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Equivalence of two optical quality metrics to predict the visual acuity of multifocal pseudophakic patients

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Abstract: This article studies the relationship between two metrics, the area under the modulation transfer function (MTFa) and the energy efficiency (EE), and their ability to predict the visual quality of patients implanted with multifocal intraocular lenses (IOLs). The optical quality of IOLs is assessed in vitro using two metrics, the MTFa and EE. We measured them for three different multifocal IOLs with parabolic phase profile using image formation, through-focus (TF) scanning, three R, G, B wavelengths, and two pupils. We analyzed the correlation between MTFa and EE. In parallel, clinical defocus curves of visual acuity (VA) were measured and averaged from sets of patients implanted with the same IOLs. An excellent linear correlation was found between the MTFa and EE for the considered IOLs, wavelengths and pupils ($R^2 > 0.9$). We computed the polychromatic TF-MTFa, TF-EE, and derived mathematical relationships between each metrics and clinical average VA. MTFa and EE proved to be equivalent metrics to characterize the optical quality of the studied multifocal IOLs and also in terms of clinical VA predictability.

1. Introduction

The optical image quality of an intraocular lens (IOL) and its variation with defocus provides an essential knowledge to interpret the pseudophakic visual performance in a distance range [1–5]. Marsack et al. [6] investigate the ability of 31 scalar metrics derived from wavefront aberration maps to predict changes in the high-contrast visual acuity (VA). They found that the area under the modulation transfer function (MTFa) and the light in the bucket (LIB) [7] were among the six metrics that accounted for over 70% of the variance in VA (see Ref. [6] for details).

Recently, more efforts have been made to obtain optical image quality metrics computed from either on-bench measurements [1,3,4] or from optical simulations [5] that may correlate well with average visual outcomes tested in clinical assessments such as VA or contrast sensitivity (CS). When a high correlation is found, such metrics becomes a preclinical metrics, meaning that it is possible to predict the relative change in the clinical average defocus VA curves from the in-vitro on-bench measurements of the IOL. Felipe et al. [1] presented linear fits, with correlation coefficient of $R^2 = 0.91$, between clinical VA outcomes and laboratory modulation transfer function (MTF) measurements averaged in the spatial frequency range up to the unit of decimal visual acuity (100 c/mm or approx. 30 cpd). Alarcon et al. [3] proposed four different quality metrics, three of them based on weighted MTF values integrated in a spatial frequency range. Nonlinear fits of through-focus values of such three MTF-based metrics versus clinical VA

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assessed in patients implanted with six different IOL designs reached high correlation coefficients ($R^2 > 0.93$). Vega et al. [4] considered one of these metrics, the MTFa in the frequency integration range from 0 to 50 cycles/mm, under green light (530 nm), and hypothesize an asymptotic limit in the VA achievable with IOL implants of exceedingly large MTFa. With these assumptions and after refining the non-linear function that fits optical bench MTFa values with clinical VA data for a modeling IOL set (consisting of one monofocal and two bifocals), they were able to predict clinical VA at different defocus levels of patients implanted with trial IOLs of different design (two trifocals and one extended range of vision).

Fernandez et al. [5] built a mean eye model using the biometric eye data of 43 patients implanted with one diffractive trifocal design. MTFa values for 4 pupil sizes and 11 defocus locations were obtained from optical simulation and correlated with the VA and CS of the patients. Using a linear model, they found a change of VA and CS per unit of MTFa of −0.01 logMAR and −0.02 logC, respectively.

The energy efficiency (EE) is a metric widely used to account for the energy directed to the image formed by the lens. In the experimental practice, the energy efficiency can be approached by the LIB measurement [7,8], which is useful to describe the through–focus (TF) performance of diffractive IOLs [8–10]. It helps to characterize the energy distribution between the foci of multifocal IOLs, and its variation with pupil [11], all of which is strongly related with the formation of haloes [12] and has an impact on photic phenomena [13] affecting pseudophakic patients. The EE metric may evaluate the depth of focus too and hence, describe the extended depth of focus (EDOF) IOLs that would facilitate an extended range of vision (ERV) once implanted. Since the EE of diffractive optical components shows a strong dependence with wavelength, LIB measurements have been extensively used to analyze multifocal and ERV diffractive IOLs under polychromatic illumination and, therefore, derive the longitudinal chromatic aberration and the spectral distribution of energy at each focus. A through-focus polychromatic analysis of EE allows the detection of unbalanced distributions of energy and its wavelength dependence [8–10,14].

In this work, we investigate the potential of EE to predict postoperative VA of subjects implanted with diffractive IOLs. From in-vitro on-bench EE measurements and clinical VA outcomes, we further analyze the correlation of a LIB-based metric (EE) with an MTF-based metric (MTFa) already used as preclinical metrics for the same purpose. Some previous works, in addition to ours, made us point to this direction. The effects of diffraction efficiency on the MTF of diffractive lenses were theoretically studied by Buralli and Morris [15], who found that the MTF at a given focus can be approached by a component caused by the order of interest, scaled by the integrated efficiency of such order (that integrated efficiency being the fraction of energy in the focused component of the point-spread function), plus a spike at the zero spatial frequency with the contribution of the remaining orders. Simpson had presented results referred to diffractive multifocal IOLs consistent with that statement in his independent and shortly earlier work [12]. More recently, He et al. [16,17] simulated one hundred virtual pseudophakic eyes and reported succinctly linear correlations between functions of image quality metrics -MTFa and EE (LIB)- and existing clinical VA data of patients implanted with multifocal IOLs.

In the first part of this work, we measure through-focus EE and MTFa for three diffractive IOLs using an on-bench eye model under the sequential illumination of red (R), green (G) and blue (B) quasi-monochromatic lights and study a hypothetical correlation between the values of both metrics. In a second part, we analyze a possible correlation between the polychromatic TF-EE, synthesized from the R, G, B measured TF-EE curves, and VA tested clinically at different defocus amounts. Our study is limited to diffractive IOLs with either a single [18,19] or two combined [20,21] parabolic phase profiles of three different designs (trifocal ERV, trifocal apodized, and ERV) and two pupil sizes. Since MTFa has been proved to serve as preclinical

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metric for average visual performance [3–5,22], the results of this study will clarify whether EE metric can be used in a similar way.

2. Method

2.1. Intraocular lenses and experimental setup

Three different IOLs of 20 D distance vision power were tested in vitro on optical bench. They had different designs concerning focality, parabolic phase profile, and other specifications (Table 1): a trifocal ERV AcrivaUD Reviol Tri-ED (VSY Biotechnology, Istanbul, Turkey), a trifocal apodized FineVision MicroF (Physiol, Lieje, Belgium), and a ERV Symfony ZXR00 (Tecnis, Abbott Medical Optics, Abbott Park, IL).

Table 1. Characteristics of the three multifocal IOLs tested.

	Acriva^{UD} Reviol Tri-ED	Fine Vision MicroF	Tecnis ZXR00 Symfony
Material	Acrylic with Hydrophobic surface	Hydrophilic Acrylic	Hydrophobic Acrylic
Refractive index n	1.462	1.46	1.47
Aspheric surface	Anterior	Posterior	Anterior
Spherical aberration[a] **SA = c[4,0] (μm)**	−0.165	−0.11	−0.27
Nominal Power (D)	20	20	20
Add Power (D)	+1.5; +3	+1.75; +3.5	ERV[b]

[a] *For a 6 mm Entrance Pupil. Value of the Zernike coefficient* $C_{4,0}$.
[b] *Some authors* [10,23] *found a bifocal design with an equivalent add power of 1.75 D at the design wavelength.*

We have used an on-bench eye model to obtain the in vitro images formed by the IOLs. The setup, described in detail elsewhere [24,25], is constituted by an illumination system, an eye model and an image acquisition system. The eye model, in turn, consisted of an artificial cornea lens, an iris diaphragm, and a cuvette filled with saline solution where the IOL is immersed. The eye model met the recommendations of the International Standard Organization 11979-2:2014 [26]. Concerning the use of an aberration-inducing artificial cornea for evaluation of aspheric IOLs, we used an achromatic doublet (Lambda-X, Belgium) that induced + 0.17 μm of spherical aberration (SA) at the IOL plane (for a 5.0 mm pupil).

The light sources of the illumination system were three light emitting diodes (LED) (Thorlabs GmbH, Munich, Germany) with emissions centered nominally at 455 nm (B), 530 nm (G) and 625 nm (R) with a full-width half-maximum spectral band width of 18 nm, 33 nm and 18 nm, respectively. These LEDs illuminated sequentially an object test, which was either a four-slit test (two horizontal and two vertical slits of 10 μm width) for MTF measurements, or a 200 μm pinhole for EE evaluation. The test object was optically located at infinity by placing it at the front focal plane of a collimator (200 mm focal length).

The image acquisition system was composed of a 10X, infinity-corrected, plan-achromatic, microscope objective assembled to an 8-bit CCD camera, mounted on a high precision, three-axis translation holder for through focus analysis. The image acquisition system (microscope and camera) was nearly diffraction limited across the visible spectrum with a cutoff frequency of 675 cycles/mm. To reduce the impact of electronic noise, each image was the result of averaging eight frames at a time.

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2.2. On-bench measurement of MTFa and EE quality metrics

The optical quality of the IOLs was assessed under separate R, G and B illumination, taking MTFa and EE measurements within a through-focus range of image vergence (−4.0 D to + 2.0 D, 0.10 D step). Two pupils - 3.0 mm and 4.5 mm, measured at the IOL plane-, were considered.

To calculate the MTFa for a given image vergence within the TF range, we computed first the MTF from the images of the four-slit test produced by the model eye with the IOL under study as reported elsewhere [4]. We averaged the four MTF curves in the horizontal and vertical directions and integrated the average MTF curve in the spatial frequency range from 0 to 50 cycles/mm, which turned out to be the MTFa value.

Regarding the measurement of the EE for a given image vergence within the TF range, we used the pinhole test and calculated the LIB value from the image formed by the model eye as reported in detail elsewhere [8]. Essentially, the pinhole image core to total energy ratio approaches the LIB value [7] in experimental practice.

In our experiment, the origin of image vergence and defocus (0.0 D) was set at the distance image for the G light (530 nm, close to the standard design wavelength of 546 nm [26]). Negative dioptric value corresponds to near vision vergence according to the clinical convention. For each IOL and pupil size, the R, G, and B TF-MTFa and TF-EE curves were experimentally obtained. From these data, we calculated the polychromatic functions [15,27]

$$F_{poly}(x) = \frac{\sum_{\lambda}^{R,G,B} S_\lambda F(x)_\lambda}{\sum_{\lambda}^{R,G,B} S_\lambda} \tag{1}$$

where S_λ denotes the spectral distribution and responsivity of the source-detector combination, F is the experimentally obtained metric F={EE, MTFa}, and the variable x (D) is the image vergence or defocus position within the TF range. For the equal–energy white spectrum and standard observer photopic sensitivity function V(λ), the weight coefficients are approximated by S_λ (R,G,B) ≈ {V(625), V(530), V(455)} = {0.321, 0.862, 0.048} [28]. From Eq. (1) we calculated $\mathrm{MTFa}_{\mathrm{poly}}$(x) and $\mathrm{EE}_{\mathrm{poly}}$(x) in the TF range.

2.3. Clinical data

The clinical data for this study were obtained from 102 eyes of 52 patients recruited in two clinical trials carried out at two ophthalmology centers (Table 2). Both studies were prospective, consecutive and non-randomized and followed the tenets of the declaration of Helsinki. The patients underwent bilateral and symmetrical cataract surgery followed by IOL implantation into the capsular bag. Previously, they had been fully informed about the study and signed a consent form. The local ethics committee approved the corresponding trial. At each setting, all the data were collected by one experienced staff member (N.G. at Miranza IOA and I.A. at the hospital Miguel Servet, Zaragoza).

Table 2. Resume of the clinical studies.

Setting	IOL implanted	Number of patients (eyes)
Miranza IOA, Innova Ocular, Madrid (Spain)	Acriva^{UD} Reviol TRI-ED (trifocal ERV)	15 (30)
	FineVision (trifocal apodized)	11 (21)
Miguel Servet University Hospital, Zaragoza (Spain)	Tecnis ZXR00 Symfony (ERV)	26 (51)

Monocular defocus VA curves from −5.00 D to + 3.00 D, with the patients having their best distance correction, were measured in logMAR scale during the last postoperative follow up. Measurements were taken following the procedure described by Wolffsohn et al. [29] using the 100% contrast Early Treatment Diabetic Retinopathy Study (ETDRS) chart at 4 m under

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photopic conditions and with natural eye pupil. The clinical VA outcomes VA(x) obtained in the studied defocus range (i.e., VA defocus curves) will be related to the on-bench F(x) polychromatic metrics [Eq. (1)]. Further details about the inclusion criteria, the surgical procedure and the clinical outcomes are reported elsewhere [4].

3. Results

3.1. Correlation between EE and MTFa

Figure 1 shows, for the three lenses studied and 3mm pupil, the R, G, B TF-MTFa curves [Figs. 1(a)–1(c)] and the R, G, B TF-EE curves [Figs. 1(d)–1(f)]. Figure 2 shows the equivalent curves for a pupil of 4.5 mm. Comparing both groups of curves, one realizes that, for a given lens and wavelength, the TF-MTFa and TF-EE curves appear to be very similar. From these results, it is straightforward for every IOL and wavelength, to obtain at each defocus position x (D) the corresponding pair of values $(\mathrm{MTFa, EE})_x$ and to study a possible correlation between both magnitudes. The results are shown in Fig. 3 which reveals a very good linear correlation between the MTFa and EE functions. To help with the interpretation of these results, we have highlighted in Fig. 1 (dashed circles) the values around the peaks of the MTFa [Fig. 1(c) -values about 30 to 35-] and EE [Fig. 1(f) -values about 0.6 to 0.8-] obtained with the blue wavelength in the case of the Symfony IOL, and the position of the corresponding $(\mathrm{MTFa, EE})_x$ pairs in Fig. 3(a) (points inside the dashed circle).

Figure 3(b) shows a similar correlation for the results obtained with 4.5 mm pupil. From these results, we obtain linear correlations between MTFa and EE for the two pupils. Their equations and correlation coefficients (R^2) are

$$Pupil\ 3.0\,mm:\ MTF_a = 40.81 * EE + 3.36, \quad R^2 = 0.97 \tag{2}$$

$$Pupil\ 4.5\,mm:\ MTF_a = 35.42 * EE + 3.82, \quad R^2 = 0.91 \tag{3}$$

Both linear fits show excellent correlation ($R^2 > 0.90$), but with a slightly different slope. The lower slope in the case of 4.5 mm pupil [Fig. 3(b)] indicates that, for a given EE value, the MTFa decreases when opening the lens aperture.

3.2. Correlation between the polychromatic EE and clinical VA

From the R, G, B TF-MTFa and TF-EE curves (Fig. 1) and Eq. (1), we compute the polychromatic TF functions $\mathrm{MTFa_{poly}}(x)$ and $\mathrm{EE_{poly}}(x)$ for the three IOLs and two pupils (3.0 and 4.5 mm) (Fig. 4). For each IOL, the similarity between its $\mathrm{MTFa_{poly}}(x)$ and $\mathrm{EE_{poly}}(x)$ curves is maintained.

Following the methodology outlined in closely related studies [3,4], we represent the clinical VA(x) outcome versus the $\mathrm{MTFa_{poly}}(x)$ for every defocus or image vergence (x) by means of blue dots (experimental points) in Fig. 5(a). A similar representation for clinical VA(x) outcomes versus $\mathrm{EE_{poly}}(x)$ appears in Fig. 5(b). We recall that the points of Fig. 5 refer to the clinical VA defocus curves obtained from patients implanted with some of the three IOLs: trifocal ERV Acriva Reviol Tri-ED, trifocal apodized FineVision, and ERV Tecnis ZXR00 Symfony. To study possible correlations between clinical VA and on-bench $\mathrm{MTFa_{poly}}$ and $\mathrm{EE_{poly}}$ metrics, we fit the experimental points of Figs. 5(a) and 5(b) either to a rational function (red solid line) [Eqs. (4), (5)] or to an exponential function (green solid line) [Eqs. (6) and (7)]. The mathematical expressions and their correlation coefficients (R^2) are

$$VA = \frac{2.361}{MTFa_{poly} - 1.896} - 0.098 \qquad R^2 = 0.94 \tag{4}$$

$$VA = \frac{0.065}{EE_{poly} - 0.060} - 0.1 \qquad R^2 = 0.85 \tag{5}$$

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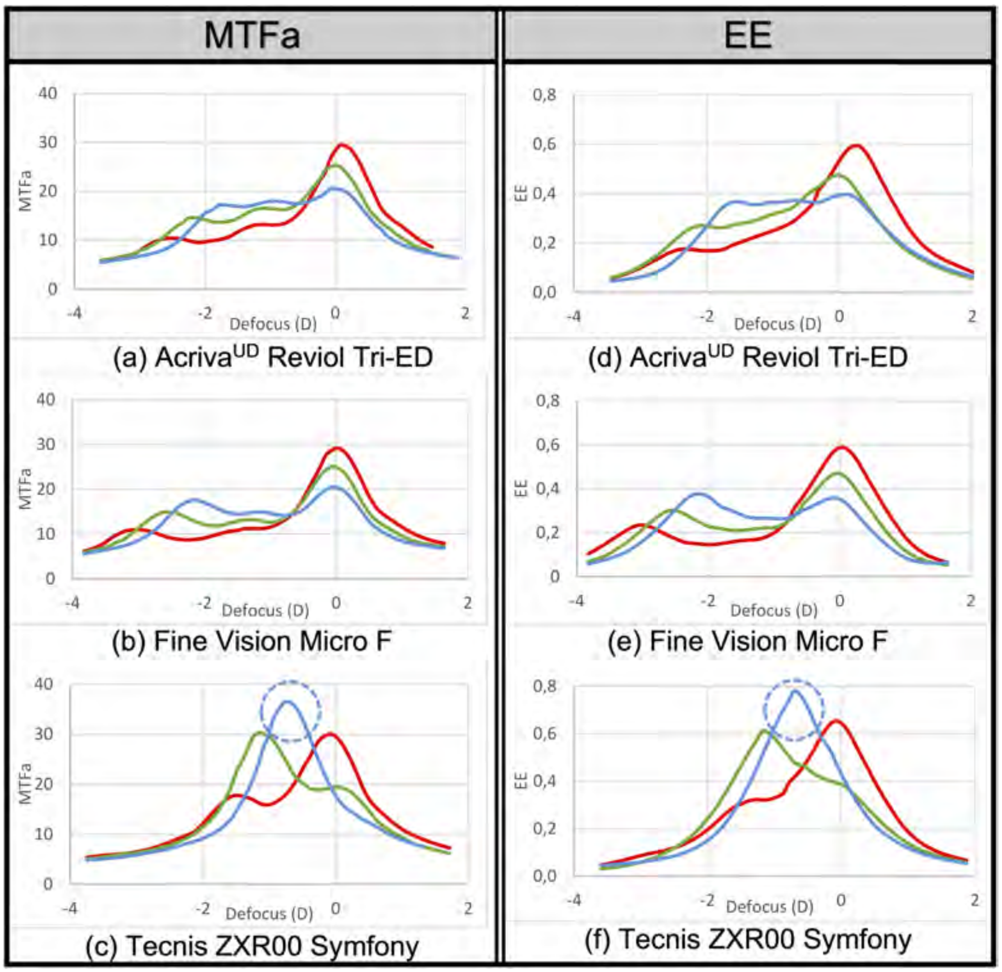

Fig. 1. R, G and B through-focus MTFa (left column) and EE (right column) curves obtained with a 3.0 mm pupil for three IOLs: trifocal Acriva Reviol Tri-ED (a, d), trifocal FineVision Micro F (b, e), and ERV Tecnis ZXR00 Symfony (c, f). Blue, green and red solid lines correspond to blue (455 nm), green (530 nm) and red (625 nm) illumination, respectively. Dashed circles in (c, f) indicate the peak values of blue MTFa and EE of the Symfony IOL (see text for details).

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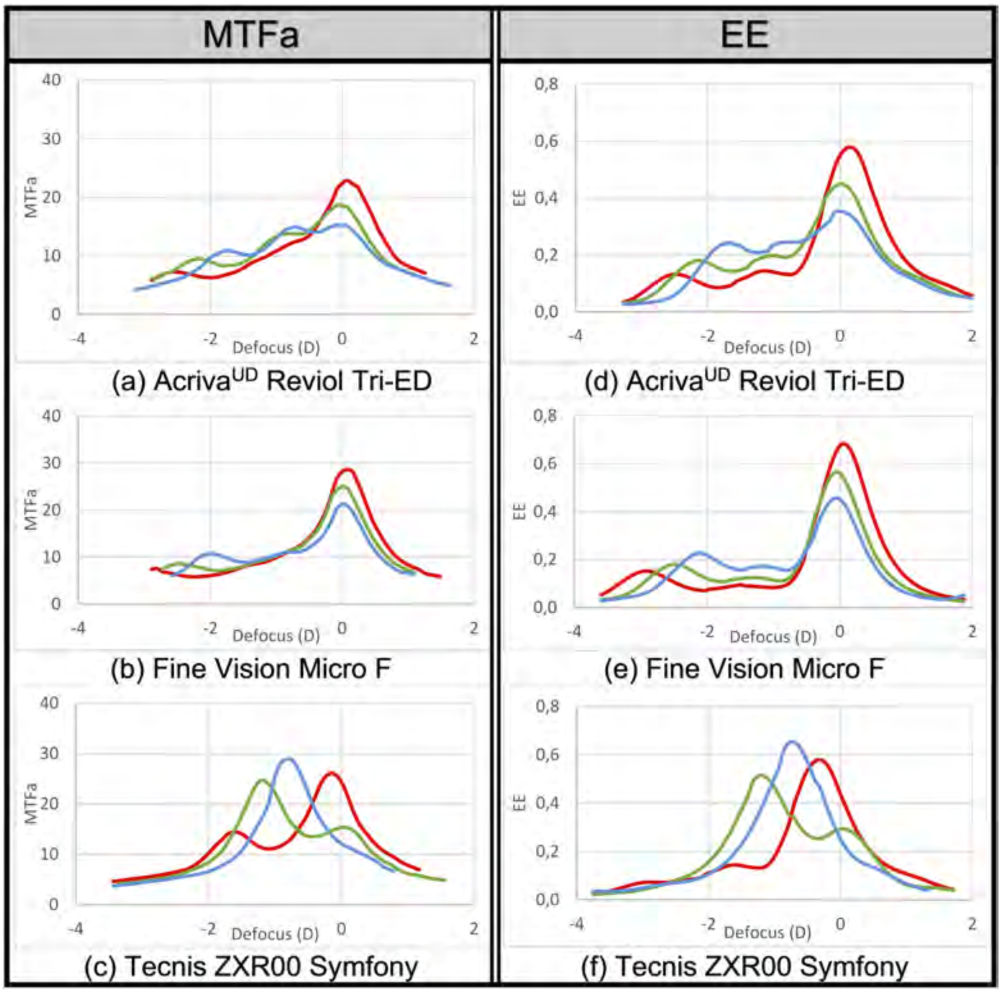

Fig. 2. R, G and B through-focus MTFa (left column) and EE (right column) curves obtained with a 4.5 mm pupil for three IOLs: trifocal Acriva Reviol Tri-ED (a, d), trifocal FineVision Micro F (b, e), and ERV Tecnis ZXR00 Symfony (c, f). Blue, green and red solid lines correspond to blue (455 nm), green (530 nm) and red (625 nm) illumination, respectively.

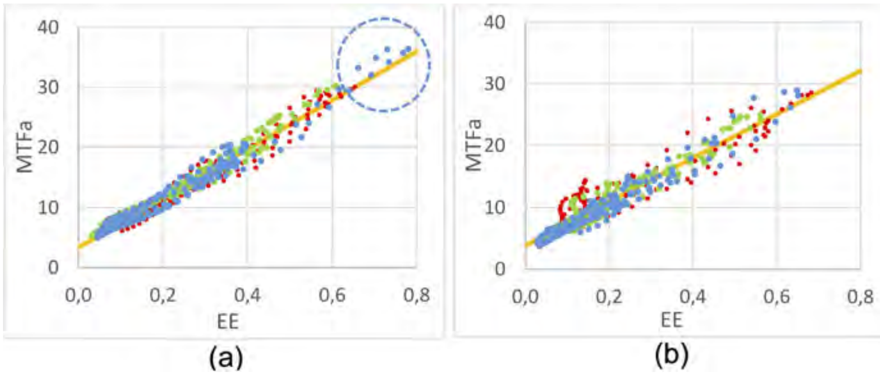

Fig. 3. Representation of $(\mathrm{MTFa}, \mathrm{EE})_x$ pairs for the three IOLs; R, G, B, illuminations; and defocus position (x) within the TF-range contained in (a) Fig. 1 (3.0 mm pupil) and (b) Fig. 2 (4.5 mm pupil). Blue, green and red dots correspond to measurements obtained under blue (455 nm), green (530 nm) and red (625 nm) illumination, respectively. Points inside the dashed circle correspond to the peak values of blue MTFa (Fig. 1(c)) and EE (Fig. 1(f)) of the Symfony IOL (see text for details). The solid yellow lines indicate the linear fits.

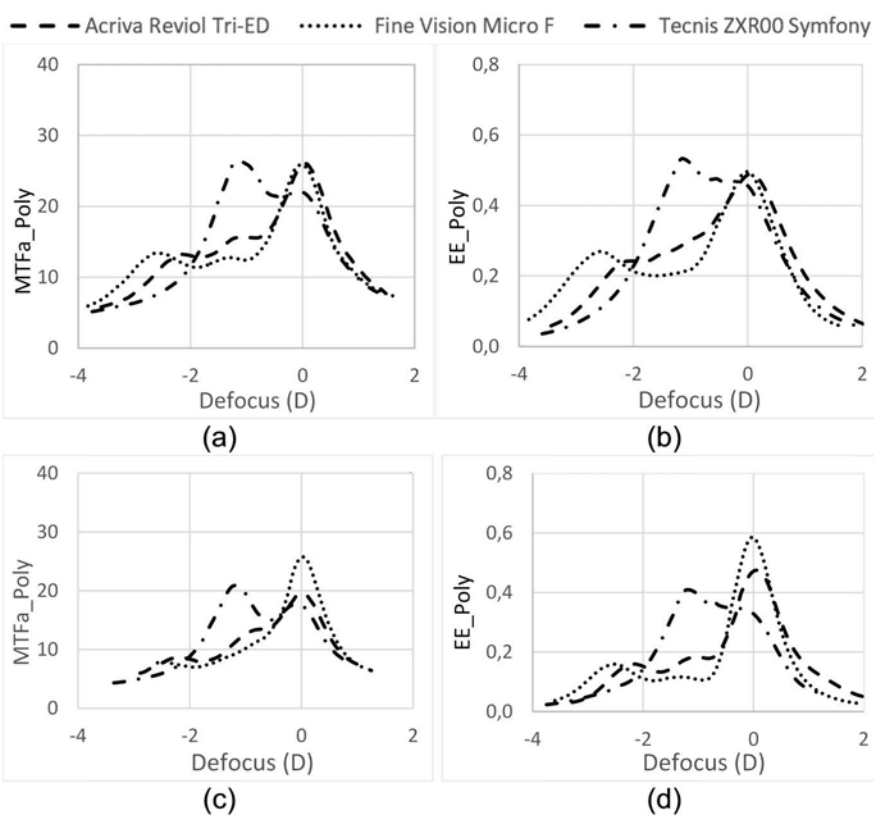

Fig. 4. Polychromatic through-focus functions for the three IOLs: (a) $\mathrm{MTFa}_{poly}(x)$ and (b) $\mathrm{EE}_{poly}(x)$ with 3.0 mm pupil and (c) $\mathrm{MTFa}_{poly}(x)$ and (d) $\mathrm{EE}_{poly}(x)$ with 4.5 mm pupil.

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$$VA = 1.828 * e^{-0.230*MTFa_{poly}} + 0.014 \quad R^2 = 0.94 \tag{6}$$

$$VA = 0.707 * e^{-7.746*EE_{poly}} \quad R^2 = 0.90 \tag{7}$$

The independent term of Eq. (5) was set to −0.1 when fitting the mathematical expression so as to allow VA to take more realistic values (see the discussion for details).

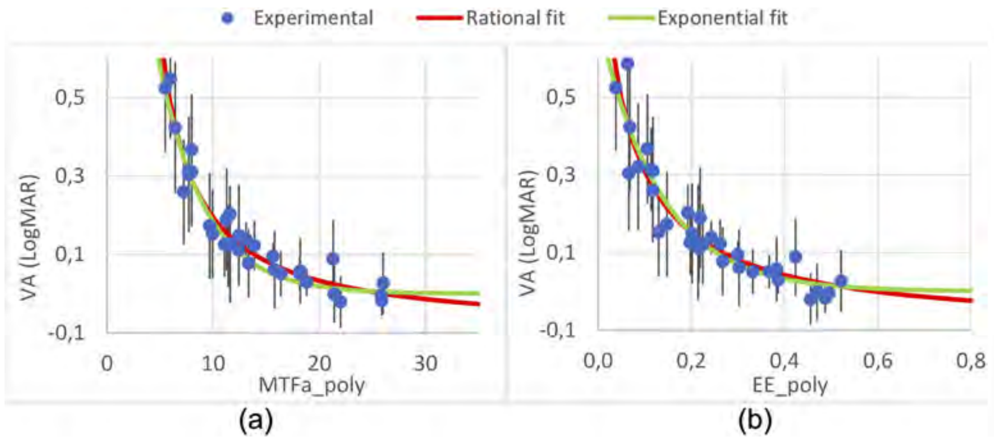

Fig. 5. Clinical Visual Acuity versus (a) $MTFa_{poly}$ and (b) EE_{poly} for 3.0 mm pupil. Each experimental point (blue dot) is represented along with its VA standard deviation (vertical bar). Rational (red line) and exponential (green line) fit functions.

4. Discussion

MTFa and EE are objective metrics to assess in-vitro the optical image quality of IOLs; the larger the value of any of these two metrics, the better the optical quality of the IOL. Buralli and Morris [15] provided an approximated expression between the optical transfer function (OTF) and the EE for a diffractive lens and concluded that "a complete characterization of an optical system that contains one or more diffractive components should include, in addition to conventional aberration analysis, an evaluation of the integrated efficiency (i.e. the fraction of energy in the focused component of the point-spread function), which will generally be a function of field position". They did not attempt to provide a rigorous calculation of diffraction efficiency for a diffractive lens, which depends on parameters of the particular optical system such as wavelength, lens material, incidence angles, polarization and surface profile description. However, they found that the OTF consists of a spike at zero spatial frequency and a component, corresponding to the diffraction order of interest that contributes to the analyzed focus, scaled by the EE. The approach is also valid when the diffractive lens works under broadband illumination.

Our experimental results (Fig. 3) show a clear linear correlation between MTFa and EE values ($R^2 > 0.90$). Although we consider MTFa rather than MTF values at a single spatial frequency, our results are fully consistent with the predictions of Buralli and Morris [15].

Interestingly, we report a lower slope of the linear fit when the lens aperture increases from 3.0 mm [Fig. 3(a), Eq. (2)] to 4.5 mm pupil [Fig. 3(b), Eq. (3)]. This result shows that the image degradation caused by the presence of larger amounts of aberration with increasing pupil affects differently to MTFa and EE. More precisely, the MTFa decreases quicker than EE when the optical quality is lower as a consequence of larger pupils and amounts of aberration. This result leads us to consider the EE metric preferable to MTFa when dealing with images of relative low optical quality due to defocus or higher-order aberrations.

From the experimental on-bench results demonstrating a linear correlation between MTFa and EE values and the potential of MTFa as a preclinical metric for predicting average VA assessment at different defocus values (i.e. defocus VA curves) of pseudophakic patients [3–5], we have studied the ability of the EE metric to predict clinical VA outcomes of patients implanted with three different designs of diffractive IOLs. TF-EE curves have proved to provide valuable information about the optical quality of IOLs being tested: particularly, for multifocal or ERV diffractive lenses, the energy distribution between the foci [11], longitudinal chromatic aberration [10], chromatic distribution of energy in each focus [8], halo size, [12,13], and so on. However, energy-based metrics such as the EE have been rarely studied as predictors of VA [16,17] and, to the best of our knowledge, no results from experimental in-vitro EE measurements and clinical in-vivo VA assessments have been reported yet.

Clinically, the VA acuity is assessed under white light illumination. For this reason, we have considered polychromatic TF-EE and TF-MTFa prior to correlate with VA outcomes. Let us remark that in this work, although for different IOLs and patients, we have obtained better correlation [$R^2 = 0.94$, Eq. (6)] by using polychromatic MTFa than in a previous study [4], where we used just monochromatic (green) MTFa ($R^2 \approx 0.90$).

The results show, in agreement with former works [3,4,22], non-linear relationships between VA and both, $MTFa_{poly}$ and EE_{poly} metrics. The mathematical functions that fit the data of Fig. 5 with high correlation coefficients [$R^2 = 0.94$ for VA versus $MTFa_{poly}$ - Eq. (6) and $R^2 = 0.90$ for VA versus EE_{poly} - Eq. (7)] clear a path to predict average VA of pseudophakic patients from objective on-bench quality metrics.

We have seen that MTFa and EE metrics are linearly correlated for the studied IOLs, does this mean that one of them is redundant or superfluous? In our opinion, not necessarily, because they provide complementary information that can be easier to interpret for other instances. Moreover, such a correlation is approximated and, as evidenced through Figs. 3(a) and 3(b), may change their mathematical expression depending on the pupil [5]. Other optical parameters concerning, for example, the surface profile or the diffractive design may have an influence too on the mathematical expression of the correlation between MTFa and EE, and deserve further investigation.

With regard to the use of a rational or an exponential function for fitting clinical VA vs $MTFa_{poly}$ and EE_{poly} (Fig. 5), it must be taken into account that the best VA reported in pseudophakic patients is typically around 0.0 logMAR for distance vision. Then, it would be desirable that the fitting functions (rational or exponential) of VA vs $MTFa_{poly}$ or EE_{poly} [Fig. 5(a) and 5(b)] reflected this constraint in the achievable VA, i.e., the largest $MTFa_{poly}$ and EE_{poly} values should correspond with the best achievable VA. In other words, it is a realistic assumption that for large $MTFa_{poly}$ and EE_{poly} values the fitting functions tend to the best VA achievable in the clinics. In practice, to fit the patients' VA versus $MTFa_{poly}$ or EE_{poly} [Figs. 5(a) and 5(b), respectively] with a rational function, one needs to add a constant C (i.e. a horizontal asymptote) that corresponds to the potentially best VA achievable by the pseudophakic patients with these IOLs. The highest correlation (i.e. best fit) with a rational function is obtained with C constant of ≈ -0.1 logMAR [Eqs. (4) and (5)], which is somewhat better than the best VA clinically measured in the patients. Alternatively, the exponential fitting shows with either the $MTFa_{poly}$ or EE_{poly} metrics [Eqs. (6) and (7)] an asymptote value of VA of 0 logMAR [Eq. (7)] or very close to 0 logMAR [Eq. (6)], which seems to be more realistic regarding the VA outcomes in these type of patients.

5. Conclusions

From the independent on-bench measurements of MTFa and EE for three different multifocal IOLs we have found a linear correlation between MTFa and EE indicating that both metrics can be a good indicator to measure the optical quality of multifocal IOLs and to predict average VA. When increasing the pupil, MTFa diminishes more quickly than EE, indicating that MTFa is more sensitive to pupil dependent aberrations.

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From the measurements with RGB wavelengths, we have computed the through-focus polychromatic metrics, $MTFa_{poly}$ and EE_{poly}, and obtained very good correlation with clinical average postoperative VA. We have derived mathematical expressions that can be used to predict average VA from $MTFa_{poly}$ and EE_{poly}.

In the conditions of our study, TF-EE is a very useful metric. As MTFa, EE provides information about longitudinal chromatic aberration, depth of focus and predicts average postoperative VA. The dependence of EE with wavelength at each focus of a diffractive multifocal IOL provides information about possible chromatic unbalance in the image when changing the focusing distance (vergence). And last but not least, TF-EE allows analyzing the photic phenomena associated with multifocal IOLs. In the conditions of our experiment, the MTFa can be replaced by EE as an optical quality parameter.

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Disclosures

The authors declare that there are no conflicts of interest related to this article.

References

1. A. Felipe, F. Pastor, J. M. Artigas, A. Diez-Ajenjo, A. Gené, and J. L. Menezo, “Correlation between optics quality of multifocal intraocular lenses and visual acuity. Tolerance to modulation transfer function decay,” J. Cataract Refractive Surg. **36**(4), 557–562 (2010).
2. A. B. Plaza-Puche, J. L. Alió, S. MacRae, L. Zheleznyak, E. Sala, and G. Yoon, “Correlating Optical Bench Performance With Clinical Defocus Curves in Varifocal and Trifocal Intraocular Lenses,” J. Refract. Surg. **31**(5), 300–307 (2015).
3. A. Alarcon, C. Canovas, R. Rosen, H. Weeber, L. Tsai, K. Hileman, and P. Piers, “Preclinical metrics to predict through-focus visual acuity for pseudophakic patients,” Biomed. Opt. Express **7**(5), 1877–1888 (2016).
4. F. Vega, M. S. Millán, N. Garzón, I. Altemir, F. Poyales, and J. M. Larrosa, “Visual acuity of pseudophakic patients predicted from in-vitro measurements of intraocular lenses with different design,” Biomed. Opt. Express **9**(10), 4893–4906 (2018).
5. J. Fernández, M. Rodríguez-Vallejo, J. Martínez, N. Burguera, and D. P. Piñero, “Prediction of Visual Acuity and Contrast Sensitivity From Optical Simulations With Multifocal Intraocular Lenses,” J. Refract. Surg. **35**(12), 789–795 (2019).
6. J. D. Marsack, L. N. Thibos, and R. A. Applegate, “Metrics of optical quality derived from wave aberrations predict visual performance,” J. Vis. **4**(4), 322–328 (2004).
7. L. N. Thibos, X. Hong, A. Bradley, and R. A. Applegate, “Accuracy and precision of objective refraction from wavefront aberrations,” J. Vis. **4**(4), 329–351 (2004).
8. M. S. Millán, F. Vega, and I. Ríos-López, “Polychromatic image performance of diffractive bifocal intraocular lenses: Longitudinal chromatic aberration and energy efficiency,” Investig. Ophthalmol. Vis. Sci. 57(4), 2021–2028 (2016).
9. S. Ravikumar, A. Bradley, and L. N. Thibos, “Chromatic aberration and polychromatic image quality with diffractive multifocal intraocular lenses,” J. Cataract Refractive Surg. **40**(7), 1192–1204 (2014).
10. M. S. Millán and F. Vega, “Extended depth of focus intraocular lens Chromatic performance,” Biomed. Opt. Express **8**(9), 4294–4309 (2017).
11. F. Vega, F. Alba-Bueno, and M. S. Millán, “Energy distribution between distance and near images in apodized diffractive multifocal intraocular lenses,” Investig. Ophthalmol. Vis. Sci. 52(8), 5695–5701 (2011).
12. M. J. Simpson, “Diffractive multifocal intraocular lens image quality,” Appl. Opt. **31**(19), 3621–3626 (1992).
13. F. Alba-Bueno, N. Garzón, F. Vega, F. Poyales, and M. S. Millán, “Patient-Perceived and Laboratory-Measured Halos Associated with Diffractive Bifocal and Trifocal Intraocular Lenses,” Curr. Eye Res. **43**(1), 35–42 (2018).
14. G. Łabuz, G. U. Auffarth, A. Özen, T. J. T. P. van den Berg, T. M. Yildirim, H.-S. Son, and R. Khoramnia, “The Effect of a Spectral Filter on Visual Quality in Patients with an Extended-Depth-Of-Focus Intraocular Lens,” Am. J. Ophthalmol. **208**, 56–63 (2019).
15. D. A. Buralli and G. M. Morris, “Effects of diffraction efficiency on the modulation transfer function of diffractive lenses,” Appl. Opt. **31**(22), 4389–4396 (1992).
16. L. He, M. Choi, X. Hong, and R. Sarangapani, “Correlation between clinical visual acuity and optical/visual metrics at varied defocuses in pseudophakic patients implanted with monofocal and multifocal IOLs,” Invest. Ophthalmol. Vis. Sci. **56**(7), 2978 (2015).
17. L. He, X. Hong, and R. Sarangapani, “Population-based simulation using image quality metrics to predict visual acuity in pseudophakic patients implanted with trifocal IOLs,” Invest. Ophthalmol. Vis. Sci. **58**(8), 4209 (2017).

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18. A. L. Cohen, "Practical design of a bifocal hologram contact lens or intraocular lens," Appl. Opt. **31**(19), 3750–3754 (1992).
19. F. Castignoles, M. Flury, and T. Lepine, "Comparison of the efficiency, MTF and chromatic properties of four diffractive bifocal intraocular lens designs," Opt. Express **18**(5), 5245–5256 (2010).
20. D. Gatinel, C. Pagnoulle, Y. Houbrechts, and L. Gobin, "Design and qualification of a diffractive trifocal optical profile for intraocular lenses," J. Cataract Refractive Surg. **37**(11), 2060–2067 (2011).
21. J. Loicq, N. Willet, and D. Gatinel, "Topography and longitudinal chromatic aberration characterizations of refractive–diffractive multifocal intraocular lenses," J. Cataract Refractive Surg. **45**(11), 1650–1659 (2019).
22. G. Cardona, F. Vega, M. A. Gil, C. Varón, J. A. Buil, and M. S. Millán, "Visual acuity and image quality in 5 diffractive intraocular lenses," Eur. J. Ophthalmol. **28**(1), 36–41 (2018).
23. D. Gatinel and J. Loicq, "Clinically Relevant Optical Properties of Bifocal, Trifocal, and Extended Depth of Focus Intraocular Lenses," J. Refract. Surg. **32**(4), 273–280 (2016).
24. F. Vega, F. Alba-Bueno, and M. S. Millán, "Energy Distribution between Distance and Near Images in Apodized Diffractive Multifocal Intraocular Lenses," Invest. Ophthalmol. Visual Sci. **52**(8), 5695–5701 (2011).
25. F. Vega, F. Alba-Bueno, and M. S. Millán, "Energy efficiency of a new trifocal intraocular lens," J. Eur. Opt. Soc. Rapid Publ. **9**, 14002 (2014).
26. International Organization for Standardization, ISO 11979–2:2014 "Ophthalmic Implants - Intraocular Lenses -Part 2: Optical Properties and Test Methods." Geneva; ISO. (2014).
27. S. Ravikumar, L. N. Thibos, and A. Bradley, "Calculation of retinal image quality for polychromatic light," J. Opt. Soc. Am. A **25**(10), 2395–2407 (2008).
28. International Organization for Standarization, ISO 23539:2005 Photometry — The CIE System of Physical Photometry (2005).
29. J. S. Wolffsohn, A. N. Jinabhai, A. Kingsnorth, A. L. Sheppard, S. A. Naroo, S. Shah, P. Buckhurst, L. A. Hall, and G. Young, "Exploring the optimum step size for defocus curves," J. Cataract Refractive Surg. **39**(6), 873–880 (2013).