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Equivalence of two optical quality metrics to predict 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.

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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 assessed in patients implanted with six different IOL designs reached high correlation coefficients.
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 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>
<thead>
<tr>
<th>Material</th>
<th>Acriva(^b)</th>
<th>Reviol Tri-ED</th>
<th>Fine Vision MicroF</th>
<th>Tecnis ZXR00 Symfony</th>
</tr>
</thead>
<tbody>
<tr>
<td>Refractive index n</td>
<td>1.462</td>
<td>1.46</td>
<td>1.47</td>
<td></td>
</tr>
<tr>
<td>Aspheric surface</td>
<td>Acrylic with Hydrophobic surface</td>
<td>Hydrophilic Acrylic</td>
<td>Hydrophobic Acrylic</td>
<td></td>
</tr>
<tr>
<td>Spherical aberration(^a)</td>
<td>−0.165</td>
<td>−0.11</td>
<td>−0.27</td>
<td></td>
</tr>
<tr>
<td>Nominal Power (D)</td>
<td>20</td>
<td>20</td>
<td>20</td>
<td></td>
</tr>
<tr>
<td>Add Power (D)</td>
<td>+1.5; +3</td>
<td>1.75; +3.5</td>
<td>ERV (^b)</td>
<td></td>
</tr>
</tbody>
</table>

\(^a\)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.

2.2. On-bench measurement of MTF\(^a\) and EE quality metrics

The optical quality of the IOLs was assessed under separate R, G and B illumination, taking MTF\(^a\) 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 MTF\(^a\) 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

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]

\[ F_{poly}(x) = \frac{\sum_{\lambda} R, G, B \cdot S_{\lambda} \cdot F_{\lambda}(x)}{\sum_{\lambda} R, G, B \cdot S_{\lambda}} \]  

where \( S_{\lambda} \) 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 \( V(\lambda) \), the weight coefficients are approximated
by \( S_{\lambda}(R,G,B) \approx \{V(625), V(530), V(455)\} = \{0.321, 0.862, 0.048\} \) [28]. From Eq. (1) we
calculated \( MTFa_{poly}(x) \) and \( EE_{poly}(x) \) 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 IOA Madrid and I.A. at the hospital
Miguel Servet, Zaragoza).

Table 2. Resume of the clinical studies.

<table>
<thead>
<tr>
<th>Setting</th>
<th>IOL implanted</th>
<th>Number of patients (eyes)</th>
</tr>
</thead>
<tbody>
<tr>
<td>IOA Madrid, Innova Ocular, Madrid (Spain)</td>
<td>Acryla UD Reviol TRI-ED (trifocal ERV)</td>
<td>15 (30)</td>
</tr>
<tr>
<td></td>
<td>FineVision (trifocal apodized)</td>
<td>11 (21)</td>
</tr>
<tr>
<td>Miguel Servet University Hospital, Zaragoza (Spain)</td>
<td>Tecnis ZXR00 Symfony (ERV)</td>
<td>26 (51)</td>
</tr>
</tbody>
</table>

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
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 3 mm 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 (MTFa, EE). 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 (MTFa, EE) 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:

For pupil 3.0 mm:
\[
MTFa = 40.81 \times EE + 3.36, \quad R^2 = 0.97
\]

For pupil 4.5 mm:
\[
MTFa = 35.42 \times EE + 3.82, \quad R^2 = 0.91
\]

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 MTFapoly(x) and EEpol(x) for the three IOLs and two pupils (3.0 and 4.5 mm) (Fig. 4). For each IOL, the similarity between its MTFapoly(x) and EEpol(x) curves is maintained.

Following the methodology outlined in closely related studies [3,4], we represent the clinical VA(x) outcome versus the MTFapoly(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 EEpol(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 Acrixa Reviol Tri-ED, trifocal apodized FineVision, and ERV Tecnis ZXR00 Symfony). To study possible correlations between clinical VA and on-bench MTFapoly and EEpol 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:

For R
\[
VA = \frac{2.361}{MTFa_{pol} - 1.896} - 0.098, \quad R^2 = 0.94
\]

For G
\[
VA = \frac{0.065}{EE_{pol} - 0.060} - 0.1, \quad R^2 = 0.85
\]

For B
\[
VA = 1.828 \times e^{-0.230 \times MTFa_{pol}} + 0.014, \quad R^2 = 0.94
\]

\[
VA = 0.707 \times e^{-7.746 \times EE_{pol}}, \quad R^2 = 0.90
\]

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. 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). Red, green and blue 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).
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). Red, green and blue solid lines correspond to blue (455 nm), green (530 nm) and red (625 nm) illumination, respectively.
Fig. 3. Representation of \( [\text{MTFa, EE}]_x \) 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). Red, green and blue 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) \( \text{MTFa}_{\text{poly}}(x) \) and (b) \( \text{EE}_{\text{poly}}(x) \) with 3.0 mm pupil and (c) \( \text{MTFa}_{\text{poly}}(x) \) and (d) \( \text{EE}_{\text{poly}}(x) \) with 4.5 mm pupil.
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

Fig. 5. Clinical Visual Acuity versus (a) $MTF_{a,poly}$ 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.
[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 predictor 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 \[R^2 = 0.94, \text{Eq. (6)}\] 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, MTFapoly and EEpoly metrics. The mathematical functions that fit the data of Fig. 5 with high correlation coefficients \[R^2 = 0.94\] for VA versus MTFapoly \[- \text{Eq. (6)}\] and \[R^2 = 0.90\] for VA versus EEpoly \[- \text{Eq. (7)}\] 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 MTFapoly and EEpoly (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 versus MTFapoly or EEpoly [Fig. 5(a) and 5(b)] reflected this constraint in the achievable VA, i.e., the largest MTFapoly and EEpoly values should correspond with the best achievable VA. In other words, it is a realistic assumption that for large MTFapoly and EEpoly values the fitting functions tend to the best VA achievable in the clinics. In practice, to fit the patients’ VA versus MTFapoly or EEpoly [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 \[\approx -0.1\] 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 MTFapoly or EEpoly metrics [Eqs. (6) and (7)] an asymptote value of VA of zero logMAR [Eq. (7)] or very close to zero 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.

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

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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