A Model for the Effect of Disturbances in the Optical Media on the OCT Image Quality

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Purpose. The loss of quality of optical coherence tomography (OCT) images resulting from disturbances in the optical media has been modeled.

Methods. OCT measurements were performed in two healthy volunteers using time domain (TD)-OCT (StratusOCT; Carl Zeiss Meditec, Dublin, CA). Optical disturbances were approached in three ways simulated with filters. The studied effects were: light attenuation (absorptive and reflective filters), refractive aberrations (defocusing lenses), and light scattering/straylight (scattering filters). The same examiner scanned the subjects with the filters placed in front of the eye. The signal strength (SS) values of the scans were then collected. The strength of the filters were expressed in optical density (OD), determined for the 830 nm central wavelength of the OCT, ($OD_{\lambda=830}$).

Results. A linear relationship has been found between the $OD_{\lambda=830}$ of the absorptive and reflective filters and the SS of the corresponding OCT images. Assuming that reduction of light from the OCT scanning spot on the retina is the critical factor, this light loss was determined for the scattering filters and defocusing lenses. A comparable linear relationship was found between the SS value and the $OD_{\lambda=830}$ of these filters.

Conclusions. The model indicates that the loss of OCT image quality in patients with disturbances in the optical media is explained by attenuation of the light in the OCT scanning spot on the retina. A linear relationship between the SS and the single pass logarithmic attenuation of the OCT signal is shown, according to SS = constant - (9.9 [−9.4 to −10.6] $\cdot$ $OD_{\lambda=830}$). (*Invest Ophthalmol Vis Sci.* 2009;50:787–792) DOI:10.1167/iovs.08-2364

In 1995, optical coherence tomography (OCT) was introduced in ophthalmology as an imaging technique for in vivo imaging of the human retina.[1] Nowadays, OCT has become a widely established and useful diagnostic tool, especially in diseases of the macula and vitro-retinal interface.[2–9] This technology is a noncontact, noninvasive imaging technique of the human retina that uses low coherence interferometry of light. Image formation depends on differences in optical backscattering properties of the tissue under investigation.

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OCT is an optical measurement technique and its image quality is influenced by opacities in the optical path. In the elderly population, cataract is the most common cause of media opacity that is known to influence the diagnostic OCT image quality and measurements. This study focused on the influence of optical disturbances, in particular cataract, on the OCT image quality.

In the first versions of OCT software (StratusOCT; Carl Zeiss Meditec, Dublin, CA), the only parameter to objectively evaluate the quality of acquired images was the signal-to-noise ratio (SNR). The SNR takes into account the single a-scan that demonstrates the strongest signal and does not account for the distribution of the signal strength throughout the scan image. Currently, the quality of the obtained OCT images is automatically reported by the software of the time domain (TD)-OCT (StratusOCT) as a metric called signal strength (SS). The SS, ranging 0 to 10, is based on the originally used SNR and the uniformity of signal strength within a scan. Due to its proprietary nature, the manufacturer does not further specify the SS. Perhaps the SS value is not the best image quality score, but at present it is the only parameter provided by the TD-OCT software for clinical use.[10] For reliable clinical measurements, the quality of the images needs to be sufficient; therefore, some studies defined thresholds for the SNR and SS that allow accurate measurements.[10,11] Opacities (in particular cataract) and OCT image quality is a well studied subject. Several studies[11–13] investigated the influence of cataract on the SNR showing a significant increase in signal quality after cataract removal. Furthermore, OCT based investigation of macular changes after cataract surgery has drawn much interest. Most of the investigators studying this topic[11,13–19] found a subtle but significant increase in retinal thickness (RT) after cataract surgery, whereas others found no differences in RT or demonstrated a decrease of RT in their postoperative patients.[13,20] A number of published studies[11,13,17] discussed whether the change of the optical quality of the media may have confounded these results.

Optical imaging can be described by means of the so-called point spread function (PSF). The PSF is the light distribution on the retina of a point of light that is focused on the retina. Whereas the ideal PSF would be a small point, the true PSF is spread out. Even in a healthy eye, the beam of light is spread out slightly, but cataract introduces more profound changes. Opacification of the lens causes refractive irregularities in the optical pathway, degrading the quality of retinal imaging. Refractive disturbances, for example lower and higher order aberrations, affect the central peak of the PSF, causing spreading over angles of around 0.1°. Irregular particles in the ocular system of small extent (on the order of 10 μm) cause very different effects. They cause scattering of light and affect the skirts of the PSF, causing light spreading over angles larger than 1°. This light is usually called straylight.[21–24] Apart from refractive type disturbances and scattering type disturbances, a third type of disturbance can be distinguished. Light is lost due to both absorption in the media and also, for a minor part, due to reflection.

The purpose of this study was to develop a model using artificial means to describe the effects of optical disturbances, in particular those introduced by cataract, on the OCT image quality. In summary, a cataract was approached by studying three effects: (1) light-attenuation (absorption and reflection), (2) refractive type disturbances (defocus/aberrations) and (3) light scattering/straylight.

Materials and Methods

Subjects

OCT measurements of two healthy volunteers (27 and 51 years old) were performed using the TD-OCT (StratusOCT, software version 4.0.1; Carl Zeiss Meditec, Dublin, CA). The research followed the tenets of the Declaration of Helsinki and verbal informed consent for a Medical Ethical Committee–approved protocol was obtained from the subjects.

Artificial Filters

Optical disturbances were approached using the three main effects as mentioned above and each was simulated with artificial filters. A series of each type of filter was used to cover the range of disturbances as can be observed in the clinic. The three effects and corresponding filters were: (1) light attenuation, simulated with absorptive (Schott, Mainz, Germany; $n = 3$) and reflective (Balzers, Balzers, Liechtenstein; $n = 8$) filters; (2) refractive aberrations, simulated with defocusing lenses ($n = 6$) because no physical models are as yet available that mimic the aberrations of cataracts; and (3) light scattering/straylight, mimicked using scattering filters ($n = 7$) that were discussed earlier as potential models for the light scattering characteristics of cataracts.[25] The different types of filters used in the photographic industry and in the study by de Wit et al.[25]—Lee (Lee Filters, Hampshire, England), SO (Cataract simulation glasses; Stereo Optical Co., Inc., Chicago, IL), BWF (B+W fog; B+W Filterfabrik Johannes Weber GmbH & Co., Bad Kreuznach, Germany), BPM (Black Pro-Mist; Tiffen, Hauppauge, NY), and Hoya (Tokina Co. Ltd., Tokyo, Japan)—proved to cover a wide range of these scattering characteristics. For the present study, we also used a P087 filter (Cokin S.A.S., Silic, France). The BPM and SO filters proved to resemble cataract relatively well (within the range of 2.5°–40°). The BPM filters represent realistic levels of light scattering, whereas the SO filter represents more extreme levels which are uncommon in clinical practice.[25] The strength of all types of filters was expressed in optical density. This optical density was determined for the 830 nm central wavelength of the used OCT system ($OD_{\lambda=830}$), as described in the model section below.

OCT Acquisition

After pupil dilation, b-scans of the macula were acquired with the 6-mm line scan protocol. During scanning, no filter and the model filters ($n = 24$) were placed in front of the eye respectively. Accurate care was taken in positioning of the filters by checking for potential tilt of the filters. This tilt was estimated to be <10°. In this manner a maximum error of 2% was accepted. All measurements were done twice at two time points (i.e., four measurements per filter in each person) by the same experienced examiner. The SS value of each scan as provided by the OCT software was collected as parameter for the OCT image quality.

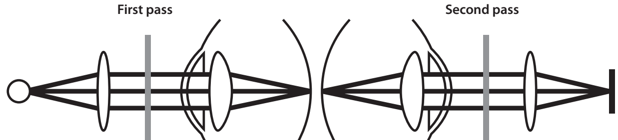

Figure 1. Schematic illustration of the first and second passes in OCT scanning with a filter placed in front of the eye.

Table 1. $OD_{\lambda=830nm}$ Values (single pass) of the Attenuating Filters

Filter Nominal Value (%)	$T_{\lambda=830nm}$ (%)	$OD_{\lambda=830nm}$ (single pass)
Absorptive		
15	66.6	0.18
30	41.7	0.38
60	30.2	0.52
Reflective		
97	96.1	0.02
79.2	91.3	0.04
63	78.2	0.11
51.2	55.1	0.26
25.4	28.8	0.54
19.6	21.3	0.67
16.1	17.9	0.75
11.5	12.0	0.92

Statistical Analysis

Statistical analysis was performed on the measurements as independent values (SPSS 12.02 software; SPSS Inc, Chicago, IL). The Pearson's correlation test was performed to investigate the correlation between the SS values and the $OD_{\lambda=830}$ of the filters. Throughout this article, Briggsian logarithms (base 10) were used.

Basics of the Model

In OCT the eye is illuminated using a coherent light source from either a laser or a superluminescent diode and like the scanning laser ophthalmoscope (SLO), its energy is delivered to a small spot on the retina. Because of the confocal principle applied in the OCT apparatus, the detected light derives also from this spot. The size of the spot on the retina is 20 μm, corresponding to 4 minutes of arc. Attenuation through absorption and reflection is the simplest type of image degradation; in both passes in OCT scanning the beam of light is equally reduced while passing these filters. In Figure 1, the first and second passes in OCT scanning are schematically illustrated.

In optical literature it is common to express the attenuation of a filter using optical density, which is wavelength dependent (OD_{λ}). The fraction of light that passes through a sample is known as the transmittance (T_{λ}). OD_{λ} is defined as $-\log T_{\lambda}$. An OD_{λ} of 0.3, for example, corresponds to a T of 0.5 ($T_{\lambda\%} = 50\%$). Since the central wavelength of the OCT light source is 830 nm, the absorptive and reflective filters were spectrometrically tested in an optical setup and the 830 nm values were used in this study, shown in Table 1.

Preliminary results demonstrated a linear relationship between the SS and the $OD_{\lambda=830}$ of the attenuation filters, suggesting that the SS value purely represents attenuation of the recorded light by the OCT. Hence, the loss of light from the scanning spot, which determines the loss of OCT image quality (decrease in SS) when using the scattering filters and defocusing lenses, was investigated. In case of scattering, light is redistributed and some is lost from this spot before it falls on the retina (first pass), so attenuation to this small spot on the retina can be expected. The loss of light, caused by each of the light scattering filters at a wavelength of 830 nm, was measured in an optical setup, using an artificial diaphragm to mimic the 20-μm area at the retina. The measured light losses (single pass) were transformed to $OD_{\lambda=830}$ values (depicted in Table 2) and plotted against the SS values obtained by placing the scattering filters in front of the eye.

Subsequently, the effect of refractive errors (aberrations) was studied, simulated with defocusing lenses. In case of defocusing, a blur circle is introduced on the retina instead of a sharp point-wise image. The diameter of the blur circle is equal to $d \cdot D$, where d is the pupil diameter and D is the refractive error. The blur circle is expressed in radians (1 radian = 57°). For example, for a pupil diameter of 4 mm

TABLE 2. $OD_{\lambda=830nm}$ Values (single pass) of the Scattering Filters

Filter	$OD_{\lambda=830nm}$ (single pass)
Scattering I (P087)	0.06
Scattering II (BWF1)	0.17
Scattering III (BWF2)	0.29
Scattering IV (BPM1)	0.29
Scattering V (BPM2_1.14)	0.36
Scattering VI (BPM2_1.18)	0.40
Scattering VII (SO)	0.47

($d = 4 \cdot 10^{-3}$ m) and a refractive error of 1 diopter ($D = 1\ m^{-1}$), the diameter of the blur circle is $4 \cdot 10^{-3}$ radians, which is equal to 0.228° or 14 minutes of arc (1° = 60 minutes of arc). When the blur circle exceeds the size of the scanning spot on the retina of 4 arc minutes in diameter, attenuation of the retinal illumination can be expected. In the above example, the intensity in this 4 arc minute area would be lowered by a factor $4^2/14^2 = 0.08$, corresponding to an OD_λ of 1.10.

However, preliminary results indicated that the influence of refractive errors on the OCT signal is smaller as a consequence of a different effective entrance pupil. When correcting for the effective pupil diameter, one has to consider the fact that in the recording process (second pass) of the OCT, the projected blur circle is blurred again. However, this second blurring does not enlarge the spot again by a factor of three as in the example above, but the diameter of the blur circle increases by approximately a factor of $\sqrt{2}$. This was taken into account by plotting the SS values (obtained by placing the defocusing lenses in the OCT/eye optical path) against ½ ([first pass effect] + [second pass effect]), with the pupil diameter as free parameter.

RESULTS

In Figures 2 and 3, the SS values of the scans acquired with the attenuation filters (pure absorption [Fig. 2] and pure reflection [Fig. 3]) placed in the OCT/eye optical path are plotted against the $OD_{\lambda=830}$ values of these filters. The Pearson's correlation coefficients (r) of these data indicated a statistically significant ($P < 0.001$) linear relationship between the SS value and the $OD_{\lambda=830}$ value. For the absorptive filters, r was 0.96 (subject 1) and 0.95 (subject 2); for the reflective filters, r was 0.97 (subject 1) and 0.96 (subject 2). The average slope of the trend lines of the four sets of data points was −10.3. Both trend lines and a straight line with a slope of −10 are included in the figures. The SS value correspond to the logarithmic attenuation of the signal, averaging out at SS = constant − $10 \cdot OD_{\lambda=830}$. The offset of subject 2 was slightly higher compared to subject 1.

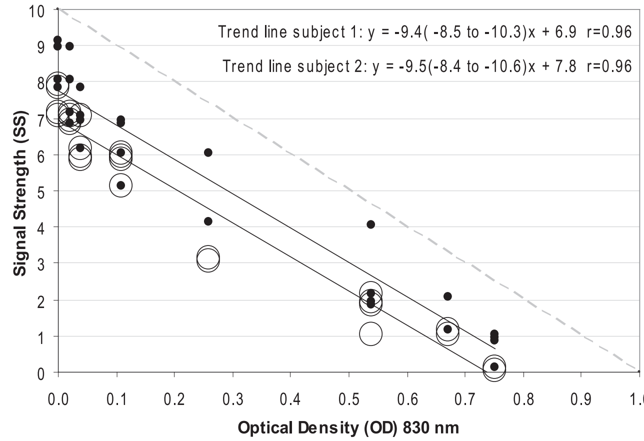

FIGURE 3. SS plotted against effective $OD_{\lambda=830}$ values (single pass) of reflective filters. *Open symbols*: subject 1; *filled symbols*: subject 2. Both trend lines (*black solid*) and a straight line with a slope of −10 (*gray dashed*) are included in the sets of data points. $P < 0.001$.

The linear relationship between the SS and the $OD_{\lambda=830}$ of the attenuation filters suggested that the SS value purely represents attenuation of the recorded light by the OCT instrument.

Figure 4 shows comparable results with Figures 2 and 3, when inserting light scattering filters instead of attenuation filters. The Pearson's correlation coefficients (r) showed the same linear relationship between the SS value and the $OD_{\lambda=830}$ value for both scattering filters and attenuation filters. The r values were 0.93 (subject 1) and 0.92 (subject 2), both statistically significant at $P < 0.001$.

The decrease in SS value actually found for refractive errors from 1 to 12 diopters (D) is presented in Figure 5. Note that an error of 1 D showed only a minor effect on the SS value, which was in contrast with the SS expected for an attenuation of 0.95 with an effective entrance pupil diameter of 4 mm.

The effective pupil diameter of the OCT light bundle was estimated, using the measured SS values and their corresponding refractive error D, by fitting the data points of Figure 5 to

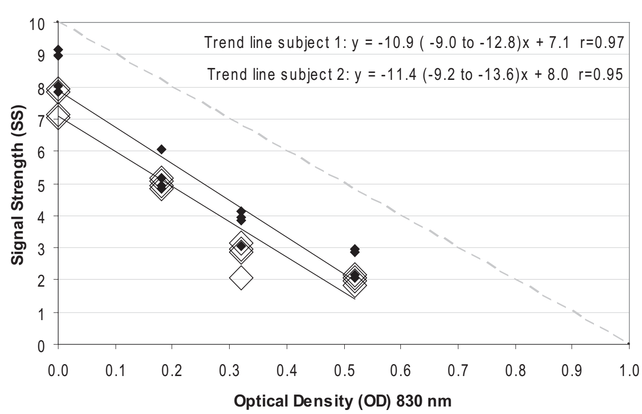

FIGURE 2. SS plotted against effective $OD_{\lambda=830}$ values (single pass) of absorptive filters. *Open symbols*: subject 1; *filled symbols*: subject 2. Both trend lines (*black solid*) and a straight line with a slope of −10 (*gray dashed*) are included in the sets of data points. The Pearson's correlation coefficients (r) of the data, as shown in the figure, are statistically significant ($P < 0.001$).

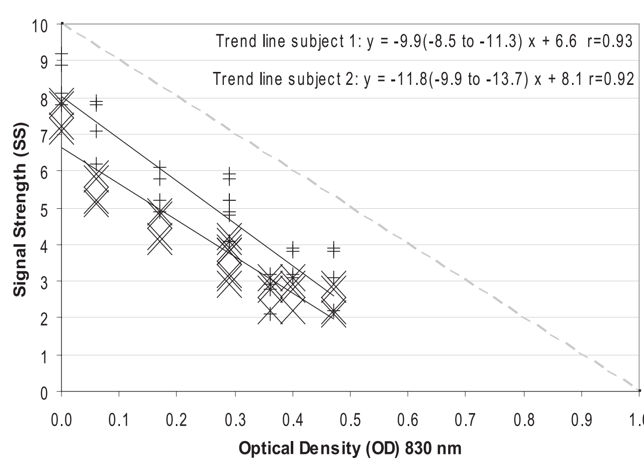

FIGURE 4. SS plotted against effective $OD_{\lambda=830}$ values (single pass) of scattering filters. *Large symbols*: subject 1; *small symbols*: subject 2. Both the trend lines (*black solid*) and a straight line with a slope of −10 (*gray dashed*) are included in the sets of data points. $P < 0.001$.

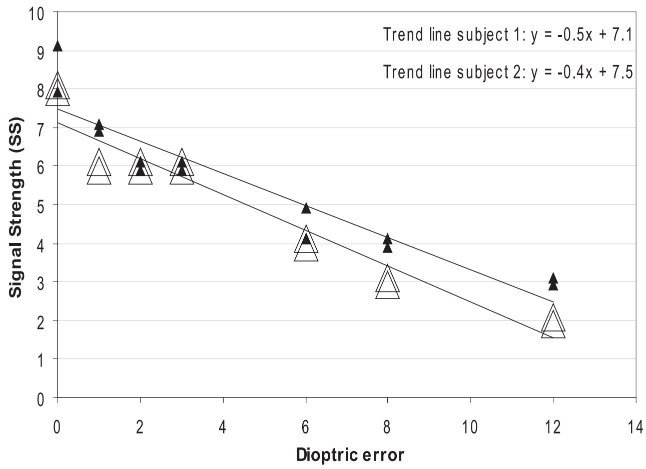

FIGURE 5. SS plotted against refractive error raging 1 to 12 diopter (D). *Open symbols*: subject 1; *filled symbols*: subject 2.

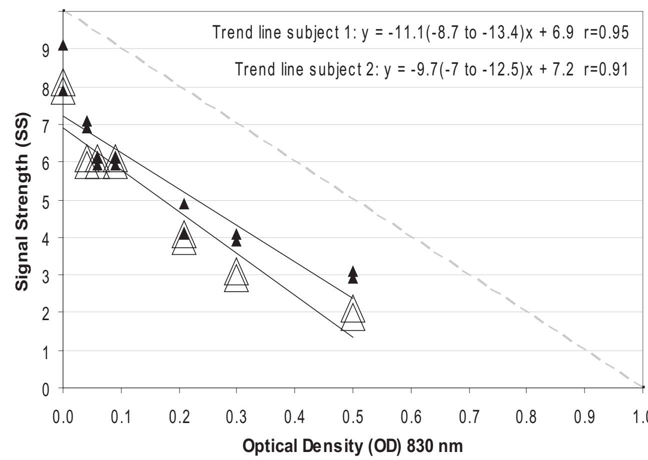

FIGURE 6. SS plotted against effective $OD_{\lambda=830}$ values (corresponding to single pass) of defocusing lenses. *Open symbols*: subject 1; *filled symbols*: subject 2. Both the trend lines (*black solid*) and a straight line with a slope of −10 (*gray dashed*) are included in the sets of data points. $P < 0.001$.

the characteristic given in Figures 2, 3, and 4. This resulted in an effective entrance pupil diameter of 0.32 mm. Using this estimated diameter, the corresponding OD values were calculated corresponding to a one pass transmission (Table 3). In Figure 6, the SS values are plotted against the effective $OD_{\lambda=830}$ values (corresponding to single pass) of the defocusing lenses. The *r* values were 0.95 (subject 1) and 0.91 (subject 2), both statistically significant at $P < 0.001$.

Figure 7 shows all the data of both subjects after correcting for the offset of the subjects. The trend line is according to SS = constant - 9.9 · $OD_{\lambda=830}$. The Pearson's correlation coefficient of 0.93 ($P < 0.001$) together with normally distributed residuals (data not shown) demonstrate the linear relationship between the SS and the single pass attenuation of the OCT signal.

In Figure 8, example images of subject 2 illustrate approximately the same decrease in SS for all types of filters at almost identical $OD_{\lambda=830}$ values, in comparison with a scan at the highest $OD_{\lambda=830}$ value and a scan without a filter placed in the OCT/eye optical path.

DISCUSSION

In this study, we attempted to describe the effects of optical disturbances, in particular those introduced by cataract, on the OCT image quality by simulating opacifications with filters. Due to the complexity of optical disturbances, in particular in cataract, we simplified the optical disturbances and approached them by introducing three main effects on the optical light path, namely pure attenuation, scattering, and refraction. It was found that loss of OCT image quality, due to disturbances in the optical media, is fully explained by attenuation of the light in the OCT scanning spot on the retina.

TABLE 3. Calculated $OD_{\lambda=830nm}$ Values (single pass) of the Defocusing Lenses, Based on an Estimation of the Effective Pupil Diameter of 0.32 mm

Spherical Lens (Diopter)	$OD_{\lambda=830nm}$ (single pass)
1	0.04
2	0.06
3	0.09
6	0.21
8	0.30
12	0.50

It should be mentioned that in the figures throughout this article the SS value was plotted against the single pass (or corresponding to single pass) OD values. The three different aspects of media disturbances work out differently in both passes in OCT scanning. The first component, attenuation of light intensity, works straightforward in this respect. It causes equal attenuation in both passes such that the total attenuation in log units equals two times the single pass attenuation. The light scattering component is more complicated. If the spot size would have been relatively large, then the first and second passes would have shown different effects. The light scattered outside the spot in the first pass re-illuminates the spot size on the second pass. However, the spot size of 4 minutes of arc is relatively small and since scattering causes light spreading over much wider angles (in degrees rather than minutes of arc), the contribution of this re-illumination might be negligible. In case of the aberration type disturbances, there is an important difference between the first and second pass. Due to a reoccurred blurring of the spot in the second pass, but to a lesser extent, the first pass attenuation is much larger compared to that of the second. This was taken into account in the fit formula which gives an estimate for the effective pupil diam-

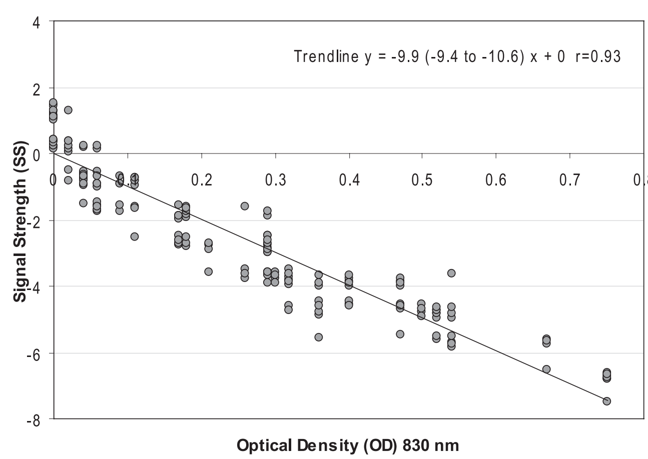

FIGURE 7. Data points of all filters of both subjects after correction for their offset are plotted and a trend line is fitted into the data points. $P < 0.001$.

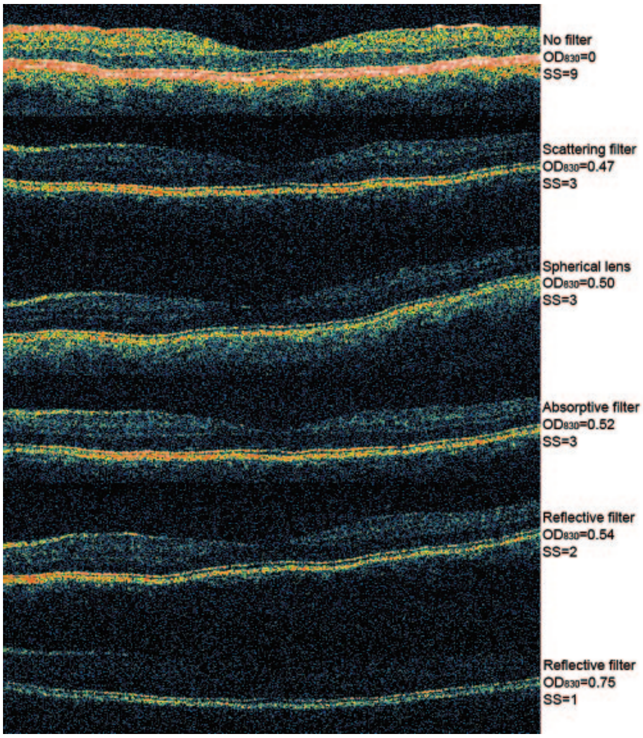

FIGURE 8. Example images of subject 2 illustrating approximately the same decrease in SS for all types of filters at roughly identical $OD_{\lambda=830}$ values in comparison with a scan at the highest $OD_{\lambda=830}$ value and a scan without a filter placed in the OCT/eye optical path.

eter (Fig. 6). The 8% reflectance, inherent to the use of a filter in the optical path, was also substituted in this formula.

Approaching a cataract by introducing absorption and reflection, that is, pure attenuation, is in principle straightforward. This component is well researched as knowledge of the spectral transmission in the eye media is important.[26–28] The major absorption properties of the lens are due to its chemical characteristics. Detailed studies have separated out the true light attenuation effects and seeming light attenuation effects that can be caused by light scattering.[29,30]

The insight in ophthalmic light scattering was developed originally from studies on the phenomenon of glare. Glare is a common visual symptom in cataract patients.[31] It was found that glare could fully be understood on the basis of the optical phenomenon of straylight on the retina.[32] Straylight is light that affects the skirts of the PSF, causing light spreading over angles larger than 1°. There is sufficient knowledge about the characteristics of light spreading in the human eye, ranging from 2.5°–40°.[22,24,33] In the present model, we simulated the scattering characteristics of the human cataract using the filters as described in the study by de Wit et al.[25] and one additional filter.

Refractive type errors introduced by optical disturbances can be described with wavefront aberrations and might result from irregularities at the surface of, for instance, the opacified lens. Wavefront aberrations introduce an irregular blur circle on the retina. On the assumption that the precise shape of the blur circle is not a critical factor in OCT image quality loss and because no physical models that mimic the complex aberrations of cataracts are thus far available, the refractive type aberrations were simulated with spherical defocusing lenses.[34,35]

Recently, Tappeiner et al.[36] modeled the impact of opacities in the optical pathway on OCT image quality by using neutral-density (i.e., absorption) filters. Their data showed a certain decline with density but not in a quantitative way because they did not correct the nominal values of the filters for the wavelength of the OCT. The present study demonstrated that the nominal values of the filters deviate strongly from the true values valid in OCT.

Simulating opacities in the optical system with filters has its limitations. In actual practice, the contribution of the three mentioned components will exist in a true cataract. Due to the approach used in this study, a systematic disturbance of the OCT signal by the different components could be demonstrated. The contribution of the three different components to the loss of light in the OCT scanning spot on the retina may differ considerably in a cataract patient. Since the OCT uses near infrared light, the attenuation component may in practice be very small.[37] Regarding light scatter, it has been found that for cataract, straylight is weakly dependent on the wavelength.[24,29,33] Refractive type errors (aberrations) can, for the purpose of this discussion, be assumed to be virtually independent of wavelength. Since disturbances in the optical media—and in particular cataract—are very complex, the present model approximates the effects of cataract on the OCT image quality. Other components of influence on the OCT signal are differences in reflectance of the retina which will be different for each individual. Notice that in the present study a different constant of the straight lines for both subjects was found. The difference in offset of the straight lines may however also reflect the difference in light loss in both individuals. More research is necessary to quantify the influence of the reflectance of the retina on the OCT signal. Another potential additional effect which can have influence on the OCT light bundle as introduced by disturbances in the optical media is polarization. However, all scans were acquired with optimal polarization and the filters used in our study are not polarizing.

As can be observed in Figure 8 at $OD_{\lambda=830}$ values around 0.5, the intra-retinal layers are more difficult to distinguish, and at an $OD_{\lambda=830}$ value of 0.75, no intra-retinal structures could be observed. The influence of media opacities on the segmentation of retinal layers by OCT software algorithms has been discussed in the literature.[38–41] We think that the present model does provide an important insight into the effects of disturbances on the OCT image quality. This can be useful in the investigation of the influence of cataract on segmentation of retinal layers and retinal thickness measurements by means of OCT, without the possible confounding effects of a cataract operation. Ongoing research confirms our model in time domain OCT and expectations are that spectral domain OCT will be influenced by media opacities in the same way.

In summary, we conclude that according to the presented model, the loss of OCT image quality in patients with disturbances in the optical media, expressed in SS, is explained by attenuation of the light in the OCT scanning spot on the retina. The SS value indicates the single pass logarithmic attenuation of the OCT signal, according to $SS = constant - (9.9\ [-9.4 \text{ to } -10.6] \cdot OD_{\lambda=830})$.

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