Portable ultra-widefield fundus camera for multispectral imaging of the retina and choroid

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Abstract: Multispectral imaging (MSI) of the retina and choroid has increasing interest for better diagnosis and treatment evaluation of eye diseases. However, currently available MSI systems have a limited field of view (FOV) to evaluate the peripheral retina. This study is to validate trans-pars-planar illumination for a contact-mode ultra-widefield MSI system. By freeing the available pupil for collecting imaging light only, the trans-pars-planar illumination enables a portable, non-mydriatic fundus camera, with 200° FOV in a single fundus image. The trans-pars-planar illumination, delivering illumination light from one side of the eye, naturally enables oblique illumination ophthalmoscopy to enhance the contrast of fundus imaging. A broadband (104 nm) 565 nm light-emitting diode (LED) is used for validating color fundus imaging first. Four narrowband (17-60 nm) 530 nm, 625 nm, 780 nm, and 970 nm LEDs are tested for MSI. With 530 nm illumination, the fundus image reveals retinal vasculature predominantly. 625 nm and 780 nm illuminations enhance the visibility of choroidal vasculature. With further increased wavelength of 970 nm, the fundus image is predominated by large veins in the choroid, with multiple vortex ampullas observed simultaneously in a single fundus image.

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1. Introduction

Fundus photography is essential for eye disease detection and treatment management. In conventional fundus cameras, a white light source is used to provide flood illumination imaging of the fundus. Using narrow-band interference filters, monochromatic light illumination has been demonstrated to increase image contrast in fundus photography [1], enabling better localization of pathological lesions in the chorioretinal layers [2]. The short wavelength, such as blue or green, light is predominantly sensitive to retinal layers; while long wavelength, such as red or near infrared (NIR), light provides enhanced penetration through the retina to reveal choroidal vasculature. Recently, light emitting diodes (LEDs) has been used to provide monochromatic illumination with discrete wavelength bands for multispectral imaging (MSI) of the fundus [3,4]. By switching the visible and NIR LEDs, the retina and choroid can be selectively observed to advance eye disease management [5]. However, currently available MSI systems have limited field of view (FOV), typically 45° as in traditional fundus camera. Morphological abnormalities such as micro aneurysms in diabetic retinopathy (DR) and tumors in retinoblastoma can occur anywhere, i.e., both central and peripheral regions of the ocular fundus. Therefore, widefield fundus examination is desirable for screening, diagnosis, and treatment evaluation of DR [6,7], retinopathy of prematurity (ROP) [8,9], sickle cell retinopathy (SCR) [10,11], retinal vein occlusion [12], intraocular tumors [13] and other eye diseases that can produce morphological abnormalities at both central and peripheral regions of the fundus.
The limited FOV of traditional fundus cameras is mostly due to trans-pupillary illumination. While the edge part of the pupil is used for delivering illumination light into the ocular fundus, the central part of the pupil is used for collecting light for imaging [14]. Based on the Gullstrand-Principle [15], the illumination and imaging pathways should be separated from each other at both the cornea and the first surface of crystalline lens. Otherwise, the illumination light may cause severe reflection artifacts from the cornea and crystalline lens, which is multiple orders of magnitude higher than the useful signal from the ocular fundus. Therefore, a traditional fundus camera has a FOV typically limited at 30° or 45° external-angle [16]. Sophisticated optical design is required to balance the illumination and imaging pathways, thus increase the instrument complexity and cost of the fundus camera. Examination of the peripheral fundus by digital imaging typically requires pharmacologically pupillary dilation and multiple field imaging to expand the effective FOV. By reducing the region of the pupil occupied for trans-pupillary illumination, miniaturized indirect ophthalmoscopy illumination has been explored for widefield fundus photography [17]. With a NIR light guidance for imaging localization and focusing adjustment, non-mydriatic imaging has been demonstrated with a FOV up to 67° external-angle, which corresponds to a 101° eye-angle [18]. Trans-cranial illumination has been also recently demonstrated for chorioretinal imaging [19]. In principle, trans-cranial illumination has the potential to expand the FOV, because the available pupil is only needed for collecting imaging light. However, it can be difficult to deliver visible light efficiently through the cranium due to heavy light scattering and diffusion. Therefore, its application might be restricted at NIR imaging, with limited contrast of retinal layers which are known to be sensitive at visible light in traditional fundus photography. Scanning laser ophthalmoscope (SLO), such as Optos (Optos, Dunfermline, UK) has been established for ultra-widefield fundus photography with a FOV up to 200° eye-angle [20]. However, multiple laser light sources and a complicated scanning system must be involved, excluding the feasibility of a portable device to foster recently emerging telemedicine ophthalmology.

Trans-scleral illumination has been proposed as one alternative illumination method to achieve ultra-widefield fundus photography, without the requirement of pharmacologically pupillary dilation [21–23]. Trans-scleral illumination has been also recently demonstrated for optical phase imaging of the human retina [24]. Panoret-1000 employed trans-scleral illumination to cover the retina from the optic disc to the ora serrata in a single fundus image [25]. However, clinical deployment of Panoret-1000 was not successful, probably due to its bulky design and difficulty to optimize the illumination efficiency, and it is no longer commercially available. In fact, the optical property of the sclera can vary over different regions. The pars plana is a smooth posterior part of the ciliary body, and lacks muscle, blood vessels and pigmentation. Therefore, the pars plana is more transparent than other scleral areas, making it an optimal location for delivering light into the eye [25–27]. Recently, we have demonstrated a portable pediatric fundus camera with trans-pars-planar illumination, enabling a 200° FOV for comprehensive examination of the retina [28]. For this study, we aim to test the feasibility of using trans-pars-planar illumination for ultra-widefield MSI. Four light wavelengths, including 530 nm, 625 nm, 780 nm, and 970 nm were tested for selective MSI observation of retinal and choroidal layers.

2. Materials and methods

2.1. Experimental setup

Figure 1 illustrates the trans-pars-planar illumination based, portable, non-mydriatic fundus camera for MSI. All off-the-shelf optical components were selected to construct the handheld prototype for a proof-of-concept validation (Fig. 1(A)). The trans-pars-planar illuminator in Fig. 1(A1) consists of an optical fiber, with 1.0 mm diameter and 0.50 numerical aperture (NA) (M59L01, Thorlabs Inc., Newton, NJ). The first component of the imager is a contact indirect ophthalmoscopy lens (CIOL) (HRX Vit, Volk Optical Inc., Mentor, OH). The CIOL consists of
two elements, a meniscus lens and a condensing lens. The CIOL produces an aerial image Retina’ in front of the relay optics (Fig. 1(A3)). The relay optics consists of three lenses, i.e., a bi-convex lens I with a focal length of 60 mm (LB1596, Thorlabs Inc., Newton, NJ), a plano-concave lens II with a focal length of -50 mm (LC1715, Thorlabs Inc., Newton, NJ), and an aspheric lens III (28D, Volk Optical Inc., Mentor, OH). In coordination with the relay optics and a camera lens CL with a focal length of 12 mm (33-303, Edmund Optics Inc., Barrington, NJ), the aerial image Retina’ is relayed to the camera sensor (CS). For color fundus imaging (Fig. 2), one USB 3.1 color camera (DFK 37AUX264, The Imaging Source Europe GmbH, Bremen, Germany) was used. The color CS has a frame resolution of 2448 × 2048 pixels, with 3.45 µm x 3.45 µm pixel size. For MSI (Figs. 3, Fig. 4 and Fig. 5), a monochrome camera (GS3-U3-41S4M-C, Flir systems Inc, Arlington, Virginia) was used. The monochrome CS has a frame resolution of 2048 × 2048 pixels, with 3.1 µm x 3.1 µm pixel size. The sensor provides quantum efficiencies of 58% at 625 nm, 22% at 780 nm, and 4% at 970 nm, respectively.

Fig. 1. (A) Photographic illustration (A1 and A2) and optical layout (A3) of the trans-pars-planar illumination based ultra-wide field fundus camera. (B) Absorption spectra of major chromophores, i.e., oxygenated/deoxygenated hemoglobin and melanin [29]. CIOL: contact indirect ophthalmoscopy lens, CL: camera lens, CS: camera sensor. (C) Light spectra of the 530nm, 565nm, 635nm, 780nm, and 970nm LEDs.

Zemax simulation (OpticStudio, Zemax LLC, Kirkland, WA) was conducted to select and arrange the lenses I, II, and III in Fig. 1(A3)) to relay the aerial image Retina’ to the camera. With trans-pars-planar illumination, the available pupil of the eye is only used for collecting light to the imaging system. In principle, the pupil aperture only affects the light efficiency, i.e., how much light can be collected into the imaging system. Therefore, we only need to arrange the optical system to relay the aerial image Retina’ in Fig. 1(A3)) to the CS, and thus the FOV of the imaging system will be identical to that of the CIOL itself. By considering the 6.6 mm aperture of the camera lens, the effective pupil size for collecting imaging light is estimated as 1.3 mm. For the pupil estimation, we assume the aerial image Retina’ in Fig. 1(A3) is right before the bi-convex lens I. According to the vendor, the magnification of the selected CIOL (HRX Vit, Volk Optical Inc., Mentor, OH) is 0.43. The magnification of the following relay system is 0.18. Therefore, total system magnification is 0.08. By considering the pixel sizes (color camera: 3.45 µm x 3.45 µm; monochrome camera: 3.1 µm x 3.1 µm), the pixel sampling resolution at the fundus is ~40 µm.

For capturing images presented in the article, the aperture of the camera lens was set to F/2. By adjusting the focusing ring in the camera lens, the refractive error of the imaged eye was compensated (Fig. 1(A2)). Camera control and image acquisition was conducted through IC
Capture software (The Imaging Source Europe GmbH, Bremen, Germany) for color images and FlyCapture2 software (FLIR Systems Inc., Wilsonville, OR) for monochrome images.

A broadband LED with 565 nm central wavelength and 104 nm bandwidth (M565L3, Thorlabs Inc, Newton, NJ) was used for validating color fundus imaging in Fig. 2 first. Four narrowband LEDs with center light wavelengths at 530 nm (M530L4, Thorlabs Inc, Newton, NJ), 625 nm (M625L4, Thorlabs Inc, Newton, NJ), 780 nm (M780L3, Thorlabs Inc, Newton, NJ), and 970 nm (M970L4, Thorlabs Inc, Newton, NJ) were tested for selective MSI observation of retinal and choroidal structures. The bandwidths of the 530 nm, 625 nm, 780 nm, and 970 nm LEDs are 35 nm, 17 nm, 28 nm, and 60 nm, respectively. For fundus imaging, the exposure times were configured to 500 ms for 530 nm LED and 100 ms for all other LEDs.

2.2. Human subject and imaging

This study was approved by the Institutional Review Board of the University of Illinois at Chicago and followed the ethical standards stated in the Declaration of Helsinki. Three healthy subjects, without diagnosed eye conditions, and one patient with retinal detachment were recruited for functional MSI validation of the prototype device. The informed consent was taken from each subject.

The imaging experiment was conducted in a room with normal light condition. The pupil size was measured as 3-4 mm, without pharmacological pupil dilation. During the imaging, topical tetracaine 0.05% was used for anesthesia. Lubricant eye gel (GenTeal, Alcon Laboratories, Fort Worth, TX) was applied between the eye and the CIOL. The trans-pars-planar illuminator was contacted to the sclera (Fig. 1). The orientation of the trans-pars-planar illumination was roughly aligned to normal direction of the scleral surface. Considering the ∼4 mm width of the pars plana located ∼3 to 4 mm posterior to limbus in adults, the illumination probe was placed ∼6 mm away from the limbus. Slightly moving the illuminator back and forth from the limbus, the optimal illumination location, i.e., the pars plana, could be identified based on the image quality [17]. When the visible light was delivered through the pars plana, the subject perceived the light as a no irritating uniform light.

The removable design of the CIOL in Fig. 1(A) enabled easy sterilization of the meniscus lens contacting to the eye. According to the instructions from the vendor, 2% glutaraldehyde solution was used for disinfestation and ethylene oxide sterilization method was performed for sterilization of the CIOL [30].

2.3. Light safety

For light safety, both photochemical and thermal hazards of the retina were quantitatively evaluated. ISO 10940: 2009 document [31] is the reference guide used by FDA for the assessment of photo-biological safety of light intensity limits of ophthalmic devices. ISO 10940: 2009 standards are defined for an average human eye. The safety limits in the ISO are at least 10 times below actual retinal threshold damage. To capture fundus images within ocular safety limits, defined by the ISO 10940: 2009 standard, maximum illumination power at the fiber edge was adjusted as 10 mW for the 530 nm green light, 7 mW for the 565 nm yellow light, 7 mW for the 625 nm red light, 4 mW for the 780 nm NIR light, and 3 mW for the 970 nm NIR light. Previous study has reported that the thickness of the sclera is ∼0.5 mm [32]. The transmission of the sclera for different wavelengths was estimated as 15% for the 530 nm green light, 20% for the 625 nm red light, 35% for the 780 nm NIR light, and 50% for the 970 nm NIR light [33]. According to the ISO standard, a maximum of 10 J/cm² weighted irradiance is allowed on the retina without photochemical hazard concern. The weighted irradiance was calculated using the photochemical hazard weighting function provided in the ISO standard. For the red and NIR light sources, aphakic photochemical hazard weighing function was lower than 0.007 and photochemical hazard was negligible. For the greenlight with maximum 10 mW power, the
Table 1. Light safety estimation

| Light wavelength | 530 nm | 565 nm | 625 nm | 780 nm | 970 nm |
|-----------------|--------|--------|--------|--------|--------|
| Maximum power   | 10 mW  | 7 mW   | 7 mW   | 4 mW   | 3 mW   |
| Scleral transmission | 15%    | 17%    | 20%    | 35%    | 50%    |
| Aphakic photochemical hazard weighting factor | 0.025 | 0.017 | 0.001 | 0      | 0      |
| Weighted power for photochemical hazard | 0.25 mW | 0.117 mW | 0.007 mW | 0      | 0      |
| The maximum allowed exposure time | 1.4 hours | 2.6 hours | >24 hours | >24 hours | >24 hours |
| Thermal hazard weighting factor | 1      | 1      | 1      | 0.69   | 0.29   |
| Weighted power for thermal hazard | 10 mW  | 7 mW   | 7 mW   | 2.76 mW | 0.87 mW |
| Equivalent irradiance for thermal hazard | 76.8 mW/cm² | 60.7 mW/cm² | 71.4 mW/cm² | 71.4 mW/cm² | 76.5 mW/cm² |

weighted power was 0.25 mW based on the 0.025 unit aphakic photochemical hazard weighting function for the 530 nm wavelength [31]. For conservative estimation of the worst case, assuming all light directly reaches to the retina behind the illuminated sclera area, the illuminated retinal area was estimated as 1.95 mm², the maximum allowed exposure times is as follows:

\[
t = \frac{10 \text{ J/cm}^2}{0.25 \text{ mW} \times 15\% / 1.95 \text{mm}^2} \approx 1.4 \text{ hours}
\]  

The maximum weighted power intensity allowed on the sclera without thermal hazard concern is 700 mW/cm². For all light sources, the thermal hazard powers were calculated and summarized in Table 1. The equivalent powers for thermal hazard estimation were 76.8 mW/cm², 60.7 mW/cm², 71.4 mW/cm², 71.4 mW/cm² and 0.87 mW/cm² for 530 nm, 565 nm, 625 nm, 780 nm and 970 nm light sources, respectively, which is 8-12 times lower than the maximum limit. Therefore, there was no thermal hazard concern.

3. Results

3.1. Color fundus imaging with a broadband LED

The 565 nm LED, which has a 104 nm bandwidth (Fig. 1(C)), was firstly used to validate ultra-widefield color fundus photography. As shown in Fig. 1(C), the 565 nm LED covers both green and red light wavelengths.

Figure 2(A) shows a representative image captured from one normal eye without previously diagnosed conditions. As shown in Fig. 2(A1), the color fundus image includes the information of both retinal and choroidal vasculatures. By separating green (Fig. 2(A2)) and red (Fig. 2(A3)) channels in the color fundus image (Fig. 1(A1)), it is confirmed that retinal and choroidal vasculatures are predominantly revealed in the green and red channels, respectively.

Figure 2(B) shows a representative image captured from one eye with retinal detachment. The trans-pars-planar illumination, delivering illumination light from one side of the eye, naturally enables oblique illumination ophthalmoscopy to readily reveal retinal detachments (green arrows) and corresponding shadows (red arrows) in Fig. 2(B).

3.2. MSI with narrowband LEDs

Figure 3 shows representative fundus images acquired with 530 nm green (Fig. 3(A)) and 625 nm red (Fig. 3(B)) LEDs. The green-light illumination confirmed retinal vasculature including both
Fig. 2. Representative images acquired with 565 nm LED illumination. (A) Color fundus image (A1), green-channel (A2) and red-channel (A3) of a normal eye. (B) Color fundus image (B1), green-channel (B2) and red-channel (B3) of one eye with retinal detachment (green arrows).

Fig. 3. Fundus images acquired with 530 nm (A) and 625 nm (B) illumination light. The green arrows in A and B point to representative retinal blood vessels. The blue arrows in B show vortex ampullas.

arteries and veins. As shown in Fig. 1(B), both oxyhemoglobin (HbO$_2$) and deoxyhemoglobin (Hb) have high absorption for the green light. Therefore, both arteries and veins show a dark color in Fig. 3(A). The melanin in retinal pigment epithelium (RPE) also has a high absorption for the green light (Fig. 3(A)), and thus exclude the visibility of the choroidal vasculature behind the RPE.

In contrast, the red light was able to partially penetrate through the RPE to reveal choroidal vasculature, while the visibility of the retinal vasculature (green arrows in Fig. 3(B)) was reduced. The blue arrows in Fig. 3(B) point to vortex ampullas which are known to be around the equator
Fig. 4. Fundus images acquired with 780 nm (A) and 970 nm (B) illumination. (C) Schematic diagram of full fundus of young adult. The red and white arrows in A2 show ora serrata and pars plana, respectively. The blue arrows in B in C show vortex vein ampullas. A 780 nm movie clip, corresponding to A, is provided as a supplemental material (Visualization 1). C is reprinted with permission from Zinn [34].

position, i.e., 90° eye-angle from the center of the retina. Therefore, the FOV can be estimated to be >200° in a single fundus image.

Figure 4 illustrates fundus images acquired with light illuminations of 780 nm (Fig. 4(A)) and 970 nm (Fig. 4(B), respectively. With 780 nm illumination, the visibility of choroidal vasculature was further improved, compared to the red illumination in Fig. 3(B). This observation is consistent to the reduced absorption of the melanin in RPE (Fig. 1(B)), and thus allows a better penetration of the NIR light illumination to detect choroidal vasculature. On the contrary, the retinal vasculature was almost disappearing in the 780 nm image. For imaging the central retina (Fig. 4(A1)), the illuminator could be placed on either temporal or nasal sclera. For reaching the ora serrata, i.e. the far end of the fundus (Fig. 4(A2)), the illuminator was placed on the opposite side of the imaged ora serrata and the imaging probe was slightly tilted from the axis. The red and white arrows in Fig. 4(A2) point to the ora serrata and pars plana, respectively. This observation further confirms that the pars-pars-planar illumination enables ultra-widefield fundus photography, up to the far end of the retina. By increasing the illumination wavelength to 970 nm, only choroidal veins were visible in the fundus image (Fig. 4(B)). As shown in Fig. 1(B), the absorption of the melanin in RPE is relatively low at the long wavelength 970 nm light, compared
Fig. 5. Comparative MSI images of a high pigmented subject, with light illumination at 530 nm (A), 625 nm (B), 780 nm (C), and 970 nm (D). Images A1, B, C, and D show single frame recordings. A2 illustrates an average of four frames, corresponding to the imaged region in A1. The white dots in A1 reflect artifacts due to camera noise at high gain used for imaging the high pigmented subject. These white dots were reduced in the average image A2. (E) Schematic diagram of anatomic landmarks of the normal fundus. Blue arrows in D and E point to ciliary nerves. E is reprinted with permission from Zinn [34].
to the 625 nm red light and 780 nm NIR light. This reflects an enhanced penetration of the illumination light to reach the choroid.

The pigmentation level of human subject was observed to affect retinal imaging. Figure 5 show representative MSI images acquired from a subject with relatively high pigmentation level, compared to the subjects in Fig. 3 and Fig. 4. It was observed that the pigmentation level primarily affects the light efficiency and image quality of visible light, particularly the short wavelength green light (Fig. 5(A)). The effect of pigmentation level difference on NIR imaging was negligible. In Fig. 5(C), the choroidal image quality was comparable to that in Fig. 4(A). Figure 5(D) consistently revealed choroidal veins dominated structures. Multiple ciliary nerves were observed (blue arrows) in Fig. 5(C) and Fig. 5(D), which are reasonably consistent to the schematic illustration of anatomic landmarks in the normal fundus (Fig. 5(E)) [34].

4. Discussion

Eye disease such as DR [3,4], ROP [5,6], SCR [7,8], retinal vein occlusion [9], intraocular tumors [10] are known to affect both central and peripheral regions of the ocular fundus. Therefore, widefield fundus examination is important for better disease detection and treatment management. In this study, we demonstrated the feasibility of using trans-pars-planar illumination for developing a portable fundus camera to achieve ultra-widefield MSI. A 200° FOV was achieved in a single fundus image. While the visible light, particularly green light, illumination predominantly reveals retinal vasculature; the NIR light illumination discloses choroidal vasculature.

It is known that the choroid is one of the sources for blood supply to the eye. Therefore, choroidal imaging can be valuable for clinical management of eye conditions. The choroid, located under the RPE, is known to have abundant melanin particles to absorb most of the visible light. In principle, a relatively longer wavelength, such as a NIR illumination can be used to image the choroid. However, the contrast of the NIR light imaging can be limited due to the weak light scattering signal from the fundus. Oblique illumination microscopy has been well established to enhance image contrast of biological tissues [35], but its practical application for fundus photography has been limited due to the limited angle available for delivering light to the fundus through traditional trans-pupillary illumination. Fortunately, the trans-pars-planar illumination, which is delivered from the sclera at one side of the eye, can naturally provide oblique illumination to enhance image contrast. Therefore, robust imaging of choroidal structures was validated using both 780 nm and 970 nm NIR light.

The 780 nm light illumination revealed both arteries and veins in the choroid (Fig. 4(A)). However, interestingly, the 970 nm light illumination disclosed only vortex vein ampullas with large converging veins (Fig. 4(B) and Fig. 5(D)). Considering color inversion, i.e. from bright in 780 nm image to dark in 970 nm image, of the vortex veins as well as high transparency of ocular tissue at 970 nm [36], we plausibly speculate that bright background in 970 nm image is due to the light reflected from the deep sclera, while significant light attenuation occurs at the large vortex veins which exit the globe through the sclera with high flow rate [37,38]. This unique phenomenon would be useful for arteriovenous differentiation in the choroid. It is known that human eye has 4-8 vortex veins [39]. Vortex vein ampullas are located on the equator of the eye. Abnormal and asymmetric appearance of the vortex veins might indicate unhealthy condition of the choroid. Dilatation of asymmetric vortex vein has been reported in central serous chorioretinopathy [39]. Engraforement of the vortex vein was observed more frequently with polyoidal choroidal vasculopathy [40]. The trans-pars-planar illumination based ultra-widefield is capable of imaging the vascular structure of these all vortex veins in a single fundus image (Fig. 4(B) and Fig. 5(D)), thereby promising a practical solution to foster objective assessment of choroidal conditions due to eye diseases. Ciliary nerves were also observed in some of the NIR images, such as Fig. 5(C) and Fig. 5(D). We speculate that the dark edges of the ciliary
nerve might result from the light absorption of the ciliary arteries accompanied with the nerve (Fig. 5(E)).

The pigmentation level of human subject was observed to affect retinal imaging, particularly the short wavelength green light. The effect of pigmentation level difference on NIR imaging is negligible for choroidal imaging (Fig. 5). One limitation of the current prototype device is the long exposure time for the proof-of-concept demonstration. In order to minimize the light power for safety control in the preliminary validation with continuous LED illumination, 500 ms and 100 ms exposure times were used for short (530 nm) and long (625 nm, 780 nm, and 970 nm) wavelength illuminations, respectively. The long exposure time might contribute to image blur due to inevitable eye movements. In principle, the LEDs can be controlled in a pulse manner which allows increased illumination power to compensate for pigmentation levels and to reduce the exposure time, and thus the effect of eye movements can be minimized. In coordination with dynamic monitoring of the overall image intensity, automated control of the required illumination power for each subject can also be achieved. Another contributing factor for the image blur, particularly in color fundus images, is the possible chromatic aberration. Because the current prototype was assembled using all off-the-shelf optics, the chromatic aberration was not quantitatively evaluated. However, the chromatic aberration might not significantly affect the MSI, because individual images were acquired with narrow band LEDs separately. A custom-optics design with optimized chromatic aberration might enhance the image contrast further. A second limitation of the current prototype device is the separate illuminator and imager, which requires the operator to use two hands simultaneously for illumination optimization and imaging adjustment. We are currently pursuing a design to integrate the illuminator and imager into a single handheld device, and thus to pave the way for clinical deployments of the ultra-widefield MSI. The contact-mode imaging modality can be valuable for pediatric applications such as ROP screening, diagnosis, and treatment assessment. Because newborns cannot make coordination for the imaging adjustment, the contact-mode is necessary to minimize the effect of eye movements. However, the contact-mode might not be favorable for adult applications such as DR assessment. In order to minimize the sterilization complications for adult imaging, one solution is to develop a disposable surface of the contact lens. Another alternative solution is to develop a contact-free mode fundus camera. Contact-free trans-pars-planar illumination has been recently demonstrated for non-mydriatic wide field photography. Further system optimization is required to develop a compact, contact-free fundus camera for ultra-widefield MSI.

5. Conclusion

Trans-pars-planar illumination has been demonstrated for ultra-widefield MSI with 200° FOV in a single image. By delivering illumination light from one side of the eye, the trans-pars-planar illumination naturally enables oblique illumination ophthalmoscopy to enhance the image contrast, such as easy detection of retinal detachments. Visible light, particularly green light, imaging predominantly reveals retinal structure, and NIR light imaging reveals choroidal structures. The 780 nm NIR light illumination consistently revealed both arteries and veins in the choroid, while 970 nm light illumination discloses large veins only.

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