measured ablation patterns from one of the lasers showed a consistent asymmetry, which was hypothesized to originate from shielding effects due to insufficient air aspiration.

Inspired by the results of that previous study, we undertook a systematic investigation of the relationship between air flow and shielding in plastic models. The question is particularly important in the calibration and evaluation of refractive surgery excimer lasers using plastic models, as the measured ablation profiles in plastic could be affected by inadequate removal of the smoke plume. Furthermore, as current lasers are increasing their repetition rates, and therefore reducing the time between single laser pulses, the shielding effects could become of even more relevance [30-31]. In this work, flat plastic samples of two different materials (Filofocon A and PMMA) were ablated with a state-of-the-art excimer refractive surgery laser. An air flow generator next to the samples provided air fluxes of different speeds. The ablated samples were measured with a non-contact optical profilometer, and the resultant ablation patterns, ablation profiles, central ablation depths, as well as ablation stability were compared across air flow and material.

2. Methods

2.1. Laser system

An Allegretto Eye-Q excimer flying spot laser platform (Wavelight GmbH, Erlangen, Germany) was used in this study. Clinically, this laser platform has the ability to treat myopia, hyperopia, astigmatism, as well as higher order aberrations, by means of different customized ablation algorithms [32–38]. Relevant nominal specifications of the laser include: 193 nm wavelength, 0.95-mm laser spot diameter with a Gaussian fluence profile, 400 mJ/cm² peak fluence, 200 mJ/cm² average fluence, and 400 Hz repetition rate.

This laser system is equipped with a removable test head to provide mechanical support to the ablation of plastic samples used for calibration and evaluation of the system. The test head also provides a laminar air flow of fixed flow speed at the sample plane to provide additional removal of smoke and debris generated during the ablation of plastic samples. For the current study, the test head was replaced by a custom mechanical support for the samples, provided with an air flow generator of variable flow speed.

The laser was calibrated at the beginning of the experiments. The pulse energy was set to 1.25 mJ, the same value used in corneal ablations. The laser calibration state was checked periodically, but no changes or adjustments were needed.

A Wavefront Optimized algorithm [39] was used, with an optical zone diameter of 6.5 mm, and an ablation zone diameter of 7.1 mm. Two spherical refractive corrections were applied, corresponding to a refractive error of −6 and −9 diopters (D) in the cornea, with nominal central ablation depths of 92 and 142 μm in corneal tissue. All the parameters were kept identical for each pair of ablations corresponding to the two different materials, and therefore similar laser pulse shape, pulse location sequence, number of pulses and energy per pulse was applied to both materials.

2.2. Samples and support

The flat samples used for ablation were made of two different materials: Filofocon A and PMMA. The flat Filofocon A cylindrical samples (14-mm diameter, 24.6-mm high) were similar to those described in a previous work [8]. They were lathe-cut and polished at the Institute of Optics (CSIC, Madrid). The PMMA flat samples (28-mm diameter, 4-mm height) were provided by the manufacturer, and were equal to those used for calibration of Wavelight
lasers [40–42]. They were manufactured by molding with no additional surface treatment. The flatness of the samples was checked before the ablations. In both types of samples, the preoperative peak to peak deviation from an ideal plane, inside the ablated zone, was below 1 μm.

The samples were placed on a tip and tilt platform, to facilitate alignment. As both types of samples had different dimensions, a mechanical adapter was designed and manufactured to allow a quick exchange between samples, while the position of the sample (surfaces center and height) was maintained.

2.3. Air flow

The clinical air aspiration integrated in the clinical laser platform and routinely used during clinical photoablations is designed to absorb the corneal ablation products while protecting the laser lenses. It removes the air at a higher plane, but not at the sample plane. An additional air flow generator was placed in the experimental setup to achieve controlled air flow conditions during ablation of the samples. This additional air flow generator consisted of a fan (50-mm diameter) to which a cylindrical tube (40-mm diameter, 100-mm length) was mounted. The fan speed could be adjusted with a conventional lab voltage source from 0 to 15 V.

The tube was located parallel to the sample surface. The surface to ablate was 18 mm below the tube center, so the sample was inside the air flow provided by the tube.

Prior to the experiments, the air flow provided by the air flow generator was calibrated. The clinical air aspiration was blocked during the calibration. The air flow was measured with an anemometer (RS 180-7111, RS Components S.P.A., Italy) at the sample plane at different voltages (1-V steps between 4-12 V) to obtain a regression curve between voltage and air flow.

Four air flow conditions were tested in the final ablation experiments: (af0), no air flow and no clinical air aspiration; (afc), clinical air aspiration and no air flow; (af1), 1.15 m/s air flow and no clinical air aspiration; and (af2), 2.3 m/s air flow and no clinical air aspiration. Although in the (afc) condition the air aspiration flow was 3.4 m/s at the clinical aspiration tube tip, no air flow was measurable with the anemometer at the eye/sample plane.

2.4. Experimental procedures

Pairs of PMMA and Filofocon A samples were ablated under identical conditions. For each condition, ablations were first made on PMMA followed by Filofocon A samples. Centration was referred either to the center of the mechanical supporting piece (PMMA samples) or an artificial iris (Filofocon A samples). The anterior surface of each sample was then positioned in the treatment plane using the pair of focusing lasers of the Allegretto Eye-Q. To achieve alignment, the tip and tilt platform was used to place the sample perpendicular to the optical axis of the instrument, using the reference laser provided by the Allegretto Eye-Q (low energy visible laser beam collinear with the ablating laser beam). The coincidence of the incoming and reflected beams was evaluated at a screen placed 200 mm above the sample. The alignment error with this procedure was estimated to be below 0.5 deg. The orientation of the sample during the ablation was marked with a line at the edge of the sample. Between ablations, centration was usually maintained, although focus and alignment were adjusted when needed.

A total of 42 samples were ablated. Ablations of −9 and −6 D were performed with the (af0), (afc) and (af1) air flow conditions in both materials. Additionally, −9 D ablations were performed with (af2) in both materials. All the ablations were performed on the same day. The room temperature was 24 degrees. The sample was isolated from the effect of the air conditioning fan of the operating room.
2.5. Shape measurements

The shape of the anterior surface of the samples was measured pre- and post-operatively with an optical profilometric microscope (PLμ, Sensofar, Barcelona, Spain) mounted on a vibration-isolated table. Details on the instrument [43] and on the measurement modes (surface topographies and cross-sectional profiles) can be found elsewhere [8]. All the post-operative surfaces were measured with the reference mark in the same orientation, to preserve the information on the relative direction between the sample and the air flow.

2.6. Data Analysis

For each profile, the central ablation depth was calculated averaging across the central 0.2 mm, for robustness. These central ablation depths were averaged across vertical and horizontal sections, and across repeated ablations under similar conditions. All the profiles were then normalized with respect to the maximum ablation depth measured (af2 condition in Filofocon A). The variability of the ablation was estimated from the standard deviation of the ablation depth at each location of the samples, for repeated ablations under similar conditions.

From the central ablation depths, we can obtain a quantitative estimation of the shielding effects at the center of the ablation. We define a Shielding Factor as:

\[ S = 1 - \frac{d_s}{d} \]

where \( d_s \) is the ablation depth with shielding, and \( d \) the ablation depth without shielding. The Shielding Factor \( S \) represents the ratio of reduction of the ablation depth due to shielding effects, and takes values between 0 (no shielding) and 1 (complete shielding, no ablation at all).

3. Results

3.1. Smoke column and air flow

Figure 1 shows the behavior of the smoke plume during the ablation, for different air flow conditions. Figure 1a (and Media 1), corresponding to the af0 condition, shows that a dense smoke column is generated during the ablation, when air flow and clinical aspiration are switched off. With clinical air aspiration alone, afc (Fig. 1b and Media 2), we observe that, as the smoke column leaves the sample, it tends to deviate towards the aspiration system. This aspiration causes instabilities of the smoke density above the sample. The smoke appears barely visible when the air flow generator is present (af1 or af2) (Fig. 1c and Media 3).

![Fig. 1. Single-frame excerpts from video recordings of the ablations of Filofocon A samples with three different air flow conditions: a) af0, no air flow, no clinical aspiration (Media 1); b) afc, no air flow, clinical aspiration (Media 2), c) af1, 1.15 m/s air flow, no clinical aspiration (Media 3).](image)
The sequences of the experiment depicted in Fig. 1 correspond to Filofocon A. The ablations with PMMA show a similar behavior, although PMMA produces a denser smoke column from the sample during the ablation. After the ablations with the air flow generator, a region of deposited material was found on the sample surface outside the ablation zone, in higher amounts with PMMA than with Filofocon A (Fig. 2).

These deposits correspond to ablation debris or particles dragged by the air flow, as they are released on the opposite direction of the incoming air flow, and their position was changed when the air flow direction was changed.

![Ablated PMMA sample in holder. A region of deposited material is visible in the upper part of the sample, outside the ablation zone. This region corresponds to the ablation debris dragged by the air flow, as it always appears in the opposite direction of the air flow generator.](image)

### 3.2. Measured ablation profiles

Figure 3 shows examples of raw topographies of the ablated samples, for each air-flow condition. Data are for Filofocon A samples. Higher air-flow produces deeper ablations. Reduction or elimination of air-flow reduces significantly the achieved ablation-depth. The topographies show that a clear asymmetry in the ablation pattern is produced when using the clinical air flow \( (\text{afc}) \) condition, as already reported in a previous study \[8\]. Increasing air-flow produces deeper ablations and eliminates the asymmetry of the ablation pattern.

![Figure 3. Topographies corresponding to ~9 D ablations in Filofocon A, for the different air flow conditions. The depth scale (colorbar) is the same for the four ablation measurements, and has been normalized to the maximum depth (center of the \( \text{af2} \) condition).](image)

Figure 4 shows a contour plot of each topography and, marked with arrows, the direction of the air flow for each condition. An outlier point has been removed in the topography of the \( \text{af0} \) condition. The \( \text{afc} \) topography has been rotated to make the direction of the air aspiration coincide with the direction of the air flow in \( \text{af1} \) and \( \text{af2} \). Clear irregularities appear in the ablation pattern achieved in the no-air-flow condition \( (\text{af0}) \). With the clinical air aspiration \( (\text{afc}) \) some irregularities are also noticeable, as well as an apparent asymmetry that occurs in the opposite direction of the air aspiration. With air flow \( (\text{af1} \text{ and } \text{af2}) \) the ablations are deeper and more symmetrical as the flow rate increases. Similar results were obtained in the PMMA ablated samples.

At a small-scale, the slight differences found at 45 and 135 deg with respect to 0 and 90 deg in the \( \text{af1} \text{ and } \text{af2} \) conditions can be attributed to the slightly higher accuracy of the
profilometer in the horizontal and vertical direction, as the structured-illumination fringe pattern projected by the non-contact profilometry is horizontal.

![Images of profilometer readings](image)

Fig. 4. Contour maps of the ablations shown in Fig. 3. The arrows represent the direction of the air aspiration (af0) or blown air flow (af1 and af2). Each contour line represents a 1/30 step of the maximum ablation depth with the af2 condition. The lines are gray-scale coded from no ablation (white) to deepest ablation (black).

Figure 5 shows horizontal depth profiles for the –9 D ablations for the different air-flow conditions, both for Filofocon A (Fig. 5a) and PMMA (Fig. 5b). The lateral position was adjusted using the ablation borders. The data are normalized to the maximum ablation depth in Filofocon A. Apart from the deeper ablations obtained as air flow increases, already observed in Fig. 1, we appreciate a much higher repeatability across equivalent profiles as air flow increases.

![Normalized ablation profiles](image)

Fig. 5. Example of normalized ablation profiles (horizontal direction). For different air flow conditions and materials, for –9D ablations. a) Filofocon A, b) PMMA.
3.3. Ablation pattern variability

Figure 6 shows the local height standard deviation along the horizontal profile (i.e. the standard deviation across equivalent profiles at each point of the profile), for −9 D. Figure 6a shows the local standard deviation for Filofocon A, and Fig. 6b for PMMA.

![Figure 6. Local standard deviations along the horizontal profiles shown in Fig. 1, across the three repetitions of each condition, for −9 D ablations. a) Filofocon A. b) PMMA.](image)

For the $af_1$ and $af_2$ conditions (blue and green curves), the standard deviation inside the ablation zone is only slightly larger than the standard deviation outside the ablation zone (roughness of the sample), indicating small surface shape changes across repetitions. The local standard deviation is increased for the low air flow conditions (no air flow, $af_0$, in red, and clinical air flow, $afc$, in orange). Also for the low air flow conditions, the local standard deviations of the ablated Filofocon A samples are higher than in PMMA. Similar results are obtained for the −6 D ablations.

![Figure 7. Relative standard deviations of the profiles, for the three repetitions of each condition. Data were computed inside the Optical Zone (central 6.5 mm of the ablation) and are relative (in %) to the maximum ablation depth, for each material.](image)

Figure 7 shows the standard deviations of the ablated surfaces for each air flow condition and material. These values were computed from the average values of the local standard deviations of the profiles of Fig. 6 within the optical zone (central 6.5 mm of each profile), for horizontal and vertical profiles. For each material the standard deviation values have been normalized to the maximum ablation depth, for comparison. With maximum air flow ($af_2$, 2.3 m/s) the standard deviation is below 1% of the maximum ablation depth. With no air flow ($af_0$), the standard deviation is 2.3% the maximum ablation depth.

3.4. Ablation depth and Shielding factor

Figure 8 shows the central ablation depths for each condition and material. We found highly statistically significant differences (p<0.001, t-test) in central depth across conditions, and between ablated PMMA and Filofocon A samples for the same ablation condition.
Consistently deeper ablations were obtained in Filofocon A, as expected due to lower Ablation threshold $F_{th}$ and absorption coefficient in Filofocon A than PMMA. Details of these material ablation properties can be found in a previous publication [14].

Fig. 8. Normalized central ablation depth for each air flow condition and material. Data obtained from a central region of 0.2 mm. Data are the average across repetitions and horizontal and vertical profiles.

For each material, the central ablation depth is highest for the highest air flow, which was set as the reference condition ($d$ in Eq. (2)). As the air flow decreases, the central depth also decreases.

Figure 9 shows the measured Shielding Factors $S$ (orange columns) for the different air flow conditions and both materials. The upper parts of the columns (brown and gray) represent the normalized ablation depth (data from Fig. 8 relative to the depth in the $a_f2$ condition) for each material.

Fig. 9. Shielding factors (in orange) for the different air flow conditions and both materials. The orange columns represent the relative ablation depth reduction attributed to shielding effects.

The measured shielding factor $S$ decreases as air flow increases and it is very similar for both materials in all conditions (except for the clinical air flow in Filofocon A, where the shielding is slightly lower). The shielding factor ranges from 0.42 (corresponding to the no air flow condition, $a_f0$) to 0.05 with 1.15 m/s air flow ($a_f1$). The Shielding Factor is 0.3 with the clinical air aspiration ($a_fc$).

4. Discussion

The use of artificial models in plastic has been identified as a tool to calibrate on polymers the ablation patterns used in corneal refractive surgery ablation patterns and to potentially improve the predictability and optical outcomes of refractive surgery procedures [1, 7-8].
PMMA has been widely used as a model, although recently Filofocon A has been proposed as an advantageous alternative to PMMA. In a previous study, Dorronsoro et al. [8] evaluated the ablation of plastic samples with different laser platforms. Important deviations from rotational symmetry were systematically found in the ablation patterns with the Wavelight Allegretto Eye-Q model laser. The authors hypothesized that shielding effects due to insufficient air aspiration might be the cause of these asymmetries. The current study was designed to study the influence of air flow in the ablation of plastic surfaces used for calibration and evaluation of refractive surgery profiles. A deeper understanding of the influence of air flow on shielding is critical to appropriately transfer the programmed patterns to the cornea.

We observed that shielding effects occur both in Filofocon A and PMMA. The ejected material (plume) interacts with the incoming laser radiation (subsequent pulses), and produces shielding and scattering which result in reduced ablation efficiency and spatiotemporal variations of the energy. We report changes across the different air flow conditions in the shape of the achieved profiles (Figs. 3, 4 and 5), in the repeatability of the profile (Figs. 6 and 7) and in the central ablation depth (Fig. 8). Increasing the air flow reduces shielding (Fig. 9), and increases the symmetry of the profile, the ablation depth and the repeatability of the profile. These shielding effects are well correlated with qualitative observations during the measurements. The videos (Fig. 1, Media 1 and Media 2) suggest that the ablating laser beam interacts with the existing plume of smoke, not only at the surface plane, but well above it. The smoke column has visible micro turbulences when the pulse train is on (Media 1), but has a laminar regime when no pulses are present (as seen at the end of the ablation process). The micro turbulences in the plume of smoke can be attributed to particle de-fragmentation and material decomposition. Other presumable interactions between the ablating laser and the plume include absorption, reflection and diffraction, which are likely to produce a reduced effective fluence, as well as irregularities in the surface of the ablated sample. Additional effects which may have an impact on the irregularities of the sample include shielding by redeposited debris material and the reduction of the effective diameter of the laser spot.

A simple model for shielding can both explain the experimental results and provide predictions on the optimal air-flow conditions to prevent shielding effects in plastic model ablations, and therefore improve the calibration of refractive surgery lasers and their predictability, based on the specifications of the laser. We can define a Shielding Transmission Coefficient \( T_S \) that accounts for the shielding by the ejected particles in the ablation process of material:

\[
T_S = F_S / F_0 ,
\]

where \( F_0 \) is the nominal fluence of the laser and \( F_S \) the effective fluence after shielding.

We can relate the Shielding Transmission Coefficient \( T_S \) with the Shielding factor \( S \), by operating with Eq. (1) (with \( F = F_S \)), Eq. (2) and Eq. (3):

\[
S = -(\ln T_S / \ln (F_0 / F_0)) ,
\]

or alternatively

\[
T_S = (F_0 / F_0)^{-S} .
\]

We can assume that the Shielding Transmission Coefficient \( T_S \) is proportional to the number of particles in suspension at the moment of the ablation, and that the ejected particles are dragged by the air flow, which, to some extent, would clean the ablation area.

This simple model can predict the ablation depth at the center of the ablation in presence of shielding. For each air flow speed \( (af) \) we can define a Particle Removal Time \( (t) \) as the maximum time needed for the particles to leave the center of the ablation, i.e., to cover the distance between the center of the ablation and the periphery \( (t = R/af) \), where \( R \) is the radius of the ablation zone. The Threshold Particle Removal Time \( (t_0) \), i.e. the time between consecutive pulses, can be obtained from the laser repetition rate. This simple model can also
be used to estimate the shielding factor $S$, under the following assumptions: (1) The Shielding Transmission Coefficient $T_S$ can be considered to be inversely proportional to the percentage of remaining particles; (2) the percentage of remaining particles can be considered in turn proportional to the percentage time exceeding $t_{th}$.

For our experimental parameters ($R=3.55$ mm; and laser repetition rate = 400 Hz), $t = 1.54$ ms, 3.09 ms and $\infty$, for $af2$, $af1$ and $af0$, respectively, and $t_{th} = 2.5$ ms between consecutive pulses. This indicates that a minimum air flow ($af_{th}=R/t_{th}$) of at least 1.42 m/s is needed in order to avoid shielding effects at the center of the ablation. This prediction is consistent with the initial assumption of using $af2$ as a reference condition (as no shielding effects are predicted for $af2 = 2.3$ m/s), as well as, with the experimental findings of the study (shielding effects occurring at $af1 = 1.15$ m/s and below). For $af1$ the removal time ($t = 3.09$ ms) is higher than the threshold removal time ($t_{th} = 2.5$ ms).

For $af0$ (0 m/s), where 100% of the shielding particles are present, we obtained the full Shielding Transmission Coefficient $T_S\{af0, FILO\} = 0.715$ (for Filofocon A), using Eq. (5). For the threshold air flow $af_{th}$ (1.42 m/s), $T_S\{af_{th}, FILO\}$ is 1, by definition. For $af1$, the model estimates that 19% of the particles are present, which yields a predicted Shielding Transmission Coefficient $T^S\{af1, FILO\} = 0.946$, a predicted Shielding Factor of $S^P\{af1, FILO\} = 0.070$ and a predicted normalized ablation depth of $d^S\{af1, FILO\} = 93\%$. The predictions of the model agree well with the experimental data for this condition: $S_{af1, FILO}=0.051$ (Fig. 8) and normalized $d^S_{af1, FILO} = 95\%$ (Fig. 5).

For PMMA, the estimated full Shielding Transmission Coefficient is lower than for Filofocon A ($T_S\{af0, PMMA\} = 0.630$), consistent with a higher absolute number of particles ejected and/or a higher interaction with the laser. For an air flow of 1.15 m/s ($af1$) we predicted a similar shielding transmission coefficient ($T^P\{af1, PMMA\} = 0.930$) than for PMMA, consistent with experimental results. The predicted Shielding Factor ($S^P\{af1, PMMA\} = 0.067$) and normalized central ablation depth ($d^P\{af1, PMMA\} = 82\%$), were also close to those found experimentally in PMMA ($S_{af1, PMMA} = 0.049$; normalized $d^S_{af1, PMMA} = 83\%$).

Despite its simplicity, the model is able to predict the effect of shielding on central ablation depth as a function of the air flow speed, within an error of 2% (less than 2 μm in absolute units). This simple model, based on an interpolation of the percentage of shielding particles and the application of theBeer Lambert Law, provides significant better estimates of the achieved ablation depth than a direct interpolation in the Shielding Factors or ablation depths. The model allows explaining consequences of shielding by the smoke column on ablation, and extrapolating the experimental factors to other laser fluences, laser repetition rates and materials.

Although the analysis has been performed for the center of the ablation, for simplicity, the conclusions can be extended to the entire ablation, due to the linearity of the equations. An increase in the ablation zone area proportionally increases the Particle Removal Time, and therefore the threshold air flow, for the same laser repetition rate.

Similarly, the model can explain the threshold air flow for different repetition rates. Higher repetition rate lasers are more prompt to produce shielding problems than lower repetition rate lasers. From our model, an increase of 100 Hz in the repetition rate causes an increase of 0.71 m/s in the threshold air flow (for an entire ablation zone of 7.1 mm). For 1000 Hz repetition rate a high air-flow speed will be needed (7.1 m/s).

For a given air flow, this model predicts a very similar Shielding factor for both materials (Fig. 10). The differences in the predicted shielding effects (Fig. 10) across materials are below 0.5%.

However, the model does not consider physical facts which may be responsible for the asymmetries of the laser profile. These include particle inertia and particle ejection directions, nozzle distance, pulse density distribution and overlapping, time sequence of the laser pulse density distribution or particle cumulative effect. Furthermore, the model is limited to the prediction of the central ablation depth. A sophistication of the model, considering the local variations of the parameters, could lead to better predictions across the entire profile and local variations of the ablation depth, and consequently asymmetries in the ablation pattern.
The experimental results and predictions from the model indicate that the clinical air flow (afc) is insufficient to remove the ejected particles during the ablation of the model plastic materials, and results in a significant shielding factor. From the curves in Fig. 10 we can obtain an effective clinical air flow for the afc condition, i.e., the air flow corresponding to the Shielding factor measured with the afc condition (shown with crosses). The estimated effective clinical air flow is higher with Filofocon A (0.5 m/s) than with PMMA (0.33 m/s). These values of effective clinical air flow do not attempt to describe the air flow at the sample plane (which is known to be negligible) nor the real air aspiration flow (much higher), but the overall effect in terms of particle removal. As the real air aspiration flow is similar in both materials, this higher effective clinical air flow in Filofocon A indicates a higher performance of the system in the aspiration of the plume of smoke produced in the Filofocon A ablations, probably due to the lower amount of smoke produced in the ablation of this material.

The ablations with clinical air flow show significant by asymmetries. The shielding is observed in particular in the direction of the aspiration (less depth and more variability in the right side of the afc curves in Fig. 4). This can be explained considering that as the air flow drags the particles, more shielding particles are present in the direction of the air aspiration.

Despite the higher effective clinical air flow, the asymmetries in the ablation pattern appear more marked in Filofocon A (Figs. 3 and 4) than in PMMA. In addition, the local standard deviation of the ablation pattern (Fig. 6) increases in the region toward which the particles are expected to have been dragged (right side of the curve), and in fact this deviation is even larger at clinical airflow than at no air flow, particularly in Filofocon A. Factors affecting the local variability of the pattern and the differences across materials in this effect include: differences in the smoke ejection pattern density and weight of the ejected patterns, among others.

In this study, the air flow was generated by a blowing fan, for convenience. Implementations based on blowing or aspiration should be equivalent in terms of shielding prevention, as far as the air flow at the sample plane (the only relevant parameter) is the same.

The conclusions obtained are limited to ablations in plastic, of importance for research studies and calibration procedures, but should not be directly extrapolated to cornea. The clinical air aspiration systems have been designed considering corneal material, which has lower particle emission and faster plume dynamics [28] and different ablation properties. Additional precautions need to be taken when ablating plastic materials, especially at high repetition rates.

5. Conclusions

Shielding effects are found with the typical air-flow of a clinical platform in the ablation of plastic models commonly used in refractive surgery laser calibrations (Filofocon A and

![Fig. 10. Predicted Shielding factors vs air flow speed for Filofocon A and PMMA for a repetition rate of 400 Hz. The effective clinical air flow (air flow corresponding to the Shielding factor found with the afc condition) is also shown (crosses).](image.png)
PMMA). The effects of insufficient air-flow speed (and therefore insufficient air-debris removal) are: a decrease in the ablation depth; irregularities in the ablation pattern; and increased variability of the ablation.

Increasing the air-flow increases the ablation depth and eliminates the irregularity of the system. For a −9 D ablation, increasing the air-flow from 0 to 2.3 m/s increased central ablation depth by 40% in PMMA and Filofocen, using a last-generation Wavelight laser.

The shielding effects for PMMA and Filofocen are similar despite the differences in full-shielding transmission coefficient (number of particles ejected and/or optical behavior of these particles) and the differences in the ablation properties of the materials.

A simple model based on an estimation of the shielding produced by the ejected debris is able to predict the measured central ablation depths within 2 μm of accuracy.

The clinical air aspiration (at least with this particular laser) is clearly insufficient for ablation experiments on plastic, which can limit the performance of the calibrations. This confirms the need for an air flow generator (as present in this laser) for the ablation of plastic samples. Air flow is a key factor in laser evaluation with plastic models.

Air flow needs to be tuned to the repetition rate of the laser, in order to avoid shielding in the ablation. In particular, an air flow over 3 m/s is recommended for a 400 Hz repetition rate laser.