Estimation of the depth of focus from wavefront measurements

Fan Yi
Contact Lens and Visual Optics Laboratory, School of Optometry, Queensland University of Technology, Australia

D. Robert Iskander
Contact Lens and Visual Optics Laboratory, School of Optometry, Queensland University of Technology, Australia

Michael J. Collins
Contact Lens and Visual Optics Laboratory, School of Optometry, Queensland University of Technology, Australia

It is possible to estimate the depth of focus (DOF) of the eye directly from wavefront measurements using various retinal image quality metrics (IQMs). In such methods, DOF is defined as the range of defocus error that degrades the retinal image quality calculated from IQMs to a certain level of the maximum value. Although different retinal image quality metrics are used, currently there have been two arbitrary threshold levels adopted, 50% and 80%. There has been limited study of the relationship between these threshold levels and the actual measured DOF. We measured the subjective DOF in a group of 17 normal subjects and used through-focus augmented visual Strehl ratio based on optical transfer function (VSOTF) derived from their wavefront aberrations as the IQM. For each subject, a VSOTF threshold level was derived that would match the subjectively measured DOF. Significant correlation was found between the subject's estimated threshold level and the HOA RMS (Pearson's $r = 0.88$, $p < 0.001$). The linear correlation can be used to estimate the threshold level for each individual subject, subsequently leading to a method for estimating individual's DOF from a single measurement of their wavefront aberrations.

Keywords: wavefront aberrations, depth of focus, image quality metrics


Introduction

Depth of focus (DOF) is an important concept in visual science and is used in many aspects of clinical practice such as the prescription of corrective lenses (Bennett, 2008; Selenow, Bauer, Ali, Spencer, & Ciuffreda, 2002) and intraocular lens implant surgery (Schmidinger et al., 2006). It can be simply defined as the variation in defocus, which can be tolerated by the eye without causing any objectionable change in sharpness of the retinal image (Wang & Ciuffreda, 2006).

The traditional goal of vision correction is to provide an optimal level of foveal acuity and contrast sensitivity. For young eyes with active accommodation, achieving a high level of vision performance for far vision allows similar levels of performance to be achieved at a range of distances from far to near. However for presbyopes, the optimal correction of far vision will obviously be inadequate at near distances. This problem is normally solved by supplementary near vision correction but is also partly compensated by the DOF of the eye.

It has been reported that the DOF of the human eye is influenced by refractive error, with myopes showing slightly greater DOF than emmetropes (Collins, Buehren, & Iskander, 2006; Vasudevan, Ciuffreda, & Wang, 2006). This could be due to higher levels of higher order aberrations (HOAs) in myopes (He et al., 2002) or a difference in sensitivity to blur in myopes (Rosenfield & Abraham-Cohen, 1999; Thorn, Cameron, Arnel, & Thorn, 1998).

The DOF of the human eye is also known to increase with age, with presbyopes shown to have higher DOF than young subjects (Nio et al., 2000). These differences are thought to arise from pupil constriction and increased levels of HOA associated with increased age (Artal, Berrio, & Guirao, 2002; McLellan, Marcos, & Burns, 2001). Some forms of optical correction of presbyopes deliberately attempt to increase the DOF by introducing higher order aberrations, such as spherical aberration, to the retinal image. The so-called “simultaneous vision” bifocal contact lenses produce variations in power across the entrance pupil or optical zone of the lens, to create an increased DOF (Plakitsi & Charman, 1995).

The DOF can be assessed using a variety of objective and subjective methods based on a range of different criteria (Atchison, Charman, & Woods, 1997; Marcos, Moreno, & Navarro, 1999). The most frequently used criteria include decrease of visual acuity, perception of just detectable image blur, and loss of visibility of target
The subjective DOF is typically larger than the DOF measured objectively (Vasudevan, Ciuffreda, & Wang, 2007). Because of different stimulus and methodologies adopted, studies have shown a wide range of DOF values (Atchison et al., 1997; Campbell, 1957; Charman & Whitefoot, 1977; Legge, Mullen, Woo, & Campbell, 1987; Marcos et al., 1999; Oshima, 1958).

In recent years, interest has been shown in methods that could estimate DOF from retinal image quality metrics (IQMs) derived from the ocular wavefront aberration (Jansonius & Kooijman, 1998; Legge et al., 1987; Marcos et al., 1999). In such methods, DOF is defined as the range of defocus error that degrades the retinal image quality calculated from the IQMs to a certain level of the possible maximum value. Although different retinal image quality metrics are used, currently there have been two arbitrary threshold levels adopted, 50% (Jansonius & Kooijman, 1998; Legge et al., 1987) and 80% (Marcos et al., 1999). Little justification has been given for the relationship between those estimated and the measured DOF.

The aim of this study was to estimate the threshold level for IQMs, which would correlate with the subjectively measured DOF and lead to a method for estimating DOF directly from a single measurement of wavefront aberration.

**Subjects and methods**

**Subjects**

The experiment was performed on 17 adult subjects (9 males and 8 females) from students and staff members of the School of Optometry, Queensland University of Technology. The mean age of the subjects was 30, ranging from 18 to 46 years. The group had a mean spherical equivalent refraction error of −0.95 D (ranging from −5.0 D to +1.0 D) and the mean cylindrical refraction was −0.32 D (ranging from 0 D to −0.5 D). All subjects had a Snellen visual acuity of at least 6/6 in the tested eye with their best correction. All subjects reported to have no history of significant eye diseases. The subjects gave written informed consent and the study met the requirements of the University Human Ethics Committee and was conducted in accordance with the Declaration of Helsinki.

**Apparatus**

A customized wavefront sensing system was constructed to measure the eye’s wavefront and DOF under different target vergences. The optical layout of the wavefront sensing system, which is based on the HASO32 Hartmann Shack wavefront sensor (Imagine Eyes, Orsay, France) is shown in Figure 1. In our pilot studies, the HASO32 wavefront sensor was calibrated and bench-marked against a Complete Ophthalmic Analysis System (COAS, Wavefront Science) and showed high correlation and good repeatability.

In the wavefront operation channel is a 10-D achromatic microscopic lens L1 with its back focal point located at the eye’s entrance pupil. Lenses L5 and L4 as well as L3 and L2 are set up in an afocal form, which produce an image of the experiment target on the back focal point of L2. The image then acts as the object of Badal lens L1 and its distance to L1 is controlled by the movement of the Badal stage. The Badal stage is based on a 300-mm-long...
travel stage driven by a fine tuning knob. In this optical setting, moving the object every 1 cm brings approximately 1 D of change in the target vergence. The target used in the experiment consists of a Snellen letter chart printed on a piece of clear transparent glass, which is attached to a piece of diffused film and back illuminated by a distant 633-nm LED light source. The target’s contrast is 80% with a luminance of approximately 600 cd/m². During the test, the subject is asked to focus on the letter in the middle of the first line of the letter chart. Through the optics, the letter size produces a visual angle of approximately 20 min of arc (0.60 logMAR detail, similar to reading print of 12 point font size at a distance of 40 cm away).

Protocol

The subject’s head was comfortably positioned in an adjustable, heavy, custom-made headrest without a bite bar. The head’s position with respect to the wavefront sensing system could be adjusted in three dimensions by the operator.

Before the commencement of the measurements, all subjects were given a short training on the system to allow them to become familiar with the task of recognizing the “just noticeable blur”, which was defined as the first detectable sign of changes in the clearness and sharpness of the displayed target. Then, the subject’s tested eye was cycloplegic and dilated by 2 drops of cyclopentolate HCl (1% Minims, 0.5 ml, Bausch & Lomb Australia). The measurement then started about 30 min later, after the maximum pharmacological effect of cyclopentolate was reached (Manny et al., 1993). The subject’s defocus level was controlled by moving the Badal stage. The operator adjusted the position of the Badal stage to approximately compensate the subject’s subjective defocus. The astigmatism derived from the individual subjective refraction was corrected with a trial lens mounted in front of the artificial pupil, while the fellow eye was fully occluded by a black eye patch. In the experiments, two pupil diameters were considered, 5 mm and 3.5 mm, to simulate the viewing under mesopic and photopic conditions. The subject was instructed to identify the “clear” position of the displayed target. The subject was asked to fixate on the target through an adjustable, heavy, custom-made headrest without a bite bar. The head’s position with respect to the wavefront sensing system could be adjusted in three dimensions by the operator.

Under full cycloplegia and pupillary dilation, the subject was asked to fixate on the target through an artificial pupil, while the fellow eye was fully occluded by a black eye patch. In the experiments, two pupil diameters were considered, 5 mm and 3.5 mm, to simulate the viewing under mesopic and photopic conditions. The subject was instructed to identify the “clear” position (corresponding to the subjective best focus) and “just noticeable blur” in both negative and positive directions, corresponding to the movement of the Badal stage toward and away from the eye.

The procedure for measuring the subjective DOF was given as follow. First, the operator adjusted the position of the Badal stage to the subject find a “clear” position in which the target could be viewed as clear and sharp as possible. Then, the operator slowly moved the Badal stage in one randomly selected direction until the “just noticeable blur” was reported by the subject. The scale reading of the Badal stage was recorded by the operator. The operator then moved the Badal stage in the opposite direction. During the movement, the subject observed the “clear” position again, and as the movement continued, the subject observed the appearance of “just noticeable blur”. The scale reading of this position was also recorded. These two limits of Badal stage movement constituted one measurement of DOF. For each pupil diameter, five sets of DOF measurements were performed. To avoid the possibility that the subject may remember the time it took to observe the “just noticeable blur” away from the “clear” position, the operator moved the Badal stage at a variable speed, and the moving speed was controlled to be less than approximately 0.2 D/s. At the end of the experiment, the subject’s accommodative response was examined to ensure that there was no significant (≤0.1 D) recovery of accommodation.

The ocular aberrations were also recorded by taking 10 wavefront measurements at each position (toward and away from the eye) when the “just noticeable blur” was observed by the subject (total 20 measurements). The higher order aberration components did not change significantly across the defocus range. Wavefront measurements were performed with the artificial pupil removed for the fully dilated pupils. The higher order aberration components of the wavefront data were then averaged and used for computing the visual Strehl ratio based on the optical transfer function (OTF), which was later used as an image quality metric for matching the subjective DOF.

Before commencing each set of measurements, the pupil position was checked by comparing the pupil positions on the sensor CCD with and without the artificial pupil in the HASO control software (Imagine Eyes, Orsay, France). The measurement had a resolution of 0.01 mm. If the displacement of the pupil was greater than 0.3 mm, then the position of the subject’s head was corrected by the operator.

Determination of the threshold for estimating DOF from wavefront data

One can estimate the theoretical DOF by calculating the range of defocus errors, which degrades the retinal image quality to a certain level of the possible maximum value. This definition has been adopted earlier by Marcos et al. (1999), who chose an 80% threshold, while a 50% threshold was used by Jansonius and Kooijman (1998) and Legge et al. (1987). In this study, we chose the threshold based on the optical transfer function (OTF) as the retinal image quality predictor to estimate the matching threshold based on the subjectively measured DOF.

The VSOTF is currently considered one of the best descriptors of visual performance that can be directly
derived from the wavefront aberrations data (Marsack, Thibos, & Applegate, 2004) and is strongly correlated with the subjective visual acuity (Cheng, Bradley, & Thibos, 2004). We have used its augmented version (Iskander, 2006), i.e.,
\[
VSOTF = \frac{\int_{-\infty}^{\infty} \int_{-\infty}^{\infty} CSF_N(f_x, f_y) \cdot \Re \{ OFT(f_x, f_y) \} \, df_x df_y}{\int_{-\infty}^{\infty} \int_{-\infty}^{\infty} CSF_N(f_x, f_y) \cdot OFT_{DL}(f_x, f_y) \, df_x df_y},
\]
where \(OFT_{DL}(f_x, f_y)\) denotes the diffraction limited optical transfer function, \(CSF_N(f_x, f_y)\) is the neural contrast sensitivity function, and \((f_x, f_y)\) are the spatial frequency coordinates. Here the VSOTF was based on calculated optical transfer function across all spatial frequencies up to 60 cycles per degree (Iskander, 2006).

To estimate DOF from an image quality metric, a through-focus calculation is required. A dedicated simulation program was written from first principles in Matlab (The MathWorks, Natick, MA) to calculate the through-focus VSOTF in the presence of the subject’s original higher order aberrations (HOAs). The flow chart of the computer simulation program is shown in Figure 2.

In the first step, wavefront data, consisting of a set of Zernike coefficients up to and including the 8th radial order, are imported. Since the wavefront data was acquired for the subject’s dilated pupils always larger than 5 mm, for consistency, in step 2, the original Zernike coefficients were resampled to a specific pupil diameter of either 5 mm or 3.5 mm using the method of Schwiegerling (2002).

Since the subject’s sphero-cylindrical error was corrected during the subjective DOF measurements, only the effect of HOAs on VSOTF is considered in the simulation. The estimates of sphero-cylinder need to be first removed from the wavefront. One can achieve that by simply setting the first six Zernike coefficients to zero. However, it has been shown that the Maloney’s best sphero-cylinder (S/C) calculated in the refractive power domain has the best correlation to the subjective sphero-cylindrical refractive error of the eye (Iskander, Davis, Collins, & Franklin, 2007). Hence, a transformation from the wavefront domain to the refractive power domain is performed. In step 3, the refractive power distribution across the pupil, \(P(r, \theta)\), is calculated from the resampled wavefront \(W(r, \theta)\) using the method of the refractive Zernike power polynomials (Iskander, Davis, Collins, & Franklin, 2007):

\[
P(r, \theta) = Z\{W(r, \theta)\},
\]
where \(Z[\cdot]\) denotes the wavefront to refractive power transformation.

Following that, in step 4, the best S/C is estimated using the method of Maloney, Bogan, and Waring (1993) and subtracted from the previously obtained refractive power. This leads to the new refractive power, given by

\[
F_{out} = F_{Zer} - F_{SC},
\]
where \(F_{Zer}\) and \(F_{SC}\) is the refractive power calculated from the subject’s original wavefront and the estimated best S/C, respectively. To simulate through focus, in the through-focus loop, a desired level of defocus is added to the refractive power from step 4. In step 5, an inverse transformation from the refractive power domain to the wavefront domain is performed (Iskander, Davis, & Collins, 2007):

\[
W_{out}(r, \theta) = Z^{-1}\{F_{out}(r, \theta)\},
\]
which is then used, in step 6, to calculate the VSOTF. From the wavefront $W_{\text{out}}(r, \theta)$ with a new defocus value, the corresponding point spread function and the optical transfer function (OTF) is calculated using fast Fourier transforms (Artal, 1990; Iskander, Collins, Davis, & Carney, 2001). The through-focus VSOTF is obtained in step 7. The calculation was repeated in a total of 49 steps corresponding to a defocus level ranging from $-3$ D to $+3$ D in 0.125-D intervals.

An example of how the matching threshold value is estimated for data acquired from averaged wavefront measurements of a subject in a 5-mm pupil is shown in Figure 3. After obtaining the through-focus VSOTF of the subject from wavefront data, an iterative calculation was performed, reducing the threshold level from 99% of the maximum achievable VSOTF value, until the effective range of defocus error produced by $D_2 - D_1$ gives the closest match to the subjectively measured DOF. This threshold value was taken as the matching threshold to estimate the DOF for this subject. The same procedure was performed for measurements of each individual subject.

### Statistical analysis

Averages are represented in terms of mean ± SD (standard deviation). Collected data including subjective DOF, individual matching thresholds, and HOA RMS in both 5-mm and 3.5-mm pupils were tested for normal distribution. For correlating the estimated VSOTF threshold values with other measures of retinal image quality, Pearson’s correlation coefficient was calculated.

### Results

We estimated the individual matching threshold of 17 subjects using the algorithm described in Figure 2. Data including the subjective DOF, the matching threshold, HOA RMS, and spherical aberration were collected for both 5-mm and 3.5-mm pupil diameters. The group mean values were shown in Table 1. The subjective DOF measured in our experiment ranged from 0.55 D to 1.05 D, with a mean value of 0.79 ± 0.15 D, in a 5-mm pupil. When the pupil

<table>
<thead>
<tr>
<th>Pupil size</th>
<th>Subjective DOF (D)</th>
<th>Matching threshold (%)</th>
<th>HOA RMS (µm)</th>
<th>Z(4,0) (µm)</th>
<th>DOF (D) for a 50% threshold</th>
<th>DOF (D) for an 80% threshold</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 mm</td>
<td>0.79 ± 0.15</td>
<td>65.6 ± 10.1</td>
<td>0.30 ± 0.08</td>
<td>0.075 ± 0.062</td>
<td>1.12 ± 0.34</td>
<td>0.58 ± 0.17</td>
</tr>
<tr>
<td>3.5 mm</td>
<td>1.30 ± 0.21</td>
<td>36.9 ± 18.4</td>
<td>0.12 ± 0.05</td>
<td>0.020 ± 0.015</td>
<td>1.07 ± 0.54</td>
<td>0.52 ± 0.24</td>
</tr>
</tbody>
</table>

Table 1. Group average results in 5-mm pupil and 3.5-mm pupil diameters.
diameter was limited to 3.5 mm, the mean DOF increased to 1.30 ± 0.21 D, while the total HOA RMS and spherical aberration reduced compared to those in a 5-mm pupil. The group means of the individual threshold estimated from the through-focus VSOTF were 65.6 ± 10.1% (ranged from 45% to 83%) and 36.9 ± 18.4% (ranged from 15% to 83%) in a 5-mm pupil and a 3.5-mm pupil, respectively.

To estimate DOF directly from wavefront measurements in a robust manner, correlation analysis was performed between subjective DOF and HOA RMS, subjective DOF and SA, matching threshold (from the through-focus VSOTF) and HOA RMS, and estimated threshold and SA. For a 5-mm pupil diameter, weak correlation was found between the subjective DOF and HOA RMS (\(r = 0.36, \ p < 0.05\)) and between subjective DOF and SA (\(r = 0.24, \ p < 0.05\)). The matching threshold showed significant correlation with the total HOA RMS (Pearson’s \(r = 0.88, \ p < 0.001\)). Moderate correlation was shown between the estimated threshold and the spherical aberration value in the eye (\(r = 0.52, \ p = 0.05\)). For a 3.5-mm pupil diameter, there was no significant correlation observed between the DOF and HOA RMS. There was weak correlation between DOF and SA (\(r = 0.49, \ p > 0.05\)). No correlation was found between the estimated threshold and the spherical aberration value (Pearson’s \(r = 0.36, \ p > 0.05\)). However, significant correlation was found between the estimated threshold and the HOA RMS (Pearson’s \(r = 0.62, \ p < 0.05\)).

It was found that the DOF threshold and HOA RMS (shown in Figure 4a, with 95% confidence bands) has the strongest correlation (Pearson’s \(r = 0.88, \ p < 0.001\)) in a 5-mm pupil. By fitting a linear function to the data, we obtained

\[
\text{Threshold Level}_{\text{predicted}} = 106.86 \times (\text{HOA RMS}) + 33.99 \pm [12.26 + 107.05 \times (\text{HOA RMS} - 0.30)^2].
\]  
(5)

This equation (including 95% confidence intervals) can be used to calculate the individual threshold level for estimating the DOF using VSOTF from wavefront measurements in subjects with normal amount of HOA.

Since the astigmatism correction by the trial lens had a limited precision of 0.25 D, it was also of interest to investigate whether the presence of the residual astigmatism can significantly affect this result. Accordingly, we have performed additional calculations in which we first found the spherocylindrical difference between the trial lens astigmatic correction and the one measured with the wavefront sensor (note that the wavefront aberrations were measured without the trial lens) and then retained the astigmatic difference (residual astigmatism) in the VSOTF calculation. To find the difference, we have transformed the two spherocylinder values to orthogonal components, subtracted them, and transformed those differences back to a spherocylindrical representation. After leaving the residual astigmatism in the through-focus simulation, the correlation between the estimated VSOTF threshold and the HOA RMS value was still significant but dropped from the original \(r = 0.88, \ p < 0.001\) to \(r = 0.77, \ p < 0.003\).

Since the HOA RMS value is a pupil-plane-based IQM, we also examined the correlation between the estimated threshold and the VSOTF value (at zero diopters of defocus), which is known to be a good representation of retinal image quality. However, for a 5-mm pupil diameter, only moderate but significant correlation was found between the estimated DOF threshold and VSOTF at zero defocus (\(r = 0.68, \ p = 0.025\)).

The DOF estimated from through-focus VSOTF using fixed thresholds (i.e., 50% and 80%) was also calculated and shown in Table 1. The group mean of the estimated DOF calculated with a fixed threshold of 50% and 80% were 1.12 ± 0.34 D and 0.58 ± 0.17 D in a 5-mm pupil and 1.07 ± 0.54 and 0.52 ± 0.24 in a 3.5-mm pupil.
Discussion

We developed a method to estimate the individual threshold from through-focus VSOTF for calculating the DOF of normal subjects from their wavefront aberrations. The threshold estimating method was based on the subjective DOF measurements of real subjects using the defining criterion of “just noticeable blur”. Therefore, it provided practical validation that the DOF estimated from wavefront aberrations can correlate with the DOF measured subjectively.

The DOF subjectively measured for the subjects in our experiment ranged from 0.55 D to 1.05 D with a mean value of 0.79 ± 0.15 D and ranged from 0.80 D to 1.61 D with a mean value of 1.31 ± 0.21 D, in a 5-mm pupil and a 3.5-mm pupil, respectively. These values match well with the range of results found in young subjects from 0.8 to 1.2 D (in a pupil size ranging from 3 to 5 mm) as reported by Ogle and Schwartz (1959), Tucker and Charman (1975), and Wang and Ciuffreda (2004). Ogle and Schwartz’s measurements were based on 50% probability of resolving a 20/25 checker board. Tucker and Charman’s measurements were based on 80% probability of achieving 90% of the optimal Snellen acuity. In Wang and Ciuffreda’s (2004) study, the subjects viewed through a dual-channel Badal optical system and made judgments of when the test target showed “just noticeable blur”.

Using different definitions of DOF are likely to affect the measured value of subjective DOF (Wang & Ciuffreda, 2006). In clinical or research applications, the range of defocus that decreases the visual acuity or contrast sensitivity to a certain limit is often used as a criterion for DOF (Legge et al., 1987; Ogle & Schwartz, 1959; Tucker & Charman, 1975). For real life scenarios, the perception of “blur” can be considered to be a more relevant criterion (Atchison et al., 1997; Campbell, 1957). Atchison, Fisher, Pedersen, and Ridall (2005) defined three levels of blur limits as: “noticeable”, “troublesome”, and “objectionable”. The authors found that the magnitudes of “troublesome” and “objectionable” limits were approximately 1.6–1.8 times and 2.1–2.5 times greater than the “noticeable” limits, respectively. We chose the widely adopted criterion “just noticeable blur” in our experiment to measure the subjective DOF, but it is expected that a larger DOF would be obtained if criteria of “troublesome blur” or “objectionable blur” were chosen.

The choice of image quality metrics to estimate the depth of focus will influence the predicted outcomes, but metrics calculated at the retinal image plane are thought to be superior to those in the pupil plane for predicting subjective refraction (Thibos, Hong, Bradley, & Applegate, 2004). To study the predicted DOF, we used the augmented visual Strehl ratio of the OTF (VSOTF) as the retinal image quality metric covering overall spatial frequencies up to 60 cpd. VSOTF has been found to correlate well with subjective visual performance in a number of studies (Cheng et al., 2004; Guirao & Williams, 2003). When calculated in through focus, it represents the interaction between HOA and defocus on retinal image quality (Collins et al., 2006). In our study, a strong correlation was found between DOF threshold and HOA RMS in a 5-mm pupil. Known as a better representative of retinal image quality, the VSOTF at zero defocus was expected to have better correlation to the DOF threshold. However, a weaker but still significant correlation was observed between the DOF threshold and the VSOTF value at zero diopters. This may be due to the fact that for most of the subjects, the peak value of VSOTF does not locate at zero defocus level.

The frequency-dependant features of the DOF was not investigated in our study (our target contained a range of spatial frequencies) but had been extensively studied by other groups (Atchison et al., 1997; Legge et al., 1987; Tucker & Charman, 1986).

We have shown that using a fixed IQM threshold (e.g., 50% or 80%) to estimate the DOF may produce results significantly varying from the subjectively measured DOF. In our study, the estimated DOF from through-focus VSOTF with a 50% threshold level had an average error of 0.33 ± 0.25 D and 0.55 ± 0.34 D in a 5-mm pupil and a 3.5-mm pupil, respectively, compared to the subjective DOF. Calculating the DOF with an 80% threshold averaged underestimated the DOF by 0.21 ± 0.15 D and 0.80 ± 0.32 D, in a 5-mm pupil and a 3.5-mm pupil, respectively. In general, use of fixed thresholds caused larger errors for the DOF estimation in a smaller pupil.

Our method to estimate the threshold for calculating the DOF from wavefront aberration was affected by the subject’s pupil size. The strong correlation between the matching threshold level and HOA RMS was only observed in a larger (5 mm) pupil. When pupil size was restricted to 3.5 mm, the eye’s blur circle was reduced. The magnitude of specific dominant HOA terms (such as spherical aberration and coma) were also significantly lower than that in a 5-mm pupil. These changes will significantly influence the details of the calculated through-focus IQMs and, therefore, affect the accuracy of threshold and DOF estimation. The matching threshold estimation method is also limited by the range of HOA RMS. It can be applied to predict the DOF of subjects with normal amount and structure of HOA (Porter, Guirao, Cox, & Williams, 2001; Wang & Koch, 2003). For the eyes of keratoconic subjects or patients who have undergone refractive surgery, their significantly higher amount of HOA may also affect the accuracy of the method or simply exceed the predictable range.

Our subjective measurements and estimating were all performed in monochromatic light. In natural scenes, the
chromatic aberrations in the human eye will also affect the DOF (Campbell, 1957). Legge et al. (1987) used the method described by Van Meeteren (1974) to calculate the depth of focus for monochromatic and white light at different spatial frequencies and pupil sizes. A very small increase was found for white light DOF compared to the one calculated for monochromatic light. Experimental measurements also showed only small differences (Campbell, 1957).

In conclusion, we have shown that the IQM threshold level used to theoretically estimate the DOF from wavefront aberrations can be adaptively optimized for each individual subject, and this method is most reliable with larger pupils (i.e., 5-mm pupil diameter). Using a fixed threshold level to estimate the DOF in different subjects or for DOF of the same subject in different pupil sizes may lead to erroneous estimates.

**Acknowledgments**

The authors wish to thank the anonymous reviewers for their very helpful comments on an earlier draft of this paper.

Commercial relationships: none.
Corresponding author: Fan Yi.
Email: f.yi@qut.edu.au.
Address: Contact Lens and Visual Optics Laboratory, School of Optometry, Queensland University of Technology, Victoria Park Road, Kelvin Grove, Qld 4059, Australia.

**References**


time course of cycloplegia using an objective measure of the accommodative response. Optometry and Vision Science, 70, 651–665. [PubMed]


