Suitability of Filofocon A and PMMA for experimental models in excimer laser ablation refractive surgery

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Abstract: Experimental corneal models in plastic (in PMMA, and more recently in Filofocon A, a contact lens material) have been proposed recently to overcome some of the limitations of the theoretical approaches aiming at improving the predictability of corneal reshaping by laser ablation. These models have also been proposed for accurate assessment of corneal laser ablation patterns. In this study Filofocon A and PMMA optical and ablation properties were studied using an experimental excimer laser set-up. The effective absorption coefficient and the ablation thresholds of these materials were obtained as a function of the number of pulses. Both materials follow a Beer-Lambert law in the range of fluences used in refractive surgery, and the number of incubation pulses is less than 4 (PMMA) and 2 (Filofocon A) above 140 mJ/cm². We found that above 40 pulses for Filofocon A and 70 pulses for PMMA, ablation threshold and effective absorption coefficients can be considered constant ($F_{th} = 90$ mJ/cm² and $\alpha_{eff} = 36000$ cm⁻¹ for Filofocon A, and $F_{th} = 67$ mJ/cm² and $\alpha_{eff} = 52000$ cm⁻¹ for PMMA, respectively). The absence of ablation artifacts (central islands), a lower number of incubation pulses, a lower pulse-number dependence of the ablation threshold, and a good correspondence between $\alpha_{eff}$ and the absorption coefficient $\alpha$ estimated from spectroscopic measurements make Filofocon A a more appropriate material than PMMA for experimental models in refractive surgery and for calibration of clinical lasers.

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References and links

28. Charles Campbell, 2908 Elmwood Court, Berkeley, California 94705, USA (personal communication 2007).
1. Introduction

Refractive surgery by excimer laser ablation is successful at eliminating refractive errors of the eye, reshaping the patient’s cornea to alter its refractive power. However, many studies [1-4] have reported the induction of high order aberrations. Post-operative wave aberration maps show large amounts of spherical aberration compared with pre-operative maps, and these induced aberrations have been shown to have a noticeable impact on vision [5].

The new generation of lasers for refractive surgery aims at optimising or even customizing the ablation pattern applied to the cornea. These systems are usually provided with precise control of the energy delivery at each point, thanks to flying spot technology, pupil tracking, and fast repetition rates. Achieving the desired post-operative corneal shape (and post-operative ocular aberrations ideally reduced from pre-operative values) using customized ablation algorithms requires an adequate profile design [6-8]. However, understanding the physical factors affecting the effective transfer of the energy to the cornea is equally critical [9].

Marcos et al. [10] and others [11-13] have demonstrated that the theoretical ablation patterns by themselves are not responsible for the increase in spherical aberration found clinically. Mrochen et al. [14] and Jimenez et al. [15-17] have studied analytically potential physical causes for the energy loss, which varies as the laser spot moves from the center to the periphery of the cornea and depends on reflection, changes in spot shape, polarization, pulse overlap, etc. The geometrical dependence of the effective laser energy results in a variation of the applied ablation pattern with respect to the theoretical ablation pattern, and consequently in a different post-operative corneal shape than expected. Recent studies by Arba-Mosquera and de Ortueta [18], and Kwon et al. [19] describe sophisticated numerical models that take into account these effects, with less approximations, and considering the temporal evolution of the process.

All theoretical and computational studies rely on the knowledge of the ablation pattern, which is proprietary in most cases. In addition, theoretical equations are subject to approximations. Corneal biomechanical effects are other potential causes for the inaccurate prediction of the corneal postoperative shape. In any case, although the biomechanical response of the cornea after the tissue removal and the wound healing may undermine the predictability and stability of refractive surgery, these are added effects to physical factors [20].
Experimental corneal models in plastic have been recently proposed [21] to overcome some of the limitations of the theoretical approaches. In corneal plastic models physical effects can be tested purely, isolated from biomechanics. The energy pattern and the laser efficiency effects can be directly measured [22], and only some assumptions and approximations are needed to extrapolate the results from the plastic material to the cornea. Experimental models in plastic have also proven useful to validate computational models [18, 19] for the cornea.

Corneal laser ablation models have not only attracted the interest of the scientific community, but of normalization agencies. For instance, the ANSI Z80.11-2007 norm requires the ablation of plastic plates for calibration (energy adjustment and laser profile testing) of corneal refractive surgery [23]. The standard also recommends the use of plastic corneal models for in vitro testing of the ablation characteristics of the laser systems and to validate experimentally mathematical models, simulations, and ablation algorithms. The Ophthalmic Devices Panel of the Food and Drug Administration (FDA) is currently evaluating the ANSI Z80.11 Standard for recognition [24], and the FDA section on Medical Devices has pointed out the necessity to develop accurate calibration techniques to assure that the achieved corneal ablation depth map equals the one intended through plastic corneal models [25-27]. Laser refractive companies have identified Filofocon A, a material used in the contact lens industry, as a material potentially suitable for refractive surgery models, and this material holds promise in the community to establish itself as standard for calibrations of laser ablation profiles [28]. PMMA has already been used to test laser ablation properties of refractive surgery laser platforms [29]. In a recent publication, we used PMMA corneal models to quantify the geometrically-induced laser efficiency losses and its effects on standard ablation profiles [22]. Filofocon A is a recent material used in artificial eyes and its laser ablation properties have never been described.

All models of excimer laser corneal refractive surgery (experimental, analytical or numerical) rely on the validity of the assumption of energy deposition occurring in a well-defined layer which is then removed by ablation. For absorbing media, which is the case for cornea, PMMA and Filofocon A at the processing wavelength (193nm), the intensity $I$ of the electromagnetic wave propagating into the material ($z$-direction) decreases exponentially following the Beer-Lambert expression, according to

$$I(z) = I_0 \cdot e^{-\alpha z},$$  \hspace{1cm} (1)

with

$$\alpha = \frac{4\pi \cdot k}{\lambda},$$  \hspace{1cm} (2)

The inverse of the absorption coefficient $\alpha$ is the optical penetration depth, i.e. the depth in which the major fraction of the energy is deposited ($1/e$ of the total amount absorbed). $k$ is the extinction coefficient. As a consequence of Eq. (1), the etch rate $d$ (etch depth/pulse) is expected to relate linearly to the natural logarithm of the laser fluence $F$ according to

$$d = \frac{1}{\alpha} \ln\left(\frac{F}{F_{th}}\right),$$  \hspace{1cm} (3)

where $F_{th}$ is the ablation threshold. In PMMA, the Beer-Lambert behavior is found only to hold under two restrictions [30, 31]. First, it is only valid for a certain range of fluence and, second, the absorption coefficient $\alpha$ estimated from low intensity spectroscopic measurements has to be replaced by an effective absorption coefficient $\alpha_{eff}$ estimated from the slope of the representation of Eq. (3), in terms of $d$ vs. $\ln(F)$. Evidence has been found that this change in absorption coefficient in the high intensity regime occurs within a single pulse for PMMA [30, 32]. While some authors consider the assumptions of Eq. (3) to be valid for corneal tissue ablation [33], recent studies also suggest a dynamic absorption coefficient in the cornea [15, 34], changing under excimer laser ablative conditions. These restrictions have recently been
proposed as a possible limitation to the suitability of PMMA for experimental models [35, 36]. In comparison, no information is available about the ablation properties of Filofocon A.

In this study we will investigate the suitability of this new material (Filofocon A) for experimental laser ablation refractive surgery models, and measure its excimer laser ablation properties. In particular, we will study the relationship between number of pulses and ablation depth (for a range of fluences typical of refractive surgery laser platforms), incubation effects, ablation threshold and absorption coefficient of Filofocon A, in comparison to PMMA. Knowledge of these properties is essential to characterize the ablation patterns on plastic plates, to calibrate the lasers, to measure laser efficiency correction factors, and to extrapolate those results to the cornea.

2. Methods

Samples of Filofocon A were ablated using an excimer laser set-up. Ablation depths were measured using microscope interferometry to assess ablation rates as a function of number of pulses and fluence, the effective absorption coefficients and the ablation thresholds. Ellipsometry was used to measure the refractive index, \(n\), and the extinction coefficient, \(k\). The procedure was also followed for PMMA, a material previously used in refractive surgery ablation models.

2.1 Samples

Filofocon A (hydro-2) is a poly-fluoro-silicone-acrylate material (Innovision, Inc., Omaha, NE), used to manufacture rigid gas permeable contact lenses. This copolymer has a soft hydrophilic outer skin when immersed in water or tear film.

PMMA (acrylic glass) is a widely used thermoplastic material. Pure PMMA is usually modified with varying amounts additives, to improve its specific properties for certain applications. Transparent extruded PMMA was used in these experiments (Horniplas, Vitoria-Gasteiz, Spain)

The samples for ablation were prepared as optically flat rectangular surfaces (10x20 mm), polished in a precision optics lathe. Two samples of each material (Filofocon A and PMMA), with 2 mm thickness, were used in the ablation experiments. For the measurement of optical properties, flat polished surfaces of 12-mm diameter, and 4-mm thickness (to avoid contributions from the back surface) were used.

2.2 Estimation of optical properties

Refractive index and extinction coefficients of Filofocon A and PMMA were obtained from spectrometric ellipsometry (WVASE J.A. Woollam, Lincoln, NE) under incident angles of 60, 65 and 70 degrees. Measurements were obtained with a xenon lamp in a wide wavelength range from 1000 to 270 nm, under ambient conditions. Measurements below 270 nm were noisy, due to the air absorption and reduced reflectivity of the material in UV, and therefore a Cauchy model extrapolation was used to retrieve \(n\) and \(k\) at 193 nm.

2.3 Laser ablation set-up

We used a laboratory Argon Fluoride (ArF) excimer laser (LPF200, Lambda Physik, Göttingen, Germany) delivering laser pulses (wavelength 193 nm, pulse duration 20 ns) to perform local ablation in Filofocon A and PMMA. The laser fluence was continuously adjusted by rotating a coated fused silica window inserted into the beam path, whose transmission depended strongly on the angle of incidence. The laser was operated at a repetition rate of 1 Hz in order to avoid remains of the expanding ablation plume to shield the sample from the next incident pulse.

The beam profile incident on the sample was designed to be top-hat to a very high purity by using an imaging setup. The latter consisted of a fused silica biconvex lens (focal length=90 mm at 193 nm) that imaged a 4.5-mm diameter circular aperture, inserted into the beam path 2 meters after the laser output, precisely onto the sample surface. The exact z-position of the sample (image plane of the circular aperture plane) was adjusted until no diffraction rings
surrounding ablation spot borders could be observed. Absolute energy measurements were performed using a calibrated energy detector (Gentec ED 100A).

The linearity of the ablation depth versus number of pulses was assessed by performing series of irradiations at constant fluence, varying the number of pulses (1 to 600 pulses). The measurements were repeated for different fluences (70 mJ/cm², 140 mJ/cm², 240 mJ/cm² and 400 mJ/cm²). The agreement with a Lambert-Beer law was assessed by performing series of irradiations with varying fluences (from 30 to 400 mJ/cm²) for three different number of pulses (1, 3 and 10).

2.4 Crater depth measurement by non-contact profilometry

An optical profilometric microscope (PLOμ, Sensofar, Barcelona, Spain), mounted on a vibration-isolated table, was used to measure the ablated samples. The interferometric mode of the microscope, with 20x and 50x interferometric objectives, was used to measure the ablation depth of each spot. A three dimensional topographic map of the spot was obtained. The instrument software for measurement analysis was used to remove tilt and to obtain the ablation depth from diametral profiles of the crater.

2.5 Data analysis

For each fluence series, ablation depths were plotted against the number of pulses. The regression coefficients of the linear fits to the data provided a numerical quantification of the linearity. The number of incubation pulses was also obtained from these fits as the number of pulses where the linear range reaches zero ablation depth.

For each series with equal number of pulses, the ablation rate (ablation depth per pulse) was plotted versus the natural logarithm of the fluence (see Eq. (3)) to obtain an experimental assessment of the Beer Lambert behavior of the material. The ablation threshold (Fth) is the fluence value separating two different regimes. Below Fth, the ablation rate is extremely low and flat. Above Fth, the ablation rate changes linearly with fluence. A linear fit to this second regime provided estimates of Fth (fluence value in which the ablation rate intersects zero) and of the effective absorption coefficient αeff (inverse of the slope of the fit).

2.6 Microscopic structure of the ablation by optical and atomic force microscopy

The bright field imaging mode of the profilometric microscope described in Section 2.4 was used to obtain optical micrographs (100x) of the ablated samples. High-resolution measurements of the surface topography in the crater centre were performed with the SmartSPM 1000 fast Atomic Force Microscope (AFM) developed by AIST-NT Co (Moscow, Russia).

3. Results

3.1 Optical properties

Table 1 shows the results of the refractive index (n) and extinction coefficient (k) measurements (and the corresponding absorption coefficient α, obtained from Eq. (2)) for both PMMA and Filofoco A, at the ablation wavelength (193 nm) and in the visible (540 nm).

Table 1. Refractive index (n) and extinction coefficient (k) measurements for both PMMA and Filofoco A. The corresponding absorption coefficient (α) is also shown. The values at 193 nm have less precision due to the extrapolation procedure to obtain them.

<table>
<thead>
<tr>
<th></th>
<th>n (193)</th>
<th>k (193)</th>
<th>α (193)</th>
<th>n (540)</th>
<th>k (540)</th>
<th>α (540)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMMA</td>
<td>1.64</td>
<td>0.0058</td>
<td>3800 cm⁻¹</td>
<td>1.495</td>
<td>0.00013</td>
<td>30.3 cm⁻¹</td>
</tr>
<tr>
<td>Filofoco A</td>
<td>1.62</td>
<td>0.0548</td>
<td>35700 cm⁻¹</td>
<td>1.456</td>
<td>0.00122</td>
<td>289.0 cm⁻¹</td>
</tr>
</tbody>
</table>
Previous works on PMMA optical properties reported a similar refractive index in the visible but a lower one (1.49) at 193 nm [37, 38]. The discrepancy is probably due to different additives or dopants in the material. The refractive index of Filofocon A measured in the visible matches the nominal data (1.46). To the authors’ knowledge, the optical properties of Filofocon A had never been reported at 193 nm, showing an extinction coefficient that is a factor of ten higher than that of PMMA.

3.2 Ablation profiles

Figure 1 shows typical profiles of the ablation craters found in PMMA and Filofocon A. PMMA ablation craters show central islands. The figure shows a typical example, although not all the central islands are so symmetric. These ablation artifacts are not present in Filofocon A.

The diameter of the spots is constant across materials and conditions (except for very high number of pulses at high fluence, in which thermal effects degrade the ablation walls). This diameter (422 microns) was used to calculate the irradiated area, and to set a multiplicative factor between the energy measurement of the detector and the fluence units (mJ/cm²). The flatness of the crater floor combined with the steep crater walls justifies this procedure to calculate the fluence values.

Fig. 1. Typical ablation profiles for PMMA and Filofocon A. These profiles are obtained as diametral cross-sections of the measured interferometric topography. The crater diameter is 422 microns. Crater depth was obtained comparing the depths measured in the proximity of the vertical arrows (see text). The profiles shown in the figure were obtained for 10 pulses at 240 mJ/cm².

Ablation depths were obtained for each ablated sample as in the example shown in Fig. 1. In order to avoid measurement artifacts near abrupt edges (oscillations in the interferogram) and central islands, depth measurements were performed in the proximity of the arrows of Fig. 1. For each crater topography, 2 to 10 depth measurements were obtained (depending on the roughness of the profile and the variability of the different diametral profiles) and the results averaged.

3.3 Linearity of ablation depth with pulse number

Figure 2(a) shows the ablation depth as a function of the number of pulses, for fixed fluences. For the sake of clarity ablations with very high number of pulses (>600) are not shown. The error bars are also not shown, as they are negligible in this scale (smaller than the symbol size). All points of each fluence series of ablation depth vs. number of pulses were fitted to a linear equation. There is a high linear relationship, both for PMMA and for Filofocon A. The regression coefficient (R²) was higher than 0.999 in all cases.

Figure 2(b) shows a closer view near the origin of the data shown in Fig. 2(a) (i.e. for few pulses and low ablation depths). For the higher fluences (400 and 240 mJ/cm²) the ablation depth is larger (deeper penetration) for Filofocon A than for PMMA, for the same number of pulses. For the lower fluences (140 and 70 mJ/cm²) this depends on the number of pulses.
Fig. 2. Ablation depth as a function of the number of pulses, for selected fluences. a) 0 to 200 pulses. b) 0 to 30 pulses (zoomed view).

Although the data are well fitted by linear regressions, the fitted lines only cross the origin at 400 mJ/cm². This is a result of incubation, which is quantified as the number of the first pulses that do not contribute to ablation depth.

Figure 3 shows the number of incubation pulses at all laser fluences, for the two materials. Filofcon A requires less incubation pulses than PMMA for a given fluence. The number of incubation pulses for Filofcon A is less than 2 above 140 mJ/cm² (4 pulses for PMMA). At 400 mJ/cm² the incubation is negligible (<0.01 pulses) for both materials.

3.4 Beer Lambert law: ablation threshold and absorption coefficient

The measured ablation depth was plotted against the natural logarithm of fluence (Beer Lambert law, Eq. (3)) for constant number of pulses. Figure 4 shows four representative examples of the experimental Beer Lambert plots for 10 and 1 pulses, for Filofcon A (Fig. 4(a)) and PMMA (Fig. 4(b)). The bends of the 1-pulse curves near the ablation threshold - green small symbols in Fig. 4(a) (Filofcon A) and orange small symbols in Fig. 4(b) (PMMA)- could be indicative of a low energy ablation regime occurring below 0.08 μm/pulse. For these points (with fluence lower than 200 mJ/cm² for Filofcon A and 275 mJ/cm² for PMMA) the incubation is close or above one pulse (see Fig. 3). For higher number of pulses, there is an abrupt change between the two regimes. The zero intercept of the linear fits provide the estimated ablation thresholds (F_th), and the slopes represent the effective
absorption coefficient $\alpha_{eff}$. As the number of pulses increases, both the ablation threshold and the slope decrease.

![Graph showing the number of incubation pulses as a function of laser fluence.](image)

Fig. 3. Number of incubation pulses as a function of laser fluence.

Figure 5 shows estimations of the ablation threshold (a) and the effective absorption coefficient (b). The large open symbols and solid line represent the values for 1, 3 and 10 pulses estimated directly from the depth/pulse vs ln(Fluence) functions, as in the examples shown in Fig. 4. The small solid symbols and dashed lines represent estimates for higher number pulses (as typically required in refractive laser ablations). These values were estimated from the linear fits in Fig. 2 to obtain interpolated depth/pulse data for an arbitrary number of pulses. This approach is valid given the linear behavior of the ablation depth as a function of the number of pulses (Fig. 2), and the Beer Lambert behavior of the ablation depth/pulse (Fig. 4). For all estimates, we checked that the number of pulses was above incubation and the fluence above the ablation threshold.

![Graph showing Beer Lambert behavior for Filofocon A and PMMA.](image)

Fig. 4. Beer Lambert behavior for Filofocon A (a) and PMMA (b) for 10 and 1 pulses. The linear fits provide the estimated ablation thresholds (zero intercepts of the line), and the effective absorption coefficient (slopes). The regression coefficient, $R^2$, and the ablation threshold, $F_{th}$, are shown for each linear fit.

For more than 10 pulses, the ablation threshold is lower in PMMA than in Filofocon A, while the effective absorption coefficient is higher for PMMA (Fig. 5). The ablation threshold decreases and the effective absorption coefficient increases with the number of pulses in both materials. Filofocon A ablation parameters approach asymptotic values faster (with lower number of pulses) than PMMA. With 40-pulse ablations, the values of ablation threshold and absorption coefficient for Filofocon A are less than 2% higher those at 100 pulses ($F_{th} = 90$ mJ/cm² and $\alpha_{eff} = 36000$ cm⁻¹). For PMMA, more than 70 pulses are required to reach 2% of the corresponding values at 100 pulses ($F_{th} = 66.79$ mJ/cm² and $\alpha_{eff} = 52000$ cm⁻¹).
3.5 Microscopic structure of the ablation craters

Visual inspection of the ablated samples shows that they are transparent up to 240 mJ/cm². At this and higher fluences a progressive darkening with the number of pulses appears in Filofocon A, but not in PMMA. High magnification (100x) reflection images confirm this darkening inside the ablation, revealing different structures in the ablated areas for Filofocon A and PMMA. These images also show a border around the ablated areas which is darker and broader in PMMA. Figure 6 shows high magnification optical images of unablated (left side of the image) and ablated (right) areas in Filofocon A (Fig. 6(a)) and PMMA (Fig. 6(b)), for 30 pulses and a fluence of 400 mJ/cm². The different structures inside the ablated area (right part of each image) are also present even with very low number of pulses. AFM images (shown in Fig. 7) confirm that these structures, close to the resolution limit of the camera of the optical microscope, are real physical structures and not optical effects such as aliasing.

4. Discussion

We have studied the excimer laser ablation of Filofocon A and PMMA materials, in the operational range of refractive surgery. These materials have been used in previous studies in experimental models of corneal refractive surgery [22, 29, 39], although their ablation properties were not well established.

Both materials show a large range of linear dependence between ablation depth and number of pulses, for different fluences (see Fig. 2) as well as between ablation rate (depth/pulse) and the logarithm of the fluence (Beer Lambert law, see Fig. 4). We have measured optical and ablation properties of these materials at 193 nm (including index of refraction and extinction coefficient, ablation threshold and effective absorption coefficient) and investigated the structural changes in the material following ablation. We have shown that incubation is not negligible for low fluences, and that the ablation parameters change with the number of pulses, more so in PMMA than in Filofocon A. All these findings are very relevant for the development of plastic models to assess refractive surgery ablation profiles and correction factors for ablation algorithms.

The results for PMMA can be compared with those in the literature in terms of ablation rate (obtained from the Beer Lambert parameters). At 500 mJ/cm², our estimates of ablation rate range between 0.39 and 0.42 microns/pulse (depending on the number of pulses) while Fisher and Hanh [9] reported 0.47 microns/pulse. Calculations from the data of Costela et al. yield an ablation rate of 0.45 [31], from Braren et al. [40] 0.47, and from Srinivasan [41] 0.43. Regarding incubation, Fisher and Hahn [9] reported the absence of appreciable incubation at 500 mJ/cm², and Pettit et al. [42] reported 18 pulses at 115 mJ/cm² and 2 pulses at 265. All these results are in good agreement with those reported in this study, specially given the...
diversity of experimental methods and ablation conditions of the different studies (i.e. fluence range), and the possible discrepancies due to different amounts and types of impurities of the polymer.

(a) Filofocon A         (b) PMMA

Fig. 6. High magnification (100x) optical micrographs of Filofocon A (a) and PMMA (b). In both pictures, the ablated zone (400 mJ/cm$^2$, 30 pulses) is on the right. The scale bar is 10 μm.

(a) Filofocon A       (b) PMMA

Fig. 7. AFM images of the ablated zone (400mJ/cm$^2$, 30 pulses) in Filofocon A (a) and PMMA (b). The scalebar is 2 μm. The peak to peak depth (color scale range) is 300 nm.

4.1 Measuring a laser ablation pattern: energy to depth relationship

Plastic plates (generally of PMMA) are typically used in clinics to calibrate the fluence of the refractive surgery excimer laser. Depth measurements are usually performed (using a gauge) in the center of the ablation only, and related to laser fluence. But new standards [23] recommend a profilometric evaluation of the entire ablated shape. Knowledge of the ablation rates of the plastic calibrating material is needed. Also, incubation and varying ablation properties with the number of pulses can affect the proportionality between the laser energy pattern and the measured ablation depth.

Clinical lasers differ in the energy profile of the laser spot. For flying spot laser systems, it is assumed that the achieved ablation depth at each point is proportional to the number of pulses. While this is only true for uniform spots (top-hat) profile, it is a good approximation also for other profiles (Gaussian or truncated Gaussian) provided that the density of pulses is high (with uniform overlapping) [43]. The fluence of operation also differs across lasers (typically between 120 mJ/cm$^2$ and 400 mJ/cm$^2$), and consequently the number of pulses to achieve a given correction in cornea is different (with higher fluence lasers using a lower number of pulses and lower fluence lasers using a higher number of pulses). Our results demonstrate that for plastic models (as others have also suggested in corneal tissue [15, 34]), the energy to depth relationship is potentially affected by incubation, proximity to the ablation threshold and variable ablation parameters, especially for low fluence lasers.

To explore these limitations in practice, it is necessary to refer the measured ablation properties to the characteristics of the refractive surgery lasers and of the clinical ablations (typical fluences and number of pulses). Previous profilometric measurements [39] of the ablation profiles (for -9 D) on Filofocon A plates performed with different clinical lasers allow us to estimate the number of laser pulses applied to each given location, using the Beer Lambert parameters obtained in the present study and the nominal peak fluences of the lasers. The refractive surgery lasers under consideration were Technolas 217 Z100, Bausch&Lomb, with a peak fluence of 120 mJ/cm$^2$ and Ladar Vision, Alcon, with a peak fluence of 400
mJ/cm². It was estimated that the laser produced 124 pulses at the edge of the 6.5-mm optical zone for the Technolas system (which is consistent with the total number of pulses of the ablation pattern, 4233, provided by the laser manufacturer) and 12 pulses for the Ladar Vision. These numbers of pulses are well above the number of incubation pulses for both materials determined in the present study (Fig. 3), for the corresponding laser fluences. These fluences are above the ablation thresholds for Filofocon A and PMMA (Fig. 5), indicating that the ablations are performed in the Beer Lambert linear region. The effect of the variation in the ablation properties can be quantified in terms of the different number of pulses estimated when considering the asymptotic (static) or the exact (variable) values for the ablation threshold and the effective absorption coefficient. In identical conditions, the data in this study predict differences in number of pulses (and therefore errors in depth estimation) up to three times larger in PMMA than Filofocon A. Nevertheless, for these particular lasers and ablation patterns, the effect of the variation of the ablation properties turns out to be lower than 1 pulse for both materials. One-pulse error (a fraction of a micron) is considered negligible in the design and characterization of ablation algorithms [23]. Therefore, in these two lasers and these two materials the energy pattern is expected to be correctly transferred to depths of ablation or, alternatively, the measured depth pattern reflects correctly the laser energy pattern. However, in general, as PMMA has a higher number of incubation pulses and a higher variability in its ablation parameters, it will be more subject to low-pulse number related artifacts than Filofocon A.

Although it is not the case for the two lasers studied (in Ref. [39]), under certain circumstances regions may be approached where the Beer Lambert law fails (number of pulses close to incubation, fluences near the ablation threshold and altered ablation properties). These circumstances include lower effective fluences (due to reduced laser fluence or radial efficiency losses), ablation patterns with shallow regions (as hyperopic patterns), edges of the optical zone for low corrections, or transition zones. As a rule of thumb to avoid these effects, at a fluence of 120 mJ/cm², ablated zones deeper than 5 μm in Filofocon A and 7 μm in PMMA can be considered correctly ablated (with alterations in the depth pattern less than 1 pulse or 0.15 μm). At a fluence of 400 mJ/cm², all ablations are correctly transferred to depths. Since both materials show high linearity with the number of pulses, a valid strategy to avoid those effects is to multiply the number of pulses at each point, applying repeated ablations to the material.

4.2 Measuring ablation efficiency

An additional application of plastic corneal models for refractive surgery is the measurement of radial laser efficiency changes and calculation of correction factors. This is a more sophisticated application and involves the ablation of curved surfaces, with a geometry similar to that of the cornea [22]. The measured depth pattern in curved surfaces is compared with that obtained from flat surfaces in identical conditions. In this case, the pulse distribution pattern is exactly the same in both conditions, and the difference in depth is only due to geometry-associated variations in ablation efficiency.

The fact that the energy pattern can be correctly measured in flat surfaces of Filofocon A, and the linearity of the Beer Lambert in the ranges of operation, indicates that local energy differences associated to efficiency changes will be transferred to depth changes. The experimental complexity of measuring efficiency effects seems then to be associated to achieving precise alignment and accurate measurement techniques for the curved surfaces, and not to the ablation properties of the material. The same can be applied to the measurement of the induced asphericity (by refractive surgery) in curved surfaces.

4.3 Central islands in PMMA

Central islands are systematically observed in the ablation spot profiles in PMMA (Fig. 1), but not in Filofocon A. Our results reveal that it is a multipulse effect, not appearing with single
pulses. This suggests that defect accumulation during the first pulses can play a role in the island formation.

AFM images (Fig. 7) also suggest different ablation mechanisms in Filofocon A and PMMA. The droplets observed in PMMA may be indicative of a more explosive mechanism than in Filofocon A, where a dense worm-like structure appears after the ablation. These are also found to be multipulse effects. As the number of pulses increases, the droplet density increases in PMMA, while the worm thickness decreases in Filofocon A.

Central islands have been extensively reported in corneal tissue as isolated areas within the optical zone where the laser ablated tissue less efficiently. Central islands have been associated to shockwaves in water, shielding effects due to the plume and particle redeposition [44, 45]. Central islands have also been observed in PMMA [44, 45]. It is remarkable that it is not observed in Filofocon A, indicating that the effect is not inherent to the laser energy distribution. In any case, as was previously reported in corneas [45], the use of flying spot lasers in refractive surgery, and the associated averaging effect, reduces the impact of inhomogeneities in the laser spot.

4.4 Absorption coefficient

It is enlightening to compare the absorption coefficient $\alpha$ determined from low intensity ellipsometric measurements (Table 1) to the effective absorption coefficient $\alpha_{\text{eff}}$ estimated from the slope of the Beer-Lambert representation and plotted in Fig. 5. The corresponding values differ strongly for PMMA, with $\alpha_{\text{eff}}$ always being much higher than $\alpha$ even for a low number of pulses. This is consistent with what is reported in the literature for PMMA [30, 32]. In contrast, we observe a perfect agreement for Filofocon A in the stationary saturation regime at high pulse numbers ($\geq 10$ pulses). This agreement points to a fundamentally different absorption mechanism in Filofocon A, truly following a Beer-Lambert law in the multipulse regime as opposed to PMMA. As a consequence, the ablation behavior is expected to be also influenced, as evidenced by the different surface morphology/topography shown in Figs. 6 and 7.

5. Conclusions

We have presented the ablation properties of Filofocon A and PMMA in the operational range of refractive surgery excimer lasers.

PMMA is often used as an ablation material in the calibration of clinical lasers, and model eyes made out of PMMA and Filofocon A have been proposed as experimental models of corneal refractive surgery. We have found that the ablation properties of Filofocon A make it a better material than PMMA for laser ablation refractive surgery research and for the development of calibration protocols. Filofocon A provides a more accurate energy-to-depth conversion (with no central islands and other ablation artifacts), shows lower number of incubation pulses and ablation parameters less dependent on the number of pulses and, unlike corneal tissue and other materials, has a predictive absorption coefficient.

These findings are useful to design calibration and validation procedures for the refractive surgery excimer lasers, and to improve the outcomes and predictability of new ablation profile designs.

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