Influence of ocular aberrations on contrast in retinal dark field imaging

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Received September 23, 2013; accepted October 16, 2013; posted online February 26, 2014

The application of dark-field (DF) technique with adaptive optics for retinal imaging in vivo is presented to improve retinal imaging contrast. The influence on the imaging contrast introduced by dynamic ocular aberrations should be considered. The qualitative analysis of the influence is discussed. Detail simulation and quantitative analysis are presented. According to our simulation results, when residual aberrations are reduced to less than 5%, it has a little impact on the contrast of DF imaging; while, on increasing the aberrations up to 10%, the contrast of the DF imaging falls off sharply.

OCIS codes: 100.2960, 100.1160, 080.1010.

doi: 10.3788/COL201412.S11002.

The eye is the most important sensor for human being, and about 80% of the external information is obtained by eyes. However, many people are suffering from ophthalmopathy. In the medical point of view, most ocular lesions, except for trauma and ametropia, develop in the fundus. In addition, it is proved medically and clinically that diabetes, hypertension, anemia, kidney disease, cancer and many other diseases can have negative effect on the microcirculation in fundus. Real-time tracking of variation in fundus morphology helps in the early diagnosis and prevention of these lesions. Therefore, high-contrast and high-resolution fundus imaging technique has great value in medical research and clinical applications. Adaptive optics (AO) technology is the primary impetus for the development of high-resolution retinal imaging technology[1–6]. In 2002, Roorda *et al.* from Houston University developed the experimental system of Adaptive Optics Confocal Scanning Laser Ophthalmoscopy (AOCSLO)[4] with lateral resolution of 3 μm and longitudinal resolution of 100 μm. In 2004, an experimental system of Adaptive Optics-Optical Coherence Tomography (AO-OCT) was developed[7] from Vienna University and Murcia University with lateral resolution of 5–10 μm and longitudinal resolution of 3 μm. These two systems had resolutions close to the diffraction limit of the pupil, but both were able to discriminate only three out of 10-layer structure of the retina, i.e., photoreceptor cells layer, nerve fiber layer and capillary layer. This is because the difference in refraction index between neighboring layers is rather little, resulting in low-imaging contrast. This predicates that grey levels across different layers of the retinal image are quite similar, which makes discrimination highly difficult. Hence, DF imaging technology was developed by improving the imaging contrast; so, it was widely used in biopolymer imaging[8], biological tissues and cells imaging[9], and lithography[10]. Considering this, it was proposed to apply DF imaging technique in the *in vivo* retinal imaging to discriminate other layers of the retina better[11]. But, the applications of DF imaging technique mentioned above belonging to static imaging or detection, which compensate aberrations in the static way and the influence of aberrations, can be ignored. However, as to retinal imaging, the real-time dynamically ocular aberrations[12–14] changing dynamically may affect, to some extent, the DF imaging result.

In this letter, we have analyzed the effect of ocular aberrations on the contrast of the retinal imaging with DF technique. The qualitative analysis of the influence was also discussed, and detail simulation and quantitative analysis were presented.

The principle of DF imaging technology is to prevent the light directly passing through the specimen into the lens, but only let light scattered by particles into the lens. Thus, there are bright particles with dark background on image plane. DF imaging technology has been widely used in different kinds of imaging fields owing to the advantage of improving the imaging contrast. The experimental platform is based on the fundus camera. The system consisted of two parts, including fundus imaging and the DF device. DF device is composed of an annular aperture and a circular aperture as shown in Fig. 1.

In order to facilitate the analysis, the mathematical model was given for the system as shown in Fig. 2.

Model specification: In the system, the parallel light illuminates the retina and the light reflected by the retina is imaged. In order to simplify the model, we modified it to the transmission type, but the reflected intensity distribution was still substituted in light intensity distribution of the object plane. The optical system of human eyes works as objective lens, and introduces aberrations. Ocular aberrations make a difference by deviating the wavefront on the exit pupil from the ideal wavefront, which is equivalent to adding an aberration surface at the exit pupil position. In the system, annular illumination and circle receiver are adopted to filter

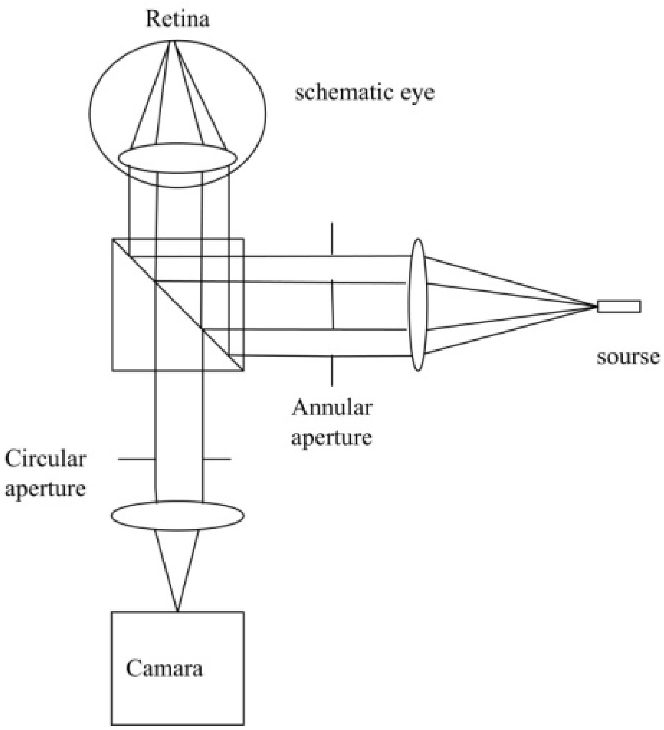


Fig. 1. Schematic diagram of the system.

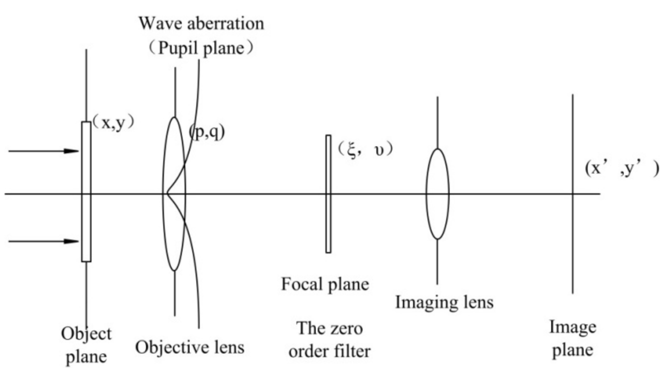


Fig. 2. Mathematical model of the system.

the direct light, i.e., the zero frequency component of the spectrum. In the model, it is equivalent to join a high-pass filter on the frequency plane to remove the zero-frequency light.

In order to understand the DF imaging technique intuitively, we simulated DF imaging process by using Matlab. Taking a target object as input, DF imaging results in the aberration-free case as shown in Fig. 3. As shown in Fig. 3(b), there are bright particles with dark background on image plane.

Considering a statistical Zernike aberration of normal human eye as an original reference aberration surface, DF imaging results with different residual aberrations are obtained. The sample data are derived from measurement results of Hartmann ocular aberration measurement system developed by Key Laboratory on Adaptive Optics, Chinese Academy of Sciences. In the actual situation, human eye defocus and astigmatism could be corrected by inserting compensating plate. Considering this, in the pretreatment, their coefficients

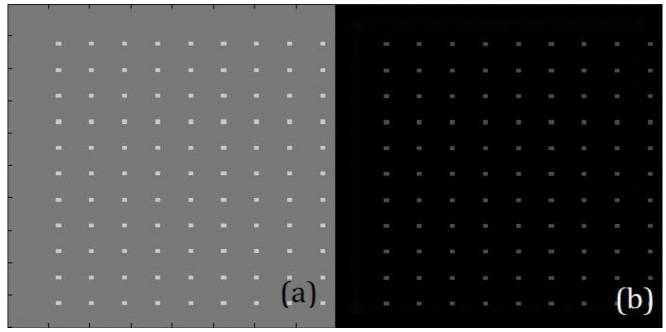


Fig. 3 (a) Input of a target object; (b) Results of dark-field imaging in the simulation.

are multiplied by the constant factor 0.1 in the process of simulation. The histogram of normal human Zernike aberration coefficient and the diagram of wave after pretreatment are shown in Figs. 4 and 5; the Root-Mean-Square (RMS) value is calculated to be 0.5646 μm. The analysis is facilitated by assuming the residual percent of the aberration coefficients to be equal in the later conditions.

DF simulation results under the conditions with four different residual aberrations (taking the wave as a new reference wavefront after pretreatment, different residual value, i.e., multiplied by, respectively, multiplying factor, such as residual weight of 10% represents being multiplied by 0.1) are shown in Fig. 6. By comparing the four pictures, the imaging contrast is reduced by increasing the aberration residual quantity.

The ratio of standard deviation to mean of the image intensity is adopted as the value of an image contrast;

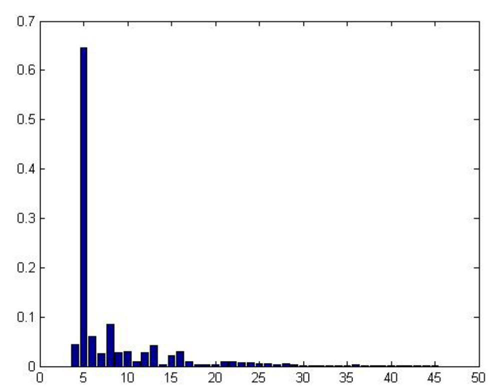


Fig. 4. Histogram of normal human Zernike aberration coefficient after pretreatment.

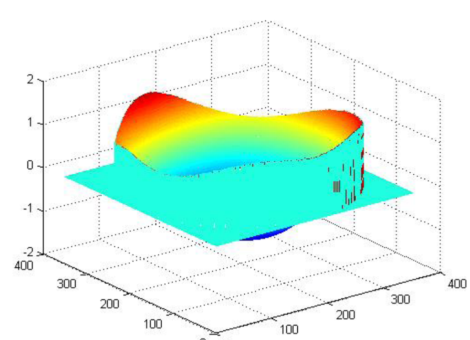


Fig. 5. Diagram of wave after pretreatment.

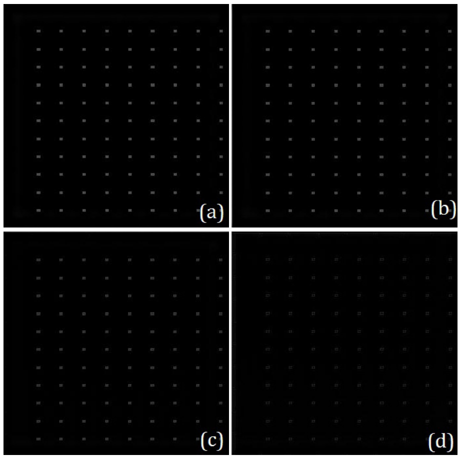


Fig. 6. The simulation results of the effect of aberration on DF imaging. (a) The result of DF imaging with residual value of 1%, (b) with residual value of 5%, (c) with residual value of 10%, and (d) with residual value of 15%, corresponding to the RMS value of 0.0056,0.0282,0.0565, and 0.0847 μm, respectively.

the values gained by the simulation are listed in Table1 and plotted in Fig. 7. As shown in Table 1, increasing residual aberration of human eyes decreases imaging contrast. However, when the residual aberration is very small, the contrast decreases very slowly. Therefore, correcting the whole aberrations in the DF imaging is not required. In Fig. 7, the imaging contrast is reduced with increasing aberration residual quantity, and when the aberration residual exceeds 10%, imaging contrast begins to reduce significantly.

Deformable mirror (DM) is an important part of the AO system; the unit number determines the correction capability of the system. The AO fundus camera in the laboratory was adopted with 37-unit DM. The 37-unit system could only correct for the first 20 order Zernike model of the residual aberrations effectively. According to the situation discussed above, results of the ideal DF imaging are obtained with the first 20 Zernike model corrected to be 0, shown in Fig. 8.

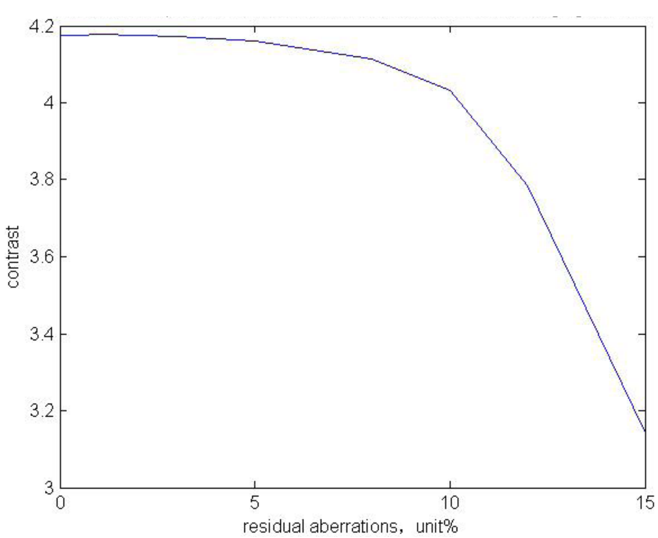
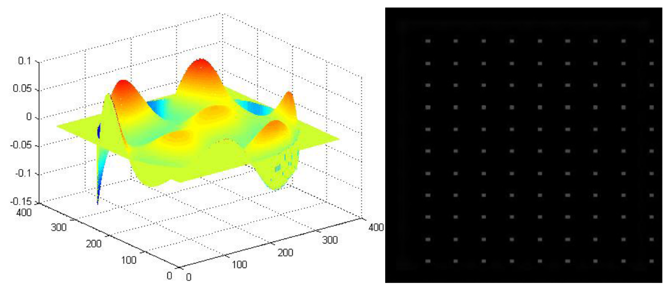


Fig. 7. Relationship between residual aberrations and dark-field imaging contrast.

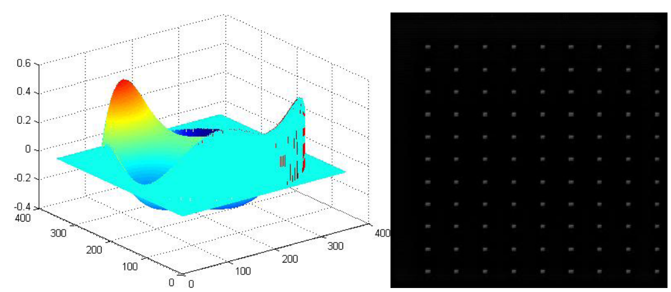
Table 1. Relationship between residual aberration and contrast

Residual value of ocular aberrations(%)	Contrast of object	Contrast of image
0	0.2225	4.1750
0.5	0.2225	4.1755
1	0.2225	4.1767
3	0.2225	4.1733
5	0.2225	4.1598
8	0.2225	4.1136
10	0.2225	4.0320
12	0.2225	3.7848
15	0.2225	3.1403



(a) (b)

Fig. 8. DF imaging for 37-unit system in an ideal condition. (a) Wave with the first 20 Zernike coefficients set at 0. (b) The result of DF imaging with the first 20 Zernike coefficients set at 0. The RMS of the wave in (a) is 0.017 μm; the contrast of the result in (b) is 4.1599.



(a) (b)

Fig. 9. DF imaging for 37-unit system in a reality condition. (a) Wave of the residual aberration in 37-unit system. (b) The result of the DF imaging in 37-unit system. RMS of the wave in (a) is 0.1230 μm; contrast of the result in (b) is 3.7272.

The results, as shown in Fig. 8, are obtained in an ideal condition with the first 20 Zernike coefficients set at 0, which could not be achieved in a real system where residual aberrations always exist. In the closed-loop tests of the real system, RMS of the residual aberration is less than 0.1266 μm. These tests showed that the 37-unit system could tackle ocular aberrations with defocus in dominance, and the correction ability for defocus was tested. The residual defocus could be less than 0.018 μm[15]. The residual coefficient of the first 20 Zernike aberrations can be calculated based on the data given above and the assumption that the residual percent of the aberration coefficients is set to be equal. The RMS of the residual aberration is 0.1230 µm. The result is shown in Fig. 9.

In conclusion, we simulate DF imaging process and analyze the effects of ocular aberrations on DF imaging contrast. The DF imaging technique can improve the contrast of retinal imaging, but increasing the residual aberration reduces the contrast. When residual aberrations are reduced to less than 5%, impact on the contrast of DF imaging is very little. While, when it increases up to 10%, the contrast of the DF imaging fall offs sharply. The results indicate the aberration tolerance for retinal DF imaging technique in vivo and present the specific technical requirements for introducing the DF to the field of retinal imaging.

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