Plastic Distributed Feedback Laser Biosensor

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Introduction

A wide variety of optical resonator structures have been used for label-free detection of chemical compounds, biomolecules, and cells [1], and several approaches have been developed commercially for application in life science research, environmental monitoring, quality control testing, and diagnostic testing [2,3]. Label-free resonant optical sensors generally detect shifts in resonant wavelength or coupling angle caused by the interaction between the target molecule and the evanescent portion of resonant modes. The narrow spectral linewidth achieved by using high Q factor ($>10^5$) passive optical resonators, enables sensor systems to resolve smaller wavelength shifts associated with the detection of low concentration analytes or low molecular weight biomolecules. While detection resolution can be substantially improved through the use of high Q factor passive resonators, the sensitivity and dynamic range of the system is generally decreased and implementation of high Q factor optical resonators typically require high precision alignment for evanescent light in/out coupling, providing potential limits their practical application in high throughput biomolecule screening.

Active resonator sensors, such as the distributed feedback laser (DFB) biosensor demonstrated in this work, have been drawing special interest because they generate their own narrow linewidth stimulated emission, while retaining simple light coupling instrumentation [4,5]. Altering the refractive index (RI) of the media exposed to the DFB laser surface or surface adsorption of biomolecules changes the effective RI associated with the resonant mode, and results in modulation of the stimulated emission wavelegnth. The DFB laser sensor biosensor was fabricated with organic material on a flexible plastic substrate using a large surface-area nanoreplica molding process.

Design and Fabrication

The DFB cavity is based on a second order Bragg grating that supports a vertically emitting mode by first-order diffraction [6]. A schematic cross-sectional diagram of the designed DFB laser structure is shown in Fig. 1. The one-dimensional grating structure is produced with a low RI ultraviolet curable polymer on polyethylene-terephthalate substrate by a nano-replica molding technique [3]. The replicated gratings have a period, depth, and RI of 400 nm, 40 nm, and 1.38, respectively. The low RI polymer layer also functions as a cladding layer, upon which a laser dye doped thin film of high RI polymer provides light confinement, feedback, and amplification. In this work, Rhodamine 590 doped SU-8 was spin coated on grating surface. The thickness of spun-on polymer is carefully controlled to enable the DFB laser to operate with a single radiation mode which facilitates the determination of laser emission wavelength shift. An optional titanium dioxide thin film was deposited on top of the DFB laser surface to improve biomolecular immobilization.

Sensor Characterization

The DFB laser was optically excited by a frequency doubled Nd:YAG laser (λ=532 nm) through a 600 μm diameter fiber and a focusing lens underneath the sensor surface. The emission from the DFB laser biosensor was coupled to a spectrometer through a detection fiber bundled with the excitation fiber. Fig. 2 shows the laser emission spectrum observed with the sensor surface exposed to air while pumped at 8.5 μJ-mm^{-2} and the spectrum was fitted to a Lorentzian profile to mathematically determine the center wavelength. The DFB laser exhibits a threshold of 1.09 μJ-mm^{-2} (inset of Fig. 2). Sensitivity to changes in the RI of media exposed to the sensor surface was measured by placing a droplet of water (RI=1.33), acetone (RI=1.36), isopropyl alcohol (RI=1.38) and dimethyl sulfoxide (RI=1.48) upon a single sensor in sequence. The recorded laser wavelengths were plotted as a function of liquid RI in Fig. 3 and a bulk RI sensitivity of $S_b=\Delta\lambda/\Delta n$=99.58 nm/RIU was measured, with linear behavior over the ~14 nm tuning range.

In order to characterize the sensor sensitivity as a function of distance from the sensor surface, stacked alternating positively and negatively charged polyelectrolyte layers were

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deposited onto the sensor surface. The used polyelectrolytes were anionic poly(sodium 4-styrenesulfonate) (PSS; M_w = 60 kDa), cationic poly(allylamine hydrochloride) (PAH; M_w = 70 kDa) and cationic poly(ethylenimine) (PEI; M_w = 60 kDa) all dissolved in 0.9 M NaCl at a concentration of 5 mg/ml. The polyelectrolyte layer coating self-limits to a single monolayer with a RI of ~1.49 and thickness of ~4.4 nm [3]. One PEI and six PSS-PAH alternating layers were deposited in sequence with a NaCl buffer rinse used after every PSS or PAH incubation. Figure 4 illustrates the temporal progression of the laser wavelength shift for the PEI and six PSS-PAH depositions. The sensor wavelength shift response is not saturated after the deposition of ~52 nm material on its surface. Results demonstrating the sensor ability to detect biomolecules and to characterize the affinity binding constant of a protein-protein interaction will be presented.

Reference

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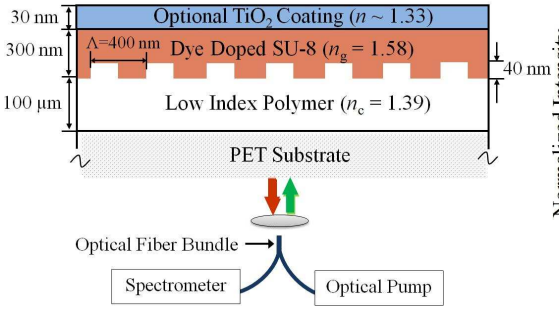

Fig 1. Cross-sectional schematic diagram of the plastic-based DFB laser biosensor.

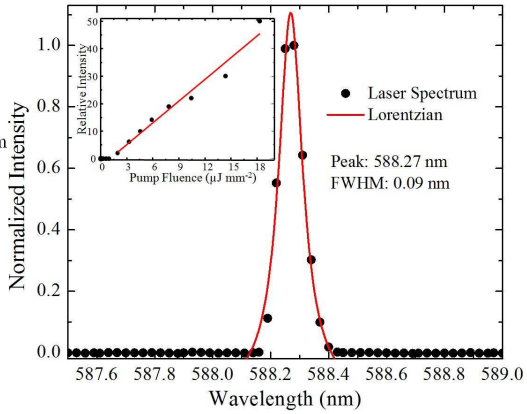

Fig 2. DFB laser emission with the sensor surface exposed to air and pumped at a fluence of 8.5 μJ mm^{-2}. Inset shows the laser threshold curve.

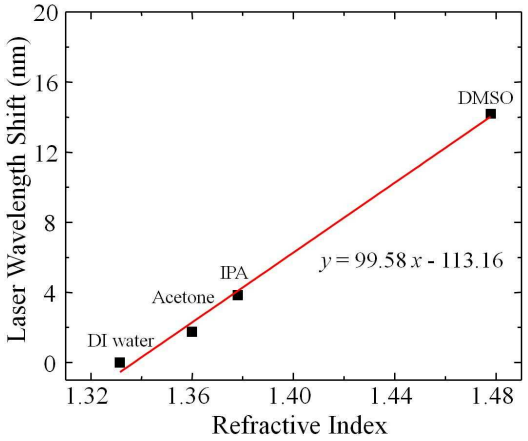

Fig. 3. Dependence of laser emission wavelength shift as a function of various solvents applied to sensor surface.

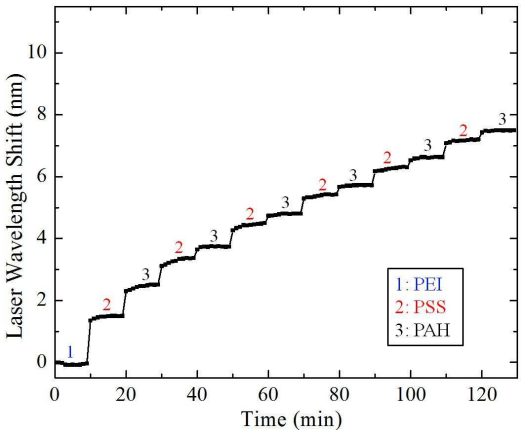

Fig 4. Dynamic detection of alternating layers of positive and negative charged polymer self-limiting monolayers.