Arrayed Waveguide Grating for Polarized Raman Spectroscopy of Human Teeth

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Abstract— **We designed an arrayed-waveguide grating spectrometer for the detection of early dental caries in teeth through polarized Raman spectroscopy. Measurement results on extracted human teeth demonstrate the feasibility of the approach. (Abstract)**

Keywords-component; Raman spectroscopy; integrated optics;

I. Introduction

The use of Raman spectroscopy, and in particular polarized Raman spectroscopy (PRS) has been shown to be applicable for the detection of dental caries [1, 2] and gives complementary information to that which can be obtained using conventional techniques such as visual inspection and OCT. Furthermore, PRS can be used reliably for detecting dental caries even at an early stage [2]. In our work we aim at realizing low-cost, hand-held devices for polarized Raman spectroscopy of teeth. Here, the speed of the device is a very important aspect since both the operator who is holding the device and the subject under examination will inevitably move during the measurement. For *in-vivo* measurements the device must be able to acquire a Raman signal over the entire spectral region of interest in a matter of a few seconds [3]. We choose as the core element of our device the arrayed-waveguide grating (AWG) [3], which enables simultaneous access to all spectral components of the input signal, leading to a fast measurement irrespective of the required spectral range and resolution. The working principle of the AWG can be described as follows: light from an input channel waveguide is guided to a free-propagation region (FPR) where it diffracts in the horizontal direction and is coupled to an array of channel waveguides which are arranged on a circle of radius R equal to the length of the FPR. On this circle the arrayed waveguides are spaced by a center-to-center distance $d \ll R$. The arrayed waveguides have a linearly increasing length with step $\Delta L = m\lambda_c/n_{eff}$, where m is an integer, λ_c is the central wavelength of the AWG, and n_{eff} is the effective refractive index of the arrayed waveguides at λ_c. Light exiting from the array enters a second FPR where the output terminations of the arrayed waveguides are again arranged on a circle with radius R. The center of this circle coincides with the entrance facet of the central output channel of the AWG. In this arrangement, when light at wavelength λ_c is sent through the input channel, a circular wave front is generated at the output of the array, and

the light is focused into the central output channel. At a different wavelength ($\lambda \neq \lambda_c$) the circular wave front at the output tilts with respect to the one for λ_c and the focal spot is located at a different spatial position. Waveguides can be placed at different positions at the output of the second FPR to collect individual spectral components of the input signal. A potential disadvantage, the limited free spectral range (FSR) achievable for a given device size, can be overcome by a technology platform providing larger refractive index contrast, or by accepting spectrally folded, yet non-overlapping peaks, as discussed in [5].

II. Polarized Raman Spectroscopy of Teeth

The method relies on measuring the Raman signal from tooth enamel in the spectral region from 375 cm^{-1} to 1175 cm^{-1}. In this region the strongest band is the symmetric P-O stretching vibration of the phosphate ions $(PO_4)^{3-}$ located near 959 cm^{-1}, which arises from hydroxyapatite crystals of tooth enamel. Information on the state of the disease is provided by the ratio of the intensities of two orthogonal polarization states (perpendicular and parallel to the polarization of the excitation laser) of the 959 cm^{-1} hydroxyapatite peak. Their intensity ratio is referred to as the depolarization ratio, $\rho_{959} = I_{959}(\perp) / I_{959}(\parallel)$. In sound enamel the hydroxyapatite signal is highly polarized in the direction parallel to that of the excitation laser, and the value of the depolarization ratio is approximately $\rho_{959} = 0.10 \pm 0.04$ [2]. As the carious lesion develops, the depolarization ratio increases to ~0.4 ± 0.12 due to a structural modification of the tooth enamel [2].

III. AWG Design and Characterization

The device is designed for an excitation wavelength of 830 nm because it induces less fluorescence background in tissue than shorter wavelengths. With this excitation wavelength the hydroxyapatite peak is positioned at ~901 nm. To resolve the peak the AWG is designed to have a central wavelength of 901 nm, a FSR of ~22 nm between 890 and 912 nm, and a resolution of 0.2 nm. The AWG is fabricated using single-mode silicon oxynitride [6] channel waveguides with with 2.2 μm width and 0.52 μm height. The waveguide core and cladding refractive indices are 1.509 and 1.445 at 830 nm, respectively. The AWG has a working order of $m = 41$ to avoid

overlap (at higher imaging orders) between the excitation wavelength and the hydroxyapatite peak. In Fig. 1 we show the measured spectral response of the central AWG channels compared to the simulation results. The characterized device has a central wavelength of 900.78 nm (for TE polarization), a resolution of 0.2 nm and a FSR of ~22 nm (as designed). Furthermore, the characterized AWG exhibits a polarization-dependent response in which the wavelength shift between the TE and the TM polarizations is 0.5 nm.
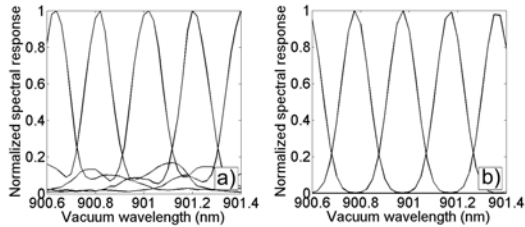

Figure 1.   Normalized spectral response of the central channels of the AWG for the transverse-electric (TE) polarization. (a) measured; (b) simulated.

IV.   EXPERIMENTAL RESULTS

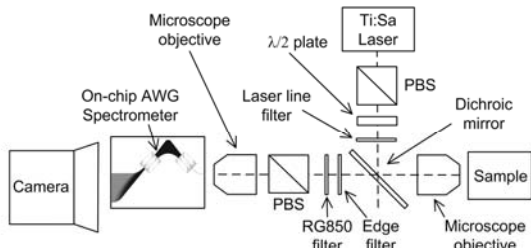The setup used to perform polarized Raman measurements with the AWG is shown in Fig. 2.

Figure 2.   Schematic diagram of the setup used for the polarized Raman experiments.

Light from a linearly polarized Ti:Sapphire laser at 830 nm is sent through a polarization beam splitter (PBS) oriented parallel to the laser polarization. A half-wave plate, initially with one of its optical axes parallel to the laser polarization, is placed after the PBS. The laser light exiting the λ/2 plate, after passing through a laser-line filter (Semrock LL01-830-12.5), is reflected from a dichroic mirror (Semrock LPD01-830RS-25) and focused onto the sample (extracted human tooth) with a ×40 microscope objective having numerical aperture NA = 0.65. The light backscattered from the sample is then collected by the same optics. The Rayleigh-scattered component is again reflected (rejected) by the dichroic mirror while the Raman-scattered wavelengths are transmitted. An edge filter (Semrock LP02-830RS-25) and an additional red-glass filter (RG 850) are used to further suppress residual reflected and Rayleigh-scattered light at the laser wavelength. At this point we position a second PBS with the same orientation as the first, and a ×50 (NA = 0.85) microscope objective to focused the light onto the input channel of the AWG spectrometer. The spectroscopic measurement is performed by imaging the output channels of the AWG onto an electron-multiplying charge-coupled device (Andor iXonEM DV887ECS-BV, EMCCD) through a camera lens (JML f0.95 50 mm). In order to perform the measurements for perpendicular polarization, the λ/2 plate is rotated by 45°, to rotate the laser polarization by 90°. The measurements were performed on extracted human tooth samples presenting clinical white spot lesions as early dental caries. We used 200 mW of incident power on the tooth surface. The Raman signal was strong and clearly visible with integration times as short as 10 s. In Fig. 3 we show representative spectra measured with 80 s integration time and 10 accumulations.
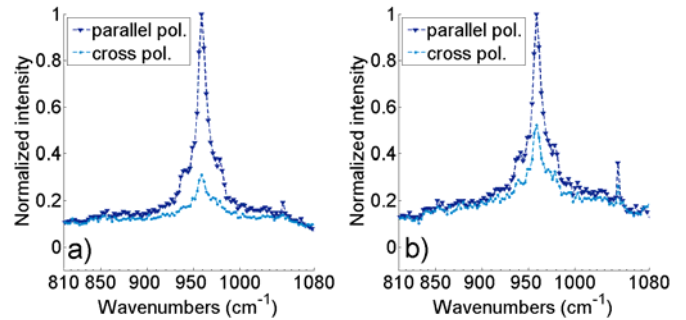

Figure 3.   Raman spectra of (a) sound and (b) carious tooth enamel for both, parallel (triangles) and cross (dots) polarizations acquired with the AWG.

We calculate depolarization ratios of $\rho_{959} = 0.16 \pm 0.047$ for the sound regions and $\rho_{959} = 0.41 \pm 0.01$ for the carious regions, which are close to the values reported in the literature [2].

V.   CONCLUSIONS

We have designed and fabricated an AWG spectrometer to be used in polarized Raman spectroscopy of human teeth. Our measurement results, which are comparable with the ones found in the literature for both, sound and carious regions of tooth enamel, demonstrate the feasibility of our approach.

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