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High-Quality Tissue Imaging Using a Catheter-Based Swept-Source Optical Coherence Tomography Systems With an Integrated Semiconductor Optical Amplifier

Youxin Mao, Costel Flueraru, *Senior Member, IEEE*, Shoude Chang, Dan Paul Popescu, and Michael G. Sowa

Abstract—We present the analysis and characteristics of an optical fiber catheter-based swept-source optical coherence tomography (SS-OCT) system for tissue imaging. A quadrature Mach-Zehnder interferometer based on multiport fiber couplers and a semiconductor optical amplifier were employed in our SS-OCT system in order to increase the signal-to-noise ratio, therefore improving image quality deeper within the tissue. We demonstrate OCT images of human skin and arterial tissue acquired with our OCT engine. The results show that this OCT system is an effective imaging modality for real-time diagnosis and *in vivo* biological applications.

Index Terms—Biomedical imaging, optical coherence tomography (OCT), optical fiber probe, quadrature interferometer, semiconductor amplifier.

I. INTRODUCTION

OVER the recent decades, the introduction of several imaging technologies has been crucial in reducing mortality associated with conditions such as skin cancer and arterial plaque. Optical coherence tomography (OCT) [1] is a recently introduced imaging modality that has shown considerable promise for the identification of high-risk arterial plaques. The advantages of OCT, compared with ultrasound, include its higher resolution, video speed acquisition rate, compactness, and portability. Because OCT uses light, a variety of functional and spectroscopic techniques are available to expand its capabilities, including polarization, absorption, elastography, Doppler, and dispersion analysis. When a small optical fiber probe, which is part of an optical catheter, constitutes the sample arm, the system becomes suitable for intravascular probing [2]–[4]. An OCT system with high signal-to-noise ratio (SNR) is important for imaging turbid tissue, such as arterial plaques, because the backscattered signals from these types of samples are extremely weak. Swept-source OCT (SS-OCT) has received much attention in recent years not only because of its higher SNR at high imaging speeds but also for its

capability to image using the 1300-nm wavelength range, where the reduced light scattering by tissue enables OCT to collect signals from deeper within the structure when compared with OCT imaging based on shorter source wavelengths. SS-OCT can also use a quadrature interferometer based on multiport fiber couplers such as a 3×3 quadrature interferometer [5], [6]. When measuring instantaneous complex signals with stable phase information using a 3×3 quadrature interferometer, the complex conjugate artifact can be suppressed, therefore doubling the effective imaging depth [7]. The phase information of the complex interferometric signals can also be exploited to gain additional information about the tissue, enhance image contrast, and perform quantitative measurements. In addition, the Mach-Zehnder configuration of the presented interferometric setup allows different options for optical power distribution between the reference and sample arms. An unbalanced input directs more optical power from the light source to the sample than that to the reference mirror [8], whereas balanced detection is used to reduce the beat noise [9]. Both techniques play their parts in increasing SNR. However, in the case of imaging turbid biomedical samples, the signal backscattered from tissue is much weaker than the reference signal, so an attenuation of optical power in the reference arm is required in order to increase the SNR.

The ability of OCT to image skin cancer and vascular plaque has been previously demonstrated [10]–[13]. However, OCT images of skin and arterial wall are limited to depths of ~ 1 mm even using the Fourier domain methods. The limited imaging depth into tissue is one of the most serious limitations for OCT to be used as a routine clinical imaging method. Further improvement of the SNR of OCT is needed in order to increase the imaging depth of OCT into tissue. Adding an optical amplifier in the path of the backscattered signal in the sample arm of an SS-OCT system with a balanced Michelson interferometer has been proposed [14]. About 2-dB improvement of SNR has been reported in the 2×2 MZI configurations [15], [16]. However, according to our knowledge, no OCT image improvement in cardiology applications has been reported.

In this paper, the characteristics of a catheter-based SS-OCT system with a quadrature interferometer (QSS-OCT) using a semiconductor optical amplifier (SOA) for the amplification of the weak signal existent in the sample arm are demonstrated [17]. Improvement of the SNR is theoretically analyzed and proved with experimental results. *In vivo* and *ex vivo* OCT

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Fig. 1. Experimental setup of our QSS-OCT system with balanced 3×3 quadrature MZI and an SOA for the amplification of the signal backscattered from the sample.

imaging of various tissues are demonstrated. Lipid-rich plaque of a Watanabe heritable hyperlipidemic (WHHLMI) rabbit is visualized and characterized with this system. Preliminary results show that vulnerable plaque with fibrous cap, macrophage accumulations, and calcification in the arterial tissue are measurable with our OCT system. The system offers the possibility to investigate tissue morphology, revealing a microstructure located on and under the tissue surface up to a 2-mm depth.

II. METHODS

A. OCT system

Fig. 1 shows an experimental setup of our QSS-OCT system that uses a balanced 3×3 with a 2×2 quadrature MZI and a SOA for weak sample signal amplification. The swept laser source (HSL2000-HL, Santac) used in the setup has a central wavelength of 1320 nm and a full scan wavelength range of 110 nm, and was swept linearly in optical frequency with a linearity of 0.2%. The bandwidth of the source corresponds to an 8- μm OCT imaging resolution in air. The average output power and coherence length of the swept source was 12 mW and 10 mm, respectively. A repetition scan rate of 20 kHz was used in our system, and the related duty cycle was 68%. The light output from the swept laser source was launched first into the 2×2 coupler, where 90% of the power was diverted toward the sample. The reference arm was arranged with a fiber collimator and a mirror. The light was directed to the sample through a lensed single-mode fiber probe [18]. A galvanometer (Blue Hill Optical technologies) scanned the fiber probe along the sample surface up to 4 mm-long trip corresponding to an OCT image width of 900 pixels. The weak (< -30 dBm) light backscattered from the sample was fed into a SOA (Covega) whose gain can be adjusted by a variable attenuator connected after the SOA. The SOA had the same center wavelength and bandwidth as the swept source. The gain could be varied from 0 to 35 dB. The reference arm has been built, so that it can match the optical distance of the sample arm without the SOA included. After introduction of the SOA, an optical jumper with a length matched to the new optical distance of the sample arm is added. Both the SOA and the added optical jumper (part of reference arm) could be removed, allowing the system to be switched back to a regular QSS-OCT system

without sample signal amplification. A polarization controller was inserted before the SOA for optimal amplification. The amplified signal was combined with the signal returning from the reference mirror through the 3×3 and 2×2 couplers, thus implementing a dual-channel balanced detection system with two complementary components of the complex interferometric signals and thus suppressing the complex conjugate artifact. Both balanced detectors (PDB150C, Thorlabs) used in this system had saturation powers of 5 mW. We selected a 3-dB bandwidth of 50 MHz to give sufficient imaging depth. The two detector outputs were digitized using a data acquisition (DAQ) card (Alazartech, Montreal) with 14-bit resolution and acquired signal at a sampling speed of 100 MS/s. The swept source generated a start trigger signal that was used to initiate the function generator for the galvanometer scanner and initiate the DAQ process for each A-scan. Because the swept source was linearly swept with the wavenumber k , A-scan data with resolved complex conjugate artifact were obtained by a direct inverse Fourier transformation (IFT) from the DAQ sampled data without performing an additional resampling step.

B. Tissues and Probes

Samples of human skin, healthy arterial tissues of mice, and an arterial plaque from WHHLMI rabbits were used in this work. WHHLMI rabbit is a suitable animal model to study hypercholesterolemia and atherosclerosis. Arteries in these rabbits develop atheromatous plaques similar to those in humans [19]. Fig. 4(a) shows a segment of the descending aorta, together with the protected forward-view ball fiber probe collecting an OCT image. The size of the OCT probe and the properties of the probing beam are important for OCT imaging. An optical fiber-lensed probe with a diameter as small as 0.5 mm is suitable for OCT imaging, particularly for *in vivo* intravascular imaging. Based on the interaction of near-infrared light with different human tissues, the penetration depth ranges from 0.5 to 3 mm [20]. The working distance range of an ultrasmall fiber-based lens [18] can be designed up to 1.2 mm to match the penetration depth into the tissues tested. Due to the tradeoff between the depth of field and beam spot size for a Gaussian beam, the depth of field can be in the range of 0.3–1.5 mm, which corresponds to a spot size range of 15–35 μm at 1300-nm wavelength.

The theoretical and experimental results of the spot size and working distance versus the length of the GRIN fiber or the diameter of the ball lens (bottom x -axis), and the length of the fiber spacer (top x -axis) are shown in Fig. 2(a) and (b), respectively. The typical measured results of the beam properties are listed in Table I, along with detailed descriptions of the samples. Numbers 1 through 4 correspond to the GRIN fiber lenses, and those from 5 and 6 to fiber ball lenses. The measured $1/e^2$ intensity beam diameters along the axial distance, the measured beam profile, and the schematic diagram of the lens structure are shown in Fig. 3. By comparing the working distance with the depth of field and the spot size, it became clear that the optimized parameters (i.e., working distance of 0.9–1.2 mm, depth of field of 0.9–1.1 mm, and spot size $< 30 \mu\text{m}$) were not at the position of the largest working distance. Instead, the optimal positions corresponded to a 0.17-mm length of the GRIN fiber

Fig. 2. Theoretical and experimental results of (a) working distance and (b) spot size versus the length of GRIN fiber or diameter of the ball lens (bottom x -axis) and length of fiber spacer (top x -axis), where lines represent the theoretical results from ZEMAX at 1300 nm. Among them, dark dotted lines represent GRIN fiber lens without a fiber spacer, dark and light solid lines represent GRIN fiber lens with a constant length of fiber spacer (0.48 mm) and a constant length of GRIN fiber (0.17 mm), respectively; and dark and light dash lines represent ball fiber lens with a constant length of fiber spacer (0.62 mm) and a constant diameter of the ball (0.30 mm), respectively. The related experimental results were represented by the points.

TABLE I
LENS STRUCTURES WITH MEASURED BEAM PROPERTIES

#	L_0 (mm)	Fiber Lens		Measured Beam Properties		
		Type	L/R (mm)	Working Distance (mm)	Depth of Field (mm)	Spot Size (μm)
1	0	50/125	0.41	0.11	0.17	13
2	0	100/140	0.52	0.28	0.50	22
3	0.36	50/125	0.12	0.50	0.60	23
4	0.48	100/140	0.17	1.00	0.95	28
5	0.70	Ball-Lens	0.15	1.00	0.50	20
6	0.62	Ball-Lens	0.15	1.20	1.10	27

with a 0.48-mm length of fiber spacer and to a 0.3-mm diameter for the ball lens with a 0.62-mm length of fiber spacer.

To further examine the beam quality of the fiber-lens system, beam profile images and normalized intensity distributions with Gaussian-fitted results in the x - and y -directions on the three typical planes (i.e., beginning plane, center plane, and end plane) were measured. As a sample, Fig. 3 shows measured results of intensity distribution and beam profile of a GRIN fiber lens. The measured beam profiles match very well the theoretical Gaussian distributions at the center plane. On the beginning plane and end planes, the measured and Gaussian-fitted

Fig. 3. Measured and Gaussian-fitted $1/e^2$ intensity beam diameters along the axial distance (zero is the position of the lens surface) at x (horizontal) and y (vertical) radial coordination in the distance range of the depth of field of samples 1–4, which were made from GRIN fiber lenses. Insets: (left top) Measured beam profile image of sample 4. (Bottom) Schematic diagram of the lens structure.

intensity distributions also generally match well, despite slight deviations on both ends of the distributions, which lead to some discrepancies between the measured and Gaussian-fitted beam diameters. In addition, the circular shapes in the beam profile images, as they appear in Fig. 3, indicate high x and y symmetries of the beam profiles through all ranges of the depth of field.

For imaging arterial tissue, a fiber ball lens was designed and fabricated in-house with a ball size of 0.3 mm, with a working distance of 1.25 mm, a depth of field of 1.0 mm, and $1/e^2$ spot diameter of $29 \mu\text{m}$ (shown in the inset of Fig. 4(a) as a forward-view probe). To form a side-viewing fiber probe, the output beam can be total internal reflected 90° – 100° by a 45° – 0° polished face on the fiber ball. A needle-delivered fiber catheter probe designed for OCT intravascular imaging is shown in Fig. 4(b). The polished lens and the uncoated portion of the SMF are protected in a transparent inner catheter (OD 0.49 mm) shown in the inset of Fig. 4(b). The buffered portion of the fiber is attached to an outer flexible catheter (OD 1.4 mm) after a syringe, which is fastened onto a modified syringe piston (not shown here), whereas the transparent inner catheter is inserted into a 21 G (OD 0.81 mm) echogenic spinal needle (VWR, Mississauga, ON, Canada). After insertion into the tissue, the needle can be drawn back a small distance to expose the optical probe to the tissue, as shown in Fig. 4(b). The probe is then scanned axially inside the tissue driven by a linear scanner, such that a 2-D OCT image is formed. For the OCT system with an optical amplifier after the sample, a very low surface reflection is required to reduce the beating noise. This ball lens probe fabricated in our laboratory has an extremely low reflectivity (~ -55 dB), which is suitable for a signal-amplified OCT system.

III. THEORETICAL ANALYSIS

SNR analysis of our quadrature interferometric platform with the SOA is performed based on the SNR analysis of a conventional balanced SS-OCT system [21], with the extra noise power from the amplifier [16] added, followed by a correction of the power distribution in the quadrature interferometer.

Fig. 4. (a) Opened left descending coronary tissue from a WHHLMI rabbit being scanned with the forward ball lens fiber probe protected by a plastic tube. Inset: Scanning electron micrograph of the fiber ball lens with forward-view fabricated in our laboratory. (b) Needle tip of a side-view fiber ball lens probe as a sample of an OCT probe for *in vivo* intravascular imaging.

Assuming that the signal in the sample arm is coming from a single-layer reflector located at the depth of z_0 , then the two channel currents on the positive and negative photodiodes in each balanced detector when the SOA is inserted into the sample arm are given by

$$I_{1\pm}(k_m, G) = \alpha \left[P_{r\pm}(k_m) + P_{s\pm}(k_m, G) + P_{SP\pm}(G) \pm 2\sqrt{P_{r\pm}(k_m)P_{s\pm}(k_m, G)} \times \cos(2k_m z_0 + \phi) \right] \quad (1)$$

$$I_{2\pm}(k_m, G) = \alpha \left[P_{r\pm}(k_m) + P_{s\pm}(k_m, G) + P_{SP\pm}(G) \pm 2\sqrt{P_{r\pm}(k_m)P_{s\pm}(k_m, G)} \times \cos(2k_m z_0 + \phi + \Delta\phi) \right] \quad (2)$$

where $\alpha = \eta e / (h\nu_0)$ is the photodiode conversion factor; η is the quantum efficiency of the detector; e is the electronic charge; h is the Planck's constant; ν_0 is the mean frequency of the incident light; $P_r(k_m)$ is the optical signal power from the reference arm arriving at the photodiodes when wavenumber $k = k_m$; $P_s(k_m, G)$ is the amplified optical signal power from the sample arm at the photodiodes when wavenumber $k = k_m$; $P_{SP}(G)$ is the spontaneous emission power of the SOA recorded at the photodiodes; G is the gain of SOA;

ϕ is an arbitrary phase shift; $\Delta\phi$ is the phase shift between the two output signals; and $m = 1, 2, \dots, M$, where M is the total sampling number of the axial pixels. The balanced detection method subtracts each positive and negative current for each channel shown in (1) and (2), so that the common dc parts are canceled and the opposite ac parts are multiplied. The complementary phase components of the signals can be calculated from a complex discrete Fourier transform. Then, the maximum-squared signal power in our system with the dual-balanced quadrature detection is given as

$$\langle I(G)^2 \rangle = 8D^2 \alpha^2 P_r P_s(G) \quad (3)$$

where the brackets denote the ensemble average, P_r is the optical power reflected from the reference mirror and impinging on each photodiode, and $P_s(G)$ is the amplified signal power incident on each photodiode backscattered from the sample. D is a multiplication factor generated by the complex IFT, with $D = 2$ in our setup.

In this configuration, there are three typical noise sources: 1) thermal noise; 2) shot noise; and 3) beat noise. By extending the noise analysis of SS-OCT [21], [22] with balanced detection [23] to our QSS-OCT system, with the SOA inserted into on the sample arm, the total noise power as a function of the SOA gain G is obtained as

$$\langle \sigma(G)^2 \rangle = DB \left\{ i_{th}^2 + 4e\alpha (P_r + P_{SP}(G)) + 2\alpha^2 \times [2\beta(2 - \beta)RIN_b P_r P_{SP}(G) + (1 - \beta)^2 \times (RIN_r P_r^2 + RIN_s P_{SP}(G)^2)] \right\} \quad (4)$$

where B is the detector bandwidth; i_{th} is the thermal current of the detector; β is a balanced factor [21], with $\beta = 1$ representing a balanced system; $RIN_{r/s} = 2/\delta\nu_{r/s}$ is the relative intensity noise of the source and the SOA; and $\delta\nu$ is their effective bandwidth. RIN_b is the relative cross-beat intensity noise from the reference light and the spontaneous emission power of the SOA, which can be estimated from the experimental results. The first and second terms in (4) are thermal noise and shot noise, as commonly expressed. The third term represents the beat noise, which includes the cross-beat and self-beat noise of the two arms. If the losses and reflectivity coefficients of the reference and sample arms are defined as γ_r , R_r , and γ_s , R_s , respectively, then the powers from the reference and sample arms impinging on the photodiodes can be described as $P_o\gamma_r A R_r$ and $P_o\gamma_s G R_s$, where A is the reference attenuation when the sample signal is not being amplified. If the loss from the SOA to the detector is defined as γ , the spontaneous emission power of the SOA impinging on the photodiodes can be written as γP_{SP} . The SNR of our QSS-OCT system with the SOA inserted for weak sample signal amplification is described as (5), shown at the bottom of the page.

$$SNR(G) = \frac{16D\alpha^2 P_0^2 \gamma_r A^2 R_r \gamma_s G R_s}{B \left\{ i_{det}^2 + 4e\alpha (P_0 \gamma_r A^2 R_r + \gamma P_{SP}(G)) + 2\alpha^2 [2\beta(2 - \beta)RIN_b P_0 \gamma_r A^2 R_r \gamma P_{SP}(G) + (1 - \beta)^2 (RIN_r (P_0 \gamma_r A^2 R_r)^2 + RIN_s \gamma^2 P_{SP}(G)^2)] \right\}} \quad (5)$$

Fig. 5. Theoretical analysis results of (left vertical axis) various noise powers and (right vertical axis) SNR versus (right horizontal axis) SOA gain G when the QSS-OCT system contains a sample signal amplifier and versus (left horizontal axis) reference arm attenuation A when the QSS-OCT system without a sample signal amplifier.

Fig. 5 shows the calculated results for various noise powers (left vertical axis) and SNR (right vertical axis) versus SOA gain G when the QSS-OCT system contains a sample signal amplifier (right horizontal axis) and versus reference arm attenuation A when the QSS-OCT system is without a sample signal amplifier (left horizontal axis). The calculations are based on the following assumptions: $\lambda_0 = 1.32 \mu\text{m}$, $\delta\lambda_r = 1 \text{ nm}$, $P_0 = 12 \text{ mW}$, $\gamma_r = -20 \text{ dB}$, $\gamma_s = -10.5 \text{ dB}$, $\gamma = -9 \text{ dB}$, $R_r = 0 \text{ dB}$, $R_s = -55 \text{ dB}$, $B = 50 \text{ MHz}$, $\beta = 0.99$, $i_{th} = 2 \text{ nA}/\sqrt{\text{Hz}}$, $\delta\lambda_s = 100 \text{ nm}$, $P_{SP} = P_{sp0}G$, and $P_{sp0} = 10^{-3} \text{ mW}$. From the theoretical analysis results shown in Fig. 5, when the QSS-OCT system is without the SOA, reference light power could be attenuated to reduce the reference power self-beat noise and to obtain the shot-noise limit. When the QSS-OCT system contains the SOA in the sample arm, then SNR can be increased as the gain of SOA increases, although the system is no longer at the shot-noise limit. When the gain of sample arm SOA is low, i.e., $< 15 \text{ dB}$, reference (ref.) self-beat noise dominates the cross-beat noise. The SNR linearly increases as the gain is increased, because the ref. self-beat noise stays constant as the gain increases. Because the cross-beat noise from the reference power and the spontaneous power emission of the SOA increases as the gain increases, when $G > 15 \text{ dB}$, the cross-beat noise raises to the level of the ref. self-beat noise, slowing the increase of SNR. As the gain increases to $G > 30 \text{ dB}$, SNR will saturate where the cross-beat noise becomes dominant. Increase of SNR up to 15 dB is calculated when the QSS-OCT system contains the optical amplifier, in comparison with the system without the optical amplifier in the shot noise limit, as shown in Fig. 5.

IV. RESULTS AND DISCUSSIONS

In optical coherence tomography, the formation of an image is based on the detection of the interference of low-coherence light that has been backscattered from the sample. *In vivo* and *ex vivo* OCT images of tissues were acquired using our catheter-based SS-OCT system. The quality of OCT images acquired with the system containing the SOA is compared with the quality of those obtained without the SOA.

Fig. 6 shows *in vivo* images of the tip of a human finger taken with the OCT system (a) with and (b) without the SOA

Fig. 6. *In vivo* images of a tip of human finger taken with the proposed OCT system (a) with SOA and (b) without SOA. The dimension bars correspond to 1 mm. Also shown are A-scan signals (black) with SOA and (purple) without SOA at the positions of the dashed lines.

in the sample arm. The gain of SOA is set at 30 dB. The image size is 600×900 pixels, respectively, which corresponds to $2.5 \times 3.5 \text{ mm}^2$. The image size was calibrated using a 1-mm glass slide and an assumed refractive index of 1.3 in the tissue. The dimension bars correspond to a length of 1 mm. The axial and lateral resolutions are 12 and $29 \mu\text{m}$, respectively. Sweat glands and other morphological features are better visualized at the epidermis/dermis junction (shown as *), with the SS-OCT system having the weak signal amplified by the SOA, as shown in Fig. 6(a), than with the system without the amplifier, as shown in Fig. 6(b). The increase in image quality at depth when acquired with the QSS-OCT using the SOA is evident in Fig. 6(a) from the ability to distinguish the deeper subcutaneous layer (shown as arrow) in the human finger. However, structures located deeper than 1 mm become difficult to discern when the SOA is removed from the system [see Fig. 6(b)].

The advantages of the higher sensitivity and the deeper imaging capability of the proposed OCT system with the SOA are visible in Fig. 6(c). Two A-scan signals at the same position, which are indicated by the dashed line/arrows, are shown in

(a)

(b)

(c)

Fig. 7. *Ex vivo* images of healthy arterial tissues of (a) and (b) mice with closed luminal channel and (c) pig with opened (tissue cut along the flow of blood) left descending coronary. Images were acquired with the proposed SS-OCT using (a) a side-view needle probe along the blood flow and (b) and (c) forward-view probe transversal on the blood flow. The bar represents 1 mm.

Fig. 6(c) for the OCT system with the SOA (Black) and the system without the SOA (purple). A clear increase of the SNR is observable for the system with the SOA, where the signal increases up to 30 dB while there is less than 10-dB noise increase, compared with the system without the SOA. Signals coming from structures located at depths of up to 2 mm can be observed in the data acquired with the system that has the SOA inserted [see Fig. 6(c)].

Cardiovascular disease is the leading cause of death in the most developed countries. Coronary atherosclerosis, which is associated with thrombosis, embolism, or cardiac infarction, is a major contributor to cardiovascular morbidity and mortality. *In vivo* assessment of coronary vessels is becoming an important tool for clinical investigation. Recently, in May 2010, OCT was approved by the U.S. Food and Drug Administration for cardiovascular applications.

Various arterial tissues were imaged with our catheter-based SS-OCT with SOA, and examples are shown in Fig. 7. Fig. 7(a)

Fig. 8. *Ex vivo* OCT images of opened aorta from a WHHLMI rabbit acquired by our catheter-based SS-OCT system (MA: macrophage accumulation, FC: fibrous cap, CA: calcified regions).

shows *ex vivo* OCT images of a healthy unopened arterial segment from a mouse, where images were acquired using a 0.4-mm catheter probe inserted along the lumen by the needle mechanism previously described and shown in Fig. 4(b). The side-view probe was linearly scanned along the direction of blood flow. The top layer shown in Fig. 7(a) represents the thickness of the surface of the probe (PS). After that, three consecutive layers, i.e., the tunica intima (TI), tunica media (TM), and tunica adventitia (TA) of the artery, are clearly visualized. The surrounding fatty tissue (FT) layer is viewed as well. An OCT image of the same arterial tissue where the catheter-based SS-OCT was used in conjunction with the forward-view probe being scanned transversally outside of the artery is shown in Fig. 7(b). More details of the muscle layers and fat tissue can be seen from this image, which was scanned transversely to the direction of blood flow. The circular orientation of the smooth muscle fibers that composed the TM is revealed in more detail after the artery is cut open (along the direction of blood flow), and the OCT image is acquired by having the luminal side exposed to the probing beam and where the scan is perpendicular to the direction of blood flow. Fig. 7(c) is an example of such an OCT image of an opened left descending coronary artery harvested from a pig.

Fig. 8 shows an *ex vivo* OCT image of a segment of aorta. In Fig. 8, a clear raised lipid core with macrophage accumulation (MA) and nonuniform reflectance coefficients within its volume are demonstrated. A thin fibrous cap (FC), which strongly scatters light, covers the lipid core. The fibrous cap thickness was defined as the minimum distance from the lumen of the vessel to the upper border of the lipid pool. A few calcified regions (CA) around the lipid core, which characterized itself by low reflectance, were clearly discernable. The lipid-rich region with nonuniform reflectance can be viewed up to depths of 2 mm, even beneath the calcified regions. Such details such as the narrowing fibrous cap, the macrophage accumulation, and the calcification boundary are also evident in this image. The identifiable important features such as the thickness of the fibrous cap and the size of the lipid pool can be quantified from these OCT images.

V. CONCLUSION

We have demonstrated high-quality OCT images of various tissues such as human skin, arteries of mice, and atherosclerotic arterial plaque of rabbit, using our catheter-based SS-OCT

system with a quadrature interferometer and a SOA. Image quality at locations deeper within the tissue structures has been improved due to an increase of SNR afforded by the presence of a SOA in the sample arm. These images have offered the possibility to investigate tissue microstructure and reveal morphological details located well within the arterial wall or into the dermis of the skin. Results have shown that the proposed swept source OCT is an effective imaging modality for real-time diagnosis with great potential for *in vivo* applications. For cardiovascular clinical conditions, integrating a SOA with the catheter-based SS-OCT system can provide superior capabilities, allowing the method to be used for scanning the lumen area; assessing lesion cap thickness; detecting calcium, lipid, thrombus, and macrophages; and even assisting in intervention.

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