Estimation of the ocular point spread function by retina modeling

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Retinal imaging often suffers from blurring aberrations. With knowledge of the blurring point spread function (PSF), better images can be reconstructed by deconvolution techniques. We demonstrate a method to enhance the contrast of retinal cells by estimating the ocular PSF. This is done by finding the cells' positions and their intensity distribution, and using these as a model for the image. The feasibility of this method is demonstrated by Wiener deconvolution both for adaptively and non-adaptively corrected images.

Keywords: 330.5310, 330.6100, 100.1830

High resolution retinal imaging is one of the direct tools to examine the retina. Thus high quality images of the retina could help shed more light on the retinal cellular structure and biological processes. Unfortunately when in vivo retinal images are needed, the acquired retinal images are often blurred and low in contrast. The degradation in the image quality is mostly caused by the ocular aberrations, in addition to scattering from the ocular optics and inner retinal layers.

Unlike imaging through the atmosphere, where the effect of aberration can be well modeled [1], the variations of ocular aberrations between subjects, and their dynamic changes, prevent reduction of aberration by conventional optics [2,3]. The implementation of adaptive optics (AO) retinal imaging helped to overcome most of these hurdles. Consequently images at cellular level resolution are obtained by AO-based setups [4], especially when aided by confocal scanning. That said, the wave front correction by those setups might be less than perfect, and the residual deviations still blur the image.

In addition to the improvement obtained by the optical schemes, reduction of the blurring is often achieved by image deconvolution methods. The instantaneous PSF can be estimated from the wave front measurement and its reconstruction [5]. When the PSF is immeasurable, the method of choice is blind deconvolution: an initial conjecture of the PSF is improved by constraint-based algorithms, as well as the reconstructed image [6-11]. However, these methods are easily biased by the initial guess, which may lead into poor image reconstruction.

In this work a method to estimate the PSF out of cells in the retinal image, is demonstrated. By modeling the photoreceptors layer, the main retinal feature (except for blood vessels), the PSF that deformed these cells can be estimated. The feasibility of this method will be demonstrated by using the estimated PSF as the degradation element in a minimum mean square error filtering (Wiener filter). This method will be applied both on images obtained by AO, and by direct imaging enhanced by weighted shift-and-add [12].

In the case of multi-frame images, made of m frames of the same object, the image formation can be written as a convolution

$$\sum_{j=1}^{m} i_j(\mathbf{r}) = \sum_{j=1}^{m} p_j(\mathbf{r}) \otimes o(\mathbf{r}) + n_j(\mathbf{r}), \qquad (1)$$

where $i_j(\mathbf{r})$ is one of the frames, $p_j(\mathbf{r})$ is its PSF, $o(\mathbf{r})$ is the object intensity distribution, and $n_j(\mathbf{r})$ is the frame noise. When we wish to replace the object $o(\mathbf{r})$ with a model $c(\mathbf{r})$ for the cells, we need to introduce an extra term $b(\mathbf{r})$ on the left. This term consists of the information in the image which cannot be acquired by the convolution of the PSF with the cells' model and by the additive noise,

$$\sum_{j=1}^{m} i_j(\mathbf{r}) - b_j(\mathbf{r}) = \sum_{j=1}^{m} p_j(\mathbf{r}) \otimes c(\mathbf{r}) + n_j(\mathbf{r}). \qquad (2)$$

When analyzing a small area of the photoreceptor layer, with no other visible features, $b(\mathbf{r})$ represents the flat background light, that is reflected back from and behind the intermediate cells area. In the small areas that are analyzed, this background can be regarded as a constant. By transforming Eq. (2) into the Fourier domain the averaged PSF can be written as

$$\sum_{j=1}^{m} P_j(\boldsymbol{\omega}) = \frac{\sum_{j=1}^{m} I_j(\boldsymbol{\omega}) - b_j \delta_j(\boldsymbol{\omega}) - N_j(\boldsymbol{\omega})}{C(\boldsymbol{\omega})}, \qquad (3)$$

where capital letters represent the Fourier transform of the corresponding terms and $\delta_j(\boldsymbol{\omega})$ is a Dirac delta function. Therefore by transforming it back into the image domain, the effective PSF, $p_e(\mathbf{r}) = \Sigma p_i(\mathbf{r})$, can be found, effective because it is found with respect to the model $c(\mathbf{r})$ which serves as a proxy for the object $o(\mathbf{r})$. To avoid dividing by infinitesimally small values of $C(\boldsymbol{\omega})$, at the image's diffraction limit, we modeled $\boldsymbol{c}(\boldsymbol{r})$ with a cell's diameter that is smaller by a few percent from the cells in the image. Moreover, by taking images with relatively high signal to noise ratio (a limit will be given later), the relevant noise level is estimated at frequencies above the image's diffraction limit. This causes the numerator of Eq. 3 to drop to

zero before the denominator does. Retinal power spectra, consisting of cells and background, are positive up to the diffraction limit [13-15]. The constant background, $\Sigma\, b_j$, only reduces the PSF by a constant term, forming negative side lobes in the estimated PSF. As noise still showed up as distant lobes of the estimated PSF, we used only the PSF area which corresponds to zero and first orders of an airy pattern for the reconstruction process.

The cells' model c (**r**) is constructed by finding the cells' positions in the image [16] and replacing each position by circular disc (Fig. 1). The diameter of all discs is determined by the half-height width of their averaged radial intensity profile. Since rod photoreceptors (~2.7 μm diameter) are near or below the diffraction limit of the imaging setup, they were ignored in this model. Thus all the resolved cells in the image can be regarded as cones, having approximately the same size in the image. Next, the intensity of each disc is determined from the radial average of each cell. Fine tuning in the disc diameter may be required in order to obtain a better PSF estimate, as measured by a better image reconstruction.

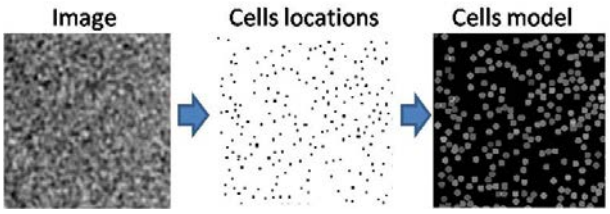

Fig 1. Cells' model construction. By using the image as an input (left), each cell's position is found (center) and is replaced by a fixed circular disc of its averaged radial intensity (right).

In order to check the method, we first used a synthetic image (Fig. 2), made of circular discs at positions derived from a retinal image (taken from Fig. 3). First, the image was used to find the corresponding PSF, which should be similar to a Dirac delta function (Fig. 2a and 2b). Small deviations were caused by failure of the locating algorithm to differentiate few adjacent cells at the left section of the image. Next, we convolved the synthetic image with a blurring function (Fig. 2c) and added Gaussian noise to it (Fig. 2d). We found an almost perfect PSF estimate for signal to noise ratio as low as 16.3 (Fig. 2e).

Fig. 3 shows an AO image and the corresponding PSF estimate. In addition, we tried to reconstruct the image using a standard Wiener filter. Since this image was good to begin with, the improvement is less significant with comparison to the direct imaging images below.

Fig. 4a shows a radially averaged power spectrum comparison between the AO image (Fig. 3a) and the reconstructed image (Fig. 3b). Each spectrum was normalized by the average value of the five lowest frequencies. The result of this analysis shows improvement in the power of median and high frequencies of the reconstruct-

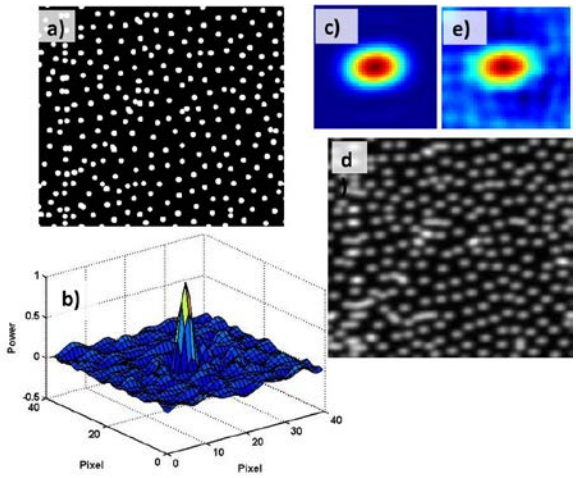

Fig. 2 (Color online): Testing the method with an artificial image. First the method is applied to an ideal image (a), so the PSF estimate is similar to Dirac delta function (b). Next, a known blurring core (c) is convolved with the image and a Gaussian noise was added (d) in order to check the correctness of the estimation (e)

ed image. The biggest improvement was found in 0.038 cycle/pixel, corresponding to a cycle of 26.3 pixels. By measuring the distance of various adjacent cells, we found average inter-cell distance of 26.5 pixels, thus the deconvolution process did increase the power of cells in the image. In addition, by measuring the intensity profile of two adjacent cells which are almost unresolved (marked in circle in Fig 3.) the local improvement of the

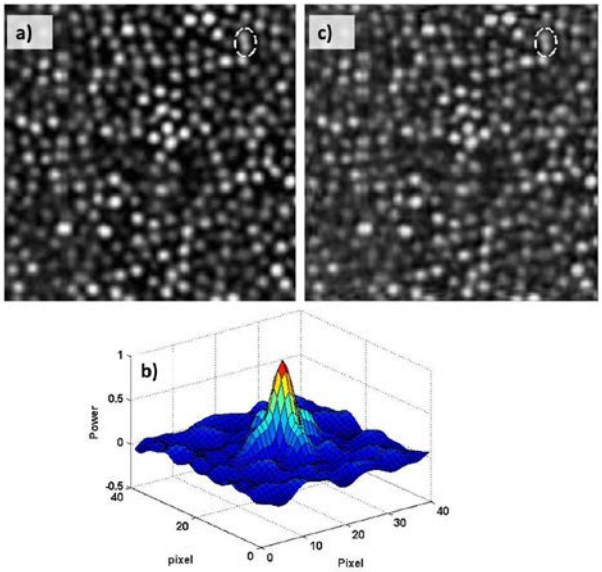

Fig. 3 (Color online): PSF estimation from an adaptive optics retinal image (a), courtesy of Dr. Laurent Vabre, Imagine Eye Ltd., France (450×450 pixels, ~0.1×0.1?? mm). The PSF (b) was used as a Wiener filter kernel to get a reconstructed image with marginal improvement (c). The dashed circles are the area used for the intensity profile in Fig. 4b.

contrast can also be seen (Fig. 4b).

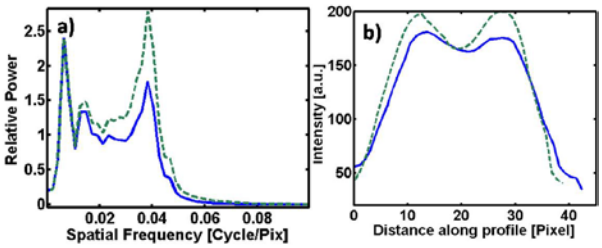

Fig. 4 (Color online): (a) Comparison between the radially averaged power spectrum of an AO image section (solid line) and the reconstructed image (dashed). (b) Intensity profile of two adjacent cells on the verge of the resolution limit.

Good results were also obtained on images that have been taken without AO correction, but after resolution enhancement by weighted shift-and-add [12]. Fig. 5 displays the input image and the reconstructed image, both at low and high magnifications. On the right, the estimated PSF and the power spectra comparison are shown. In this case the improvement is much greater and cells which were originally blurred are now much more visible. In the magnified reconstructed image we see cells of two sizes, probably cones and rods, although the smaller cells are almost irresolvable. We marked with an arrow a structure of a big cell surrounded by these smaller cells. This structure is compatible with the rods-cones photoreceptors arrangement in the corresponding retinal layer [17,18]. In addition, the improvement in the reconstructed image is also backed up by higher spectral power.

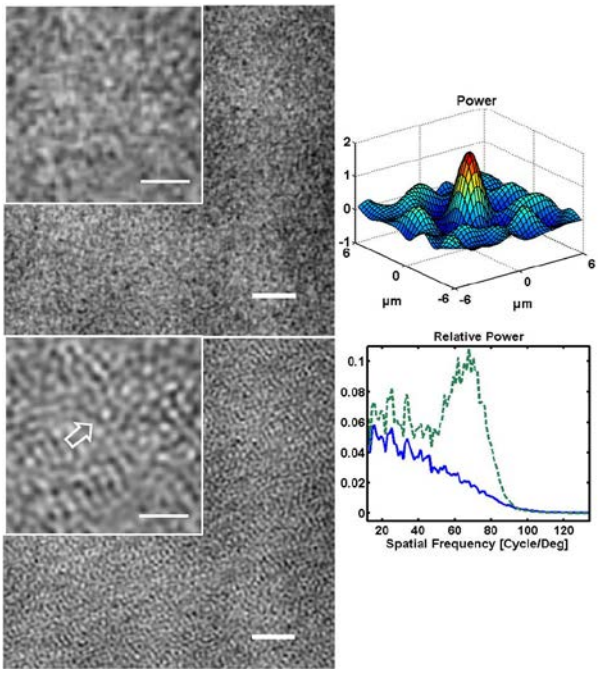

Fig. 5 (Color online): PSF estimation process and Wiener image reconstruction. Top row: image and estimated PSF. Bottom row: reconstructed image and power spectra of the image (solid) and the reconstruction (dashed). Scale bars, 30 μm and 15μm for the large and small fields of view

In conclusion, a method to estimate the point spread function of a retinal image was presented. By using the spatial and intensity distribution of cells in retinal layer, a model of the image can be constructed and the PSF can be estimated. In the absence of a measured PSF in these areas, the estimated PSF can be used to enhance the visibility of cells. Since this method is based on identification of cells, the PSF estimation is expected to be better when using more resolved images as an input.

Parts of this work were supported by the Israel Science Foundation.

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