Integrated optical confocal system for Raman spectroscopy

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Abstract. *We have designed a polarization insensitive arrayed-waveguide grating as a wavelength-selective device for Raman spectroscopy of the skin. The integrated spectrometer was characterized. Experimental results are presented and compared with a simulation. We tested our device in a novel confocal arrangement with a similar device that was used for focusing the excitation signal onto the sample. Experimental results on the collection efficiency and volume are presented together with a demonstration of multi-wavelength imaging.*

Introduction

Our challenge is to realize an integrated, low-cost, compact, hand-held spectrometer for *in vivo* confocal Raman spectroscopy of the skin, in particular to detect water concentration in the *stratum corneum*. Water concentration can be measured in terms of ratios between the detected signals in the water, lipid, and protein bands [1]. The Raman spectrum of the skin is shown in Fig. 1 for different values of water concentration, together with the required bands.

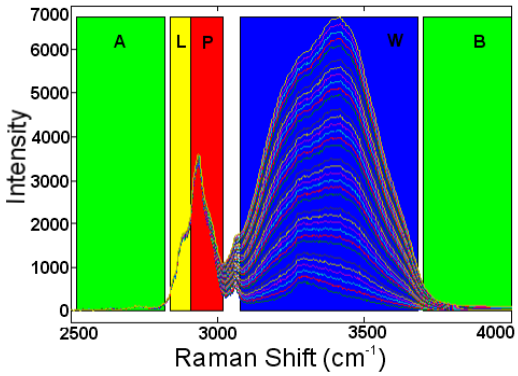

Fig. 1. Raman spectrum of the skin for different values of water concentration, with bands A and B used to correct for background variations, L is the lipid band, P the protein band and W the water band.

The lipid peak (region indicated with the letter L in Fig. 1) is the narrowest and weakest peak in the skin's spectrum, and therefore the integrated device must present the lowest losses in this wavelength region. The free spectral range FSR of the device must also be large enough to cover the entire skin spectrum from 2500 cm^{-1} to 4000 cm^{-1}, and it must be polarization independent in this wavelength range in order to be able to detect indistinctively both TE and TM polarizations of the Raman signal.

As we aimed for compactness and low cost, we investigated a novel planar confocal optical system for delivering excitation light and collecting the Raman signal, that can be integrated with the spectrometer. Both the spectrometer and the confocal system are based on arrayed waveguide gratings.

Arrayed waveguide spectrometer design

We have designed an arrayed waveguide grating (AWG) spectrometer on a thermally oxidized silicon wafer using low-birefringence silicon oxynitride (SiON) waveguides with SiO_2 cladding. The waveguides have a cross-section of 1.4 μm × 0.52 μm and a core refractive index for TE polarization of 1.509 at 830 nm. The material birefringence is $\Delta n_{\text{TM-TE}} = 2.1 \times 10^{-3}$ for SiON and 1.0×10^{-3} for SiO_2 [2]. As verified experimentally, the guiding material does not generate luminescence in the Raman wavelength range when excited with 671-nm laser light, which is the wavelength chosen for the excitation in the skin application. The requirements for wavelength selection are shown schematically in Fig. 1, where**Errore. L'origine riferimento non è stata trovata.** the Raman shift is indicated in units of cm^{-1}, which for our excitation wavelength translates into a wavelength range from 800 nm to 920 nm. The resolution of the device must be large enough to separate the lipid and protein peaks; taking into account that the lipid band is 5 nm wide and that in this wavelength region the Raman signal is weaker than in the P and W bands, we choose the central wavelength of the AWG at the center of the lipid band (832 nm) and an output channel spacing of 5 nm. In this way we guarantee the lowest insertion loss for the lipid peak.

To cover the entire Raman signal, a minimum free spectral range of 120 nm is required; an optimal choice of grating order is $m \leq 3$ (we chose 3), since higher values of m would result in the (backscattered) laser wavelength overlapping with or falling close to one of the Raman channels. The requirement of a low order, $m = 3$, makes it necessary to use a broadband s-shaped AWG layout, because for this low order the conventional horse-shoe geometry does not permit the interconnection of the arrayed waveguides to the free propagation regions with the correct angular relations.

The polarization insensitivity requires the use of low-birefringence waveguides [3]. In our particular case the waveguide geometry has been designed to have zero birefringence in the center of the Raman signal range and $\pm 1 \times 10^{-4}$ at the edges of the spectrum. Simulations of the AWG with this waveguide geometry show negligible shifts of 0.01 nm between the TE and TM peaks for the central channels, and 0.03 nm in the W band. From the simulations we also observe low losses ranging from 1.1 dB for the central channel up to a maximum of 2 dB for the outermost channels, and a non-adjacent crosstalk of -37 dB for the central channels which increases up to -27 dB at higher wavelengths.

AWG characterization

The fabricated device was characterized for both TE and TM polarizations. For this purpose a super-continuum (Fianium) source was used as an input, and the output spectrum was measured from 7 central and 7 outer channels for the two polarizations using a spectrometer (Horiba iHR550). The measured spectra, shown in Fig. 2, were normalized with respect to a reference spectrum from a straight waveguide channel.

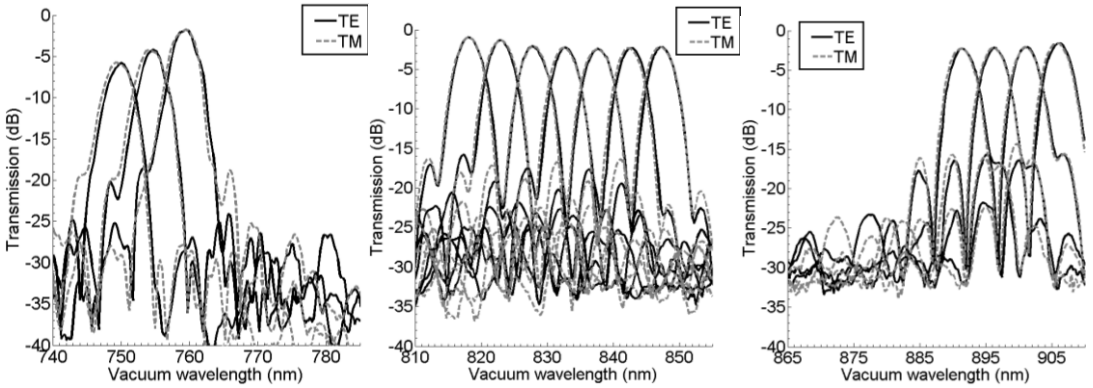

Fig. 2. Response of AWG in the wavelength range from 740 nm to 910 nm, for both TE and TM polarizations: (ii) central channels; (i,iii) outer channels.

From the measurements (with spectral resolution of 0.25 nm) we observe that the device intrinsic losses (excluding input and output fiber-to-chip coupling losses) are approximately 1.2 dB for the central channel and increase to around 2.5 dB for the outermost channels; while the shift between TE and TM peaks is less than 0.5 nm in the measured range from 740 nm to 910 nm.

Integrated confocal light delivery and signal collection

To combine the functions of light focusing and signal collection on the same chip we made use of two AWGs in a confocal arrangement displayed in Fig. 3.i. In this way the output free propagation region (FPR) of the focusing AWG and the input FPR of the collecting AWG merge together, forming a single sample-side FPR (SFPR). In its simplest design both AWGs have order $m = 0$ such that they form two lenses, the first one focusing a single excitation beam into the focal spot at a desired depth inside the sample and the second one collecting the light backscattered from this focal spot into a central output channel. In this arrangement [4, 5] the device behaves like a confocal microscope (in the lateral direction). Wavelength selectivity can be achieved without increasing device complexity by using AWGs with order $m > 0$. As a result, the focusing AWG can have more than one input channel, allowing one to focus several excitation wavelengths into the same focal spot. Likewise, the collecting AWG can resolve different wavelengths emitted from the focal spot, thereby enabling spectral analysis of the backscattered light.

Multi-wavelength imaging was demonstrated utilizing a device consisting of an excitation AWG of order zero in confocal arrangement with a collector AWG having 11 output channels spaced by 5 nm and centred at 831 nm. As a sample we used the facet of a 9-μm-fiber array with 250-μm spacing, partially coated with silver paint, as shown in Fig. 3.ii (a). Two lasers operating at 831.4 nm and 838 nm were coupled into adjacent fibers of the array to mimic multi-wavelength emission from the sample. For testing the confocal arrangement additional light from a super-continuum source was passed through a red-glass (RG850) filter, focused through the excitation AWG, and reflected by the silver paint. The sample was positioned at the focal spot and scanned over an area of 620 μm × 60 μm with 1 μm step size. By detecting the signals from three output channels of the collector AWG centred at 831 nm, 841 nm, and 856 nm, respectively, three images were formed (Figs. 3.ii (b-d)). These images were then combined into a single image presenting both spatial and spectral information (Fig. 3.ii (e)).

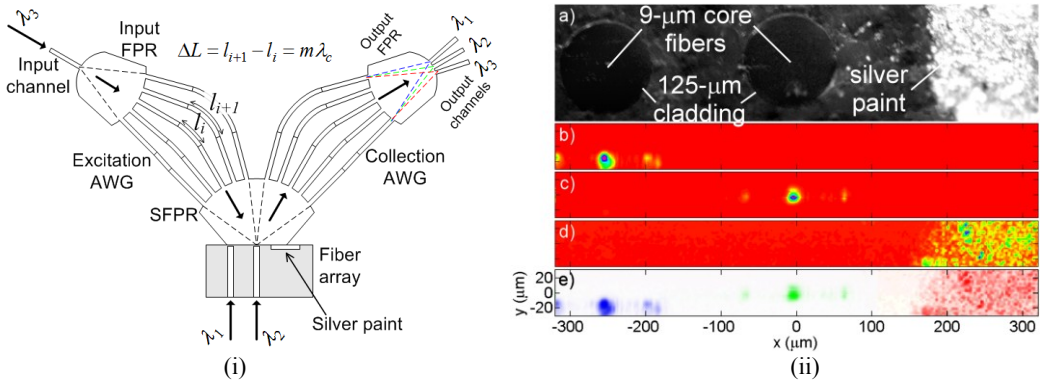

Fig. 3. Confocal arrangement of two arrayed waveguide gratings (i); (ii-a) 9 μm fiber array used as a sample;(ii-b,c,d) measurement from output channels centered at 831,841, and 856 nm; (ii-e) resulting multiwavelength image.

Conclusions

We have discussed the design of an integrated device for Raman spectroscopy of the skin working in the near infrared spectral region and presented experimental results on the characterization of the device. The device is polarization insensitive in a broad spectral range (740 nm – 910 nm) in which it presents low losses. Furthermore, we proposed a novel method for laser delivery and confocal signal detection from a confined region below the surface of a sample which may open possibilities for on-chip spectroscopic measurements without the need for external optics.

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