Quantitative Measurements of Autofluorescence with the Scanning Laser Ophthalmoscope

François Delori,1,2 Jonathan P. Greenberg,3 Russell L. Woods,1,2 Jörg Fischer,4 Tobias Duncker,3 Janet Sparrow,3 and R. Theodore Smith1,2

PURPOSE. To evaluate the feasibility and reliability of a standardized approach for quantitative measurements of fundus autofluorescence (AF) in images obtained with a confocal scanning laser ophthalmoscope (cSLO).

METHODS. AF images (30°) were acquired in 34 normal subjects (age range, 20–55 years) with two different cSLOs (488-nm excitation) equipped with an internal fluorescent reference to account for variable laser power and detector sensitivity. The gray levels (GLs) of each image were calibrated to the reference, the zero GL, and the magnification, to give quantified autofluorescence (qAF). Images from subjects and fixed patterns were used to test detector linearity with respect to fluorescence intensity, the stability of qAF with change in detector gain, field uniformity, effect of refractive error, and repeatability.

RESULTS. qAF was independent of detector gain and laser power over clinically relevant ranges, provided that detector gain was adjusted to maintain exposures within the linear detection range (GL < 175). Field uniformity was better than 5% in a central 20°-diameter circle but decreased more peripherally. The theoretical inverse square magnification correction was experimentally verified. Photoreceptor bleaching for at least 20 seconds was performed. Repeatability (95% confidence interval) for same day and different-day retests of qAF was ±6% to ±14%. Agreement (95% confidence interval) between the two instruments was <11%.

CONCLUSIONS. Quantitative AF imaging appears feasible. It may enhance understanding of retinal degeneration, serve as a diagnostic aid and as a sensitive marker of disease progression, and provide a tool to monitor the effects of therapeutic interventions. (Invest Ophthalmol Vis Sci. 2011;52:9379–9390) DOI:10.1167/iovs.11-8319

The autofluorescence (AF) of the fundus principally emanates from RPE lipofuscin.1 Lipofuscin is a byproduct of the visual cycle and is a complex mixture of bisretinoids (including A2E) and their oxidized forms.2 Defects in photoreceptor genes can have a direct impact on RPE lipofuscin levels, such as is the case for ABCA4-related retinal disorders.3–5 Several adverse effects of RPE lipofuscin has been demonstrated in vitro, including generation of free radicals,6,7 lysing of cell membrane,8,9 photoinduced apoptosis10,11 and photo-oxidation-associated complement activation.12 These deleterious effects may play a role in the pathogenesis of age-related macular degeneration (AMD) and some retinal dystrophies.13

Fundus AF has been quantified noninvasively by fluorometry14 in normal subjects,1,15 patients with AMD,16 and those with recessive Stargardt’s disease.17 However, fluorometry did not enjoy broad clinical use, in part because of its restricted availability, but also because of the difficulty in obtaining measurements from discrete areas of pathology in the presence of eye movements.

Fundus AF imaging by confocal scanning laser ophthalmoscopy (cSLO)18–21 or by fundus camera22,23 allows visualization of the spatial distribution of fundus AF. In some retinal disorders, the distribution of fundus AF deviates from normal such that AF patterns can assist in diagnosis. Generally, differences in AF intensity within images have thus far not been comparable between patients, or even between successive images of the same patient. Notable exceptions are short-duration clinical24,25 and basic26–27 studies of intensity of AF images obtained using the same cSLO and a comparative study of different cSLOs using an external fluorescence standard.28

Given the widespread use of fundus AF in clinical settings, there is a need for a standardized approach that can reliably determine AF levels at specific retinal locations so as to interpret fundus AF findings in relation to given pathologic conditions.29 Some of the challenges associated with the quantification of AF from cSLO images have been discussed,30 but no approach has been identified for clinical use.

AF quantification would aid in addressing questions such as whether a fundus area has normal or abnormal AF levels and whether AF levels correlate with disease progression. Thus, for Stargardt’s disease, AF levels could provide valuable genotype-phenotype correlations, establish whether increased AF is indicative of an ABCA4 carrier state, and serve as potential metrics for response to therapy. For retinitis pigmentosa, AF levels in rings with high AF could be studied in relation to photoreceptor changes detected on spectral domain optical coherence tomography scans. For AMD one could determine whether higher AF in normal subjects is a risk factor for AMD. As treatments for these disorders become available, this approach could be used to monitor the efficacy of therapeutic interventions such as gene therapy or drugs designed to decrease RPE lipofuscin formation.

We have developed and tested a method to perform standardized quantitative measurements of fundus AF. This technique is applicable to SLOs and, in theory, to fundus cameras. The basic principle of the method is that when the AF from the fundus is normalized to the fluorescence of a standard

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Supported by National Eye Institute Grants RO1 EY015520 and R24 EY019861; the Foundation Fighting Blindness; New York Community Trust; and the Roger H. Johnson Fund, University of Washington, Seattle, WA.

Submitted for publication July 29, 2011; revised October 9, 2011; accepted October 10, 2011.

Disclosure: F. Delori, None; J.P. Greenberg, None; R.L. Woods, None; J. Fischer, Heidelberg Engineering (E); T. Duncker, None; J. Sparrow, None; R.T. Smith, None.

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(mounted within the imaging device), the effects of variation in laser power and detector gain can be compensated. Thus, fundus AF can then be compared longitudinally, between eyes and between images obtained with different devices. In this article, we demonstrate how this approach can be implemented in two cSLOs. The underlying optical principles are presented in the Supplementary Material (http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental). We have tested the method by comparing measurements between eyes, between sessions, and between instruments and by systematically varying common operator settings.

**METHODS**

**Subjects**

Thirty-four subjects (20 women and 14 men) with normal retinal status participated in the study. Twenty-four were white, and 10 were Asian, black, or of Hispanic ethnicity. Ages ranged from 20 to 55 years, and refractive errors from −4.4 to +3.7 D. All subjects had relatively good fixation and clear media, except for some floaters.

The tenets of the Declaration of Helsinki were followed. Institutional Review Board approval was granted, and informed consent was obtained for all subjects. The pupil of the test eye was dilated to at least 7 mm in diameter using 1% tropicamide and 2.5% phenylephrine. The pupil of the test eye was dilated to at least 7 mm in diameter using 1% tropicamide and 2.5% phenylephrine. The retinal light exposures (recommended maximum power: 280 μW; 30° × 30° field; 488 nm) are below the limits recommended by the ANSI standards for durations up to 8 hours.31,52

**cSLOs and Internal Reference**

An HRA2 and a 53500 Spectralis HRA-OCT (both Heidelberg Engineering, Heidelberg, Germany) were used in this study (high-speed mode; 30° field; 768 × 768 pixels; 8.9 frames/s). With the exception of the internal reference, both devices were standard cSLOs with the excitation light (488 nm) generated by a laser (Sapphire; Coherent GmbH, Lübeck, Germany) and coupled via single-mode fiber into the camera head. The barrier filter in both devices transmitted light from 500 to 680 nm. The optical systems are similar as far as AF imaging is concerned (Appendix A, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental). Although the detectors were the same, the “advanced sensitivity mode” was used in the Spectralis, whereas this mode was not available for the HRA2. Consequently, the sensitivity settings of each device were different and not directly comparable. The laser power of the two cSLOs was monitored at intervals of 3 to 9 months and was always between 220 and 260 μW. It decreased by 0.4% to 1.4%/mo, probably varying with the use of the devices.

As the internal reference was mounted at the “intermediate” retinal plane of both cSLOs, it was always in focus with the fundus image (Fig. 1). Spectral and other characteristics of the fluorescent material are provided in Figure 2 and Table 1. An internal reference can also be readily inserted in a fundus camera as indicated in Appendix E, Supplementary Material (http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental).

**Image Acquisition**

A single operator (JG) acquired AF images with both the HRA2 and the Spectralis. Room lights were dimmed (monitor glow only) to reduce possible effects on the test and to minimize distraction of the subject. With the subject’s head positioned in the chin-head-rest, the laser was switched to the blue (488 nm) excitation mode, and the image was refocused until the whole field reached its maximum intensity. The focus was ~1 D more myopic at 488 than at 830 nm, consistent with chromatic aberrations between the two wavelengths.33 The sensitivity, S (acquisition screen: Sens.), was adjusted to avoid nonlinear effects (colored pixels appear in the image if GL is >252), followed by a bleaching period of 20 to 30 seconds to reduce phototopic absorption to <5%,34–37 Final alignment of the camera was critical to centrally align the camera to avoid obstruction by the iris. The camera was aligned in all three dimensions such that optimal image uniformity was obtained (minimizing the extent of the lowest signals at the sides and corners of the image).

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was made during that period. The patient was asked to blink a few times to provide a uniform tear film on the cornea. Eyelid interference, causing localized decreased signal, was avoided. Nine successive frames were then acquired. Frames were examined, and those demonstrating either localized (eyelid interference) or generalized (iris obstruction) decreased signal were eliminated (necessary in ~10% of images). The frames were then aligned and averaged with the system software and saved in the nonnormalized mode (no histogram stretching), whereas the latter was not.

All AF images were analyzed with a dedicated image analysis program (IGOR, Lake Oswego, OR). The software transforms the entire image into a qAF-map of the fundus. Color-coded maps can also be generated, and they may be clinically useful. Average qAF can be computed in the preset regions (Fig. 1) or in manually selected areas of the image. In this study, we mainly analyzed the fovea, where AF is highly attenuated by the ocular media for wavelengths below 535 nm. The excitation spectrum for older individuals is extrapolated. The emission spectra between 500 and 555 nm were extrapolated. The excitation spectrum from 480 to 490 nm, % 470–500-nm blue light, % 398, since both the excitation and emission are affected.

Details of the derivation of equation 1 are given in Appendix B and experimental verification of square law is described in Appendix C (Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/ iovs.11-8319/-/DCSupplemental). The parameters are:

RCF, reference calibration factor (see below)

gAF, mean GL in the fundus area of interest
GL0, zero signal level expressed in GL, provided by the system software (‘offset’ in ‘image info’ panel). This signal is measured in a horizontal scan outside the image, with the laser turned off and the detector on (images can be affected by room light).

SF, scaling factor in retinal micrometers per pixel (provided in image info). SF depends on the focus setting (refraction) and the corneal curvature of the subject (Appendix D, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/ iovs.11-8319/-/DCSupplemental). SF can also be calculated from the Gullstrand-Emsley model eye.33 SFem,7.7 is the SF for an emmetropic eye with average dimensions.

Thus, gAF represents the fundus autofluorescence relative to that which would be measured through the media of a 20-year-old emmetropic eye with average ocular dimensions.

The RCF was obtained, for each internal reference in each device, by in situ calibration with a (tentative) master reference, of the same fluorophore mounted 20 cm from the SLO’s detection pupil (Table 1). The RCF for the HRA2 was 0.89 ± 0.04 that of the Spectrals, because the former was equipped with a slanted quarter-wave plate in front of the condenser lens (to reduce a reflection artifact), whereas the latter was not.

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FIGURE 2. Excitation and emission spectra (thick lines, measured by spectrophotometry; MPF-44A; Perkin Elmer, Boston, MA) of the fluorescent material used in the study together with the spectral ranges of the excitation laser and of the barrier filter. Other characteristics of the internal reference are given in Table 1. For comparison, excitation and emission spectra of the fundus are shown for the 20- to 30-year age group (dashed lines, Y) and for the 60- to 70-year age group (solid lines, O). The spectra were measured by fluorometry14 with no correction for ocular media losses. The emission spectra between 500 and 555 nm were extrapolated. The excitation spectrum for older individuals is highly attenuated by the ocular media for wavelengths below 500 nm.

**Image Analysis**

To calculate the quantified (q)AF from gray level (GL) measurements in an image, we used:

\[
\text{qAF} = \text{RCF} \times \frac{\text{GL}_k - \text{GL}_0}{\text{GL}_k - \text{GL}_0} \times \left(\frac{\text{SF}_{\text{em},7.7}}{\text{SF}_{\text{em},7.7}}\right) \times \frac{T_{\lambda_{20}}}{T_{\lambda_{20}}} \times \frac{T_{\lambda_{20}}}{T_{\lambda_{20}}} \tag{1}
\]

Details of the derivation of equation 1 are given in Appendix B and experimental verification of square law is described in Appendix C (Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/ iovs.11-8319/-/DCSupplemental). The parameters are:

RCF, reference calibration factor (see below)

GLk, mean GL in a defined area of interest
GL0, mean GL in a defined area of interest (Fig. 1)

GLr, zero signal level expressed in GL, provided by the system software (‘offset’ in ‘image info’ panel). This signal is measured in a horizontal scan outside the image, with the laser turned off and the detector on (images can be affected by room light).

SF, scaling factor in retinal micrometers per pixel (provided in image info). SF depends on the focus setting (refraction) and the corneal curvature of the subject (Appendix D, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/ iovs.11-8319/-/DCSupplemental). SF can also be calculated from the Gullstrand-Emsley model eye.33 SFem,7.7 is the SF for an emmetropic eye with average dimensions.

Tλ, Tλ, Transmission of the ocular media at the excitation (λ) and emission (λ) wavelengths, respectively. The transmissions Tλ,20 and Tλ,20 are the average transmissions for the media of 20-year-old subjects.

Thus, qAF represents the fundus autofluorescence relative to that which would be measured through the media of a 20-year-old emmetropic eye with average ocular dimensions.

The RCF was obtained, for each internal reference in each device, by in situ calibration with a (tentative) master reference, of the same fluorophore mounted 20 cm from the SLO’s detection pupil (Table 1). The RCF for the HRA2 was 0.89 ± 0.04 that of the Spectrals, because the former was equipped with a slanted quarter-wave plate in front of the condenser lens (to reduce a reflection artifact), whereas the latter was not.

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**TABLE 1. Characteristics of the Internal Reference**

<table>
<thead>
<tr>
<th>Red Slide* + NDF† (1.3 DU)</th>
<th>Dimensions, mm</th>
<th>Peak emission, nm</th>
<th>Half-height points, nm</th>
<th>Change in excitation spectrum from 480 to 490 nm, %</th>
<th>Fluorescence decay to 5% of peak, ns</th>
<th>Change in fluorescence after a 300-h exposure with 1.3 mW/cm² of 470–500-nm blue light, %</th>
<th>RCF</th>
<th>HRA2</th>
<th>Spectralis</th>
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<tr>
<td></td>
<td>7 × 5 × 1.3</td>
<td>590</td>
<td>570–615</td>
<td>+12</td>
<td>29‡</td>
<td>+0.8 ± 0.05§</td>
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* Developed and tested for stability by Ping Chin Cheng (SUNY, Buffalo, NY). The fluorophore is Texas Red dye with other proprietary compounds embedded in a plastic matrix (Microscopy/Microscopy Education, McKinney, TX).
† Neutral-density filter (Wratten; Kodak, Rochester, NY) optically cemented to the slide. This method provides an attenuation of 10^-3.3 = 398, since both the excitation and emission are affected.
‡ The toy decay must be short enough to ensure that sufficient fluorescence is detected before the horizontal scanner moves to neighboring areas (pixel clock: 100 ns in high-speed mode).
¶ The test irradiance was 14× higher than the irradiance in the intermediate plane and 2200× higher (14 × 10^3) than the irradiance on the fluorescent material.
§ An early version of the internal reference, used here in a few tests, used a lower NDF of 1.1 DU; the RCF was 515.
lowest, and the four segments, with emphasis on the temporal segment, where fundus AF is generally highest.\textsuperscript{15}

We noticed that the focus readings necessary for best image quality in the same patients were different for the Spectralis and the HRA2 (Spectralis being 1.26 \pm 0.26 D more myopic). The focus readings for the HRA2 agreed with the refractive error determined by autorefraction (Nidek, Hiroishi, Japan). Therefore, we corrected the Spectralis’ focus by adding 1.2 D. This decreased SF, thereby reducing qAF by 3\% to 4\% (Appendix D, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental). We also found that the zero GL\(_0\), given by the software (offset) was 0.6 to 0.8 GLs (\(P < 0.001\)) higher than that measured in a completely dark room. This small error will only affect qAF when a very low AF is measured.

In the present study, we did not measure individual corneal curvatures (needed for computation of SF; Appendix D, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental). We also did not, at this point, individually correct qAF for losses of light (needed for computation of SF; Appendix D, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental) but used the default value of 7.7 mm. Also, we did not, at this point, individually correct qAF for losses of light in the ocular media.

Statistical Analyses
Since sample sizes were generally small, we used nonparametric statistical tests (performed with StatPlus; AnalystSoft, Vancouver, BC, Canada, and SPSS 19.0 for Mac; SPSS IBM, Chicago, IL). To measure test-retest repeatability and between-instrument agreement between two measures qAF\(_1\) and qAF\(_2\) (\(\Delta qAF = qAF_2 - qAF_1\)), we used the widely accepted method of Bland and Altman.\textsuperscript{38} Since stepwise regression showed a weak positive association between the absolute values of \(\Delta qAF\) and qAF (consistent with a component of the measurement noise being related to signal strength), we transformed \(\Delta qAF\) into the relative values \(\Delta qAF/qAF\). The (coefficient of) repeatability, expressed in percent, is then:

\[
\text{Repeatability} = \frac{\sigma_{\Delta qAF}}{qAF} \times 100 \quad (2)
\]

This equation provides the 95\% confidence intervals for testing under the same conditions (e.g., same day, different days). In this article, we use the term repeatability for comparisons within instrument. Agreement is also defined by equation 2 and was used when the two instruments were compared. Repeatability is the coefficient of variation in percent, multiplied by 1.96. If the variability of measurements increases (increased measurement noise), so will the repeatability (interval).

**RESULTS**

To illustrate the main features of the method and its ability to account for changes in detector gain, G, we analyzed images from an eye over a range of Ss (Fig. 3). S denotes the setting on the control panel, whereas gain G denotes the actual relative gain of the detector. The GLs of the fundus and the reference increase with S, corresponding to the change in G. The quantified autofluorescence, qAF (equation 1), was independent of S despite an increase in G by a factor 2.1 (S = 89–93). Small variations in qAF were probably due to instrument noise, subject fixation, and other sources of measurement noise.

**Linearity**

For equation 1 to be applicable, the GL output of the detection system must be linear with respect to the intensity of the actual fluorescent signal. To test the linearity of the HRA2 and the Spectralis, we imaged a pattern located 20 cm from the detection pupil of the camera (focus, 5 D). It consisted of a raster with 36 calibrated fluorescence intensities (Fig. 4, inset). Images of the pattern were recorded at different S for the two cSLOs, and the results were adjusted to simulate an equal laser power for both devices. The two devices were linear for exposures that produced GL < 175 (Fig. 4). At higher exposures, the fluorescence was increasingly underestimated. Although none of the pixels of the mean image reached a GL of 255, some pixels in the nine individual frames did, causing the mean to deviate from linearity.

**Bleaching before qAF Imaging**

To verify whether the 20-second bleaching period before AF imaging was sufficient to adequately reduce the absorption of photopigment, we recorded, in three subjects, the decrease in attenuation during bleaching (Fig. 5). Although there was a surprising lack of variability among these subjects, the bleaching duration of 20 seconds appeared adequate for rods. Longer bleaching exposure may be needed for older subjects. For foveal cones, a slight increase in attenuation was observed after 30 seconds of bleaching (Fig. 5).

**Quantitative AF at Different Sensitivities**

To assess the efficacy of the internal reference approach at accounting for different laser powers and sensitivities (S), we
recorded (as we did in Fig. 3) single images obtained at different Ss from the fundi of eight subjects (age range, 21–35 years) and from three stationary fluorescent targets A, B, and C. The target tests were repeated three times (Fig. 6). The qAFs (normalized to one S) at the two retinal sites correlated significantly (Spearman $\rho = 0.40$; $P = 0.01$), indicating that the qAF changes are due in part to factors that affected the entire image (e.g., obstruction by the iris, tear film alterations). This similarity is apparent in Figure 6 for some subjects (e.g., KD and TY), in the variation of qAF with S for the temporal and foveal sites.

All data were analyzed for the influence of S in the same range of detector gains (HRA2, $S = 72$–84; Spectralis, $S = 72$–84). In the eight subjects, the temporal measure had a higher qAF than did the foveal (repeated-measures ANOVA, $F_1,7 = 39.4$; $P < 0.001$), and there was no effect of S on qAF ($F_{1,28} = 0.78$; $P = 0.55$) at both retinal sites (i.e., no interaction, $F_{4,28} = 0.95$; $P = 0.45$). For the data on the three stationary fluorescent targets, there were large differences in qAFs between the targets (Greenhouse-Geisser corrected $P < 0.001$), but no effect of sensitivity ($P > 0.14$).

It is apparent from Figure 6 (logarithmic plot) that the measurement noise increased with decreasing qAF for both subject and pattern data (see also later discussion). Furthermore, the noise was higher for the subject data because of errors related mainly to changes in alignment and head/eye movements.

Similar tests were performed on the stationary fluorescent target by adjusting the laser power from 210 to 250 $\mu$W. No differences in qAF were detected (four sites, two sensitivities each; Wilcoxon $Z = 0.4$, $P = 0.7$).

**Field Uniformity**

We assessed the field uniformity of the fundus excitation and AF detection by acquiring five to eight fundus images in each of five subjects using different fixations provided by the HRA2 as well as some intermediate fixation positions (Figs. 7A, 7B). Uniformity profiles were obtained by plotting the qAF relative to that at the image center as a function of eccentricity (Fig. 7C). Although some small asymmetries (e.g., top-bottom or left-right) were detected in individual subjects, we assumed that the profiles were circularly symmetric.

The uniformity profiles all show a decreasing signal with increasing eccentricity: at an eccentricity of 10°, the qAF was $\approx 95\%$ of the central value, and it decreased further to $\approx 85\%$ at the edge of the field (eccentricity, 15°). The corners of the image were always the darkest (60%–80%). Thus, the area of highest uniformity was a 20°-diameter circle centered in the field, where average signals did not drop below 95%. Nonuniformities are caused by the optics between the intermediate plane and the retina (camera lens and ocular optics), because uniformity was observed in the intermediate plane (Fig. 7C, bottom profile).

**FIGURE 4.** Zero-corrected gray levels (GLs) versus the fluorescence of a calibrated pattern AFpattern at different sensitivities, S, (as indicated) for the HRA2 and the Spectralis. Inset: an image of the calibrated pattern. Dashed lines: point at which nonlinearity reached 5% at different sensitivities. Pixel-sized flashes of colored light appear slightly below the 5% level. For high exposures, the GLs saturate at a level of 255 – GL0. The range of AFpattern was chosen to correspond roughly with measurements previously obtained by fluorometry: normal subjects (AFpattern = 200–400), patients with AMD (AFpattern = 300–500), Stargardt’s disease (AFpattern = 500–1200), and Best disease (AFpattern ≈ 1600).1,13-15 Dark arrow: the equivalent exposure for the internal reference currently used in the study. For example and for the HRA2, if we set a lower limit for GL = GL0 to 25 (mainly for contrast), then the reference should be adequate to cover a large range of fundus AF levels ranging from an equivalent of 1600 (using S = 89, GL = GL0 ≈ 110, and $GL_R = GL_0 = 25$ at the limit) to an equivalent of 100 (using S = 96, $GL_R = GL_0 = 25$ at the limit, and $GL_0 = GL_0 = 165$).
Repeatability

Repeated analysis of the same image yielded repeatability of ±1% to ±1.5%, reflecting positioning of the sampling areas.

We tested repeatability of qAF in both eyes of 12 subjects (ages, 25–50 years) using both the HRA2 (S = 91–94) and the Spectralis (S = 72–90). In addition to the fovea and the temporal segment, we also considered the larger sampling area of all four segments (Fig. 1). We tested repeatability for qAFs obtained from images (1) within a session (sitting), (2) between sessions on the same day, and (3) between sessions on different days (Fig. 8). No statistically significant differences were found between those repeated measures at any site, except as mentioned below.

Within Session. Two successive images were obtained within a session (≈2–6 seconds apart) using the same positioning in the chin/head rest, alignment of the camera, and focus. Different sensitivities were used, but they were the same for each image pair. Repeatability was ±6.9% for the fovea and improved to ±2.7% for the four-segments (Fig. 8). As repeatability is a 95% confidence interval, a subsequent foveal qAF measurement taken within a session will only differ from the first measurement by more than 6.9% on 5% of occasions.

Between-Sessions, Same-Day. For qAF data obtained in two sessions on the same day (<5 minutes apart), the operator selected the same sensitivity in 58% of the cases. Having the subject move away from the instrument made the qAF measurements more variable than those within session, as repeatability was ±6% to ±11% at the three sites (Fig. 8). For the fovea, the qAF for the second session was slightly smaller than that in the first (by ≈2%; Wilcoxon Z31 = 2.0, P = 0.05). Since this effect on qAF appears to be exacerbated by low fundus AF, we speculate that the determination of the GL0 level may be responsible. This possibility is being investigated further.

Between Sessions Different Day. For qAF data obtained on different days (28–64 days apart), the same S was selected in 33% of the cases. Repeatability was ±7% to ±14% (Fig. 8), slightly worse than the same-day measurements, suggesting that much of the difference in repeatability between within-session, same-day and between-sessions was due to subject alignment and related issues.

The differences (ΔqAFs) at the foveal and temporal segments correlated with each other for both the within-session (Spearman, rS = +.041; P = 0.0007) and the same-day, between-session comparisons (rS = +.061; P = 0.0002). Furthermore, significant correlations were found between the ΔqAFs for the other sites (nasal, superior, and inferior segments; data not shown). In fact, 32% (within-session) and 47% (same-day, between-sessions) of all comparisons showed either an increase or a decrease in qAF at all five sites, although the differences were not uniformly distributed.

Internal Reference. The GLr – GL0 normalization term was used in equation 1 to account for variations in laser power, gain, and other sources of instrument noise, but can also be used to characterize measurement noise from the instrument alone under the above conditions (as shown in Fig. 8). Repeatability of GLr – GL0 is derived from images obtained at the same S. Within-session repeatability for GLr – GL0 was ±1.2% and ±2.0% for the HRA2 and the Spectralis, respectively (n = 147, corresponding to optical densities of 0.16 to 0.19 DU for the rods (500 nm). The attenuation was reduced to 1.05 (5% absorption) after a bleaching duration of 12 to 17 seconds and to 1.02 (2% absorption) after 17 to 24 seconds. Marginal increases in AF were observed at the fovea (ages, 25–50 years) using both the HRA2 (S = 91–94) and the Spectralis (S = 72–90). Images were recorded after 11, 21, 31, and 51 seconds, to document the effect of bleaching. (B) Image recorded 30 seconds after image (A). No clear differential AF was detected between the areas that were and were not initially bleached. (C) Time course of the attenuation of the AF during bleaching for three subjects (age range, 23–47 years). Error bars are SD, calculated from propagation of errors. Measurements were made within the superior (△) and inferior (▽) dark bands, at the boundary of the dark area (eccentricities, 12°–17°), and in the fovea (○). The AF measured in the bleached zone outside the fovea acted as the reference. The data were fitted by exponential functions. At t = 0 the attenuations ranged from 1.37 to 1.47, corresponding to optical densities of 0.16 to 0.19 DU for the rods (500 nm). The attenuation was reduced to 1.05 (5% absorption) after a bleaching duration of 12 to 17 seconds and to 1.02 (2% absorption) after 17 to 24 seconds. Marginal increases in AF were observed at the fovea for RL and TM (○), perhaps related to regeneration of the photopigment.
66 and \( n = 67 \). Between-session, same-day repeatability was 3.1% and 4.2% for the HRA2 and the Spectralis, respectively (\( n = 21 \) and 4; Spectralis has many more S settings than HRA2, and thus few pairs with the same S). The GLR of the internal reference was lower for the second session than the first (by 2% \( 1\% \); Wilcoxon \( Z = 3.3, P = 0.0009 \)), and that seems to be related to the decrease in foveal qAF sessions conducted on the same day, as reported above.

Between-session, different-day repeatability (over a 4-month period) was 4.7% and 5.7% for the HRA2 and Spectralis, respectively (\( n = 95 \) and 89), but this was complicated by the variability in between-session durations and gradual reduction in the GLR over time (a systematic bias for each device, Spearman \( R = 0.35; P = 0.0001 \)), presumably due a reduction in laser power. Internal reference repeatability indicates the lower limit on the repeatability that is possible with these two cSLOs, as configured (i.e., if subject-related sources of error could be minimized). As described next, those limits will also vary with the sampling area and qAF.

**Effects of Sampling Area and qAF Level.** Figure 8 shows that repeatability for the fovea was worse than for the temporal segment, which was worse than the four-segments. To evaluate the effects of sampling area and qAF level on measurement noise, we estimated within-session repeatability with smaller sampling areas for subjects and for stationary AF patterns (Table 2). Repeatability was worse as qAF decreased (multiple regression, \( t = 5.2; P < 0.001 \)) and as sampling area decreased.
Effects of Errors in Focus and Alignment

We investigated how focus errors and changes in axial and lateral position of the cSLO with respect to the eye contributed to these measurement errors. Deviation from the focus that gave maximum AF intensity (the end point in AF imaging) by \( \pm 0.6 \) D resulted in a 5% decrease in qAF (Fig. 9A). Thus, it is unlikely that this error in focus played a major role, since the absolute difference in focus between sessions was only 0.2 D (IRQ \( Z_{16} = 0.7, P < 0.01 \)) and no differences were detected for the temporal segment (Wilcoxon \( Z_{16} = 0.0, P > 0.2 \)) or the four-segments (Wilcoxon \( Z_{16} = 0.8, P = 0.4 \)). After correction for the Spectralis focus (the Methods section), the absolute difference in the foci of both devices was 0.17 D (IRQ \( Z_{16} = 0.21 \)). The agreement (between the HRA2 and the Spectralis) was <11% (Fig. 8), but did not show the benefit of larger area sampling observed in the repeatability tests. We suspect that this was due to asymmetrical nonuniformities caused by slight misalignment of one of the devices.

Table 2. Within-Session Repeatability of qAF for Different Sampling Areas

<table>
<thead>
<tr>
<th>Site (Eccentricity)</th>
<th>Image Pairs (n)</th>
<th>Sampling Area* (Pixels(^2))</th>
<th>Mean qAF (qAF Units)</th>
<th>Repeatability† (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Fundus Sites</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mean of four-segments§</td>
<td>115</td>
<td>52,120</td>
<td>240</td>
<td>5.5</td>
</tr>
<tr>
<td>Segments (6.6–8.9°)‡</td>
<td>115</td>
<td>8,030</td>
<td>60</td>
<td>5.1</td>
</tr>
<tr>
<td>Superotemporal (13°)</td>
<td>23</td>
<td>850</td>
<td>90</td>
<td>5.9</td>
</tr>
<tr>
<td>Inferior to fovea (1°)</td>
<td>23</td>
<td>850</td>
<td>90</td>
<td>6.4</td>
</tr>
<tr>
<td>Fovea (2° diam. circle)‡</td>
<td>115</td>
<td>2,540</td>
<td>190</td>
<td>6.3</td>
</tr>
<tr>
<td>Superior (10°)</td>
<td>23</td>
<td>140</td>
<td>80</td>
<td>6.4</td>
</tr>
<tr>
<td>Fovea (1.4° diam. circle)</td>
<td>23</td>
<td>1,270</td>
<td>80</td>
<td>7.2</td>
</tr>
<tr>
<td>Temporal (13°)</td>
<td>23</td>
<td>480</td>
<td>225</td>
<td>7.3</td>
</tr>
<tr>
<td><strong>AF Targets</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Master Reference</td>
<td>15</td>
<td>14,000</td>
<td>460</td>
<td>6.7</td>
</tr>
<tr>
<td>Internal Reference</td>
<td>70</td>
<td>4,140</td>
<td>250–260</td>
<td>1.4</td>
</tr>
<tr>
<td>Target B</td>
<td>15</td>
<td>7,300</td>
<td>220</td>
<td>1.6</td>
</tr>
<tr>
<td>Target C</td>
<td>15</td>
<td>2,050</td>
<td>90</td>
<td>3.1</td>
</tr>
<tr>
<td>Cell B3, Pattern (Fig. 4)</td>
<td>20</td>
<td>130</td>
<td>190</td>
<td>4.4</td>
</tr>
</tbody>
</table>

* One optic disc area in a 768 x 768 image is \( \approx 10,000 \) pixels\(^2\).
† Repeatability for image pairs obtained using the HRA2 and the Spectralis (Heidelberg Engineering, Heidelberg, Germany).
‡ Same sites as those in Figure 8.
§ Representative data from stationary AF pattern (of a total of 15 pairs).

with the Spectralis (S = 72–90) on the same day. qAFs measured with the Spectralis exhibited a tendency to be lower than those measured with the HRA2 at the fovea (by 2% ± 4%: Wilcoxon \( Z_{16} = 1.8, P = 0.08 \)) but no differences were detected for the temporal segment (Wilcoxon \( Z_{16} = 1.0, P = 0.5 \)) or the four-segments (Wilcoxon \( Z_{16} = 0.8, P = 0.4 \)). After correction for the Spectralis focus (the Methods section), the absolute difference in the foci of both devices was 0.17 D (IRQ \( Z_{16} = 0.21 \)). The agreements (between the HRA2 and the Spectralis) were <11% (Fig. 8), but did not show the benefit of larger area sampling observed in the repeatability tests. We suspect that this was due to asymmetrical nonuniformities caused by slight misalignment of one of the devices.

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and in the four-segments (Fig. 10), even without correction for qAF exhibited a significant increase with age both at the fovea and in the four-segments (Fig. 10), even without correction for qAF as a Function of Age

qAF exhibited a significant increase with age both at the fovea and in the four-segments (Fig. 10), even without correction for

ocular media absorption. All subjects were white and had normal retinas \((n = 20, 20 - 50 \text{ years})\). The measurements were made over a 6-month period with the two cSLOs. The correlation for the fovea was less significant than that at the temporal site because foveal AF is variably attenuated by macular pigment absorption. When corrected for ocular media absorption using an algorithm\(^9\) that predicts the optical density of the media at a given age and wavelength (Appendix F, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental), the rate of increase in the four-segment qAF was 8.6 qAF units/y, slightly smaller than an equivalent rate of 12 qAF units/y found by fluorometry in a much larger population.\(^1\)

**DISCUSSION**

Measurement of AF from images obtained using SLOs and fundus cameras is feasible if images of high quality and uniformity are recorded by a skilled operator,\(^4\) if care is taken to adjust exposures within the range of linearity (Fig. 4), and if the important retinal features are aligned within the central 20°-diameter circle with the highest uniformity (Fig. 7). Additional requirements include bleaching the photopigment (Fig. 5) and recording multiple images in one test session to ascertain individual variability. The advantages of the internal reference approach were demonstrated by the relative resistance of qAF to changes in \(S\) and laser power (Fig. 6), and by the fact that measurements taken at protracted time intervals and on different devices were reproduced with reasonable accuracy (Fig. 8). Therefore, qAF measurements using the established protocols can be performed in clinical settings and can be exchanged between investigators. Even though these advantages would be forfeited if an internal reference were not used, that does not preclude conducting valuable quantitative studies without an internal reference, as long as they are performed using the same instrument over a short time period. The benefit of using an internal reference in such studies will depend on the instrument-related noise (repeatability of \(\pm 3\%\) to \(\pm 6\%\) between sessions for our instruments), systematic errors (as noted above), and the effect magnitude and time interval being studied.

Limitations of our study include the difficulty of identifying technical issues and adapting protocols; recording of all images by a single operator to avoid interoperator variability during these initial studies; and the testing of only young subjects (\(\leq 55\) years) with reasonably good fixation. Furthermore, the presence of the internal reference in the image compromised the software alignment of the frames, particularly when fixation was poor (the stationary reference and the moving fundus image compete for alignment). This resulted in the blurring of the reference image and/or suboptimal alignment of the fundus features. Heidelberg Engineering has now developed modified software that aligns the fundus image but not the area occupied by the reference. This modification could also be implemented in the software of the ART (automatic real-time averaging) mode. The ART-mode, in which frames are continuously aligned and averaged into one image, was not used in this study because it did not allow for examination of individual frames and the manual rejection of degraded frames (decreased signal, distortions due to movement) before calculating the mean image. The ART software currently rejects distorted frames but could be modified to include a rejection algorithm dealing with partially obscured frames.

**Internal Reference**

The requirements for the ideal fluorescence reference material are an efficiency that is close to that of fundus AF measured at the intermediate plane, an emission spectrum matching the...
The algorithm of van de Kraats and van Norren (Appendix F, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-76777/-/DCSupplemental); the rate of increase of qAF with age was then 8.6 and 2.7 qAF units/year for the four-segments and fovea, respectively.

**Figure 10.** qAF versus age for 20 subjects (age range, 20–50 year, all white) for the average of the four-segments (●) and for the fovea (○). Error bars are ±1 SD. Linear regression lines through the data showed a significant increase in qAF with age. For the four-segments, qAF = 201 + 5.22 × (age-20) \( r_{19} = 0.70; P = 0.0005 \), and for the fovea, qAF = 76 + 1.35 × (age-20) \( r_{18} = 0.52; P = 0.02 \). Dashed line: regression lines for the data (symbols omitted for clarity) of the four-segments and foveal sites after accounting for media absorption with the algorithm of van de Kraats and van Norren (Appendix F, Supplementary Material, http://www.iovs.org/lookup/suppl/doi:10.1167/iovs.11-8319/-/DCSupplemental); the rate of increase of qAF with age was then 8.6 and 2.7 qAF units/year for the four-segments and fovea, respectively.

**Measurement Noise**

Measurement errors associated with in vivo qAF determinations were more important than the influence of statistical noise of the AF signal and digitization electronics. This finding is evident from the near tripling of repeatability for within-session tests (within 6 seconds) on a stationary pattern, compared with that measured in subjects (Table 2) and in the near doubling of repeatability for within-session compared to between-session, same-day qAF of subjects (Fig. 8). In this regard, it is unlikely that the use of 16 frames/image, instead of the 9 frames/image used in this study, would substantially improve repeatability, since it would reduce only (by 33%) the relatively small statistical noise.

The positive correlation between the ΔqAFs at different sites and the large percentage (30%-50%) of image pairs where the qAFs of all sites either increase or decrease, albeit not uniformly, suggest that common effects in the anterior eye are responsible for these errors. Such effects may be changes in the tear film, in focus, and in the axial and lateral alignment of the camera with the eye. Although errors in focus appear to be a minor possibility, axial and lateral adjustment of the camera results in a ±5% change in qAF over ranges applicable to a skilled operator. However, asymmetries are also frequently observed, possibly due to asymmetries in the optical elements of the eye, to differences between the curvature of the retina and that of the image plane, or to minor misalignment in the camera optics. Eye movements would compound these effects, particularly if pupil dilation were insufficient. However, at this point, we are unable to separate the sources of error, most likely because they may all occur simultaneously and their effects are probably correlated with each other.

**Repeatability and Sample Size**

Monitoring lipofuscin levels in eyes could be performed in many situations (e.g., aging, drug, and gene therapy) with large sampling areas (such as the four-segments), since lipofuscin would be equally affected in all areas. Using the four segments has the additional benefit of balancing out some asymmetric variability, providing better repeatability. As examples, we used repeatability to estimate sample sizes for two hypothetical studies. These use our repeatability, but it is best to have estimates for the conditions of the planned study (e.g., multicenter). Example 1: does an oral drug stop foveal lipofuscin accumulation in healthy eyes? If we presume that the initial qAF four-segment average is 200 qAF units, we felt that this may be a reasonable way to define the qAF unit.

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500 qAF units) by 75 qAF units, one fourth of the difference between them and normal qAF at age 20 yr (200 qAF units)? Presuming that repeatability is 16% (twice the value used above for the normally sighted subjects in our study), the SD will be 0.16 × 500/1.96 = 41 qAF units, so for α = 0.05 and β = 0.20, the required sample size would be five.

**Pre-RPE Light Absorption**

Quantitative fundus AF is not an absolute measurement of lipofuscin because of light losses in the ocular media and in the layers of the neurosensory retina. The measured AF would, however, correlate with the amount of lipofuscin, since the latter is the principal source of AF and the effect of all pre-RPE absorbers, outside of the fovea, is relatively small in comparison. Although some of these losses can be accounted for by appropriate protocol or corrections, some cannot.

To date, we have not corrected our data for the absorption of the ocular media. This correction is important for studies where the study population contains individuals with a large range of ages (e.g., aging and longitudinal studies). Several psychophysical and physical methods are available for the individual estimation of the optical density of the media. In some studies, such as a comparison of groups of young and old subjects, it may be sufficient to employ an algorithm that predicts the average media optical density for a given age.39,50 We have used one such algorithm50 to illustrate the effect of this correction (Fig. 10).

Absorption by photopigment reduces the excitation light to the RPE by amounts that vary with the type and quantity of photopigment, the wavelength used, and the fraction of bleached pigment. For rods, the time course of bleaching recorded in three subjects (Fig. 5) corresponded well with theoretical predictions based on the two-compartment model (with regeneration).34–37 Cones, on the other hand, cannot be efficiently bleached with the retinal irradiances available in the two cSLOs, in part because macular pigment reduces the irradiance on the cones and because 488 nm is spectrally remote from the peak absorption by cones (550 nm). Thus, studies of foveal lipofuscin should be made with an imaging system that uses an excitation wavelength longer than 540 nm, where macular pigment absorption is very low.

Furthermore, interpretation of qAF levels may have to account for light losses in the nerve fiber layer,52 the neural layers of the retina (e.g., OCT reveals multiple reflections and hence light losses from within the retina), retinal capillaries,53 and RPE melanin. The latter reduces the lipofuscin AF by amounts that are dependent on age, the spatial distribution of melanin, and the apical/basal distribution within the RPE cell.54–56 We estimated that lipofuscin AF may be attenuated by a mean factor of ≈1.2 (age range, 20–70 yr).55,57 Unknown at this point is the contribution of melanolipofuscin58 to the total qAF.

**Conclusions**

Quantitative AF imaging appears feasible with current equipment, but several technical changes could be implemented to reduce measurement errors. In addition to the new alignment software, the detection pupil could be reduced from 6 to 5 mm to decrease obstruction by the iris, automatic software rejection of degraded frames could be implemented, and additional optics could be introduced to improve the centering of the camera on the eye. Furthermore, the use of a green laser to excite the AF could minimize the effect of macular pigment and the errors associated with ocular media correction. We are developing a reflectometry method47–49 to individually estimate the light losses in the media. Further studies will focus on more specifically identifying the sources of measurement error, optimizing image acquisition and analysis protocols, and establishing a normative database of subjects with normal retinal status. Finally, hardware installations and software development for quantitative analysis will be enhanced for portability and ease of use in multicenter studies.

By offering a clinically accessible standard against which to measure AF intensity, qAF will not only facilitate clinical research, but will also offer potential diagnostic, prognostic, and therapeutic applications.

**Acknowledgments**

The authors thank Tilman Otto (Heidelberg Engineering) for his numerous insights and for developing the modified alignment software.

**References**


