Enhancing the Angular Sensitivity of Plasmonic Sensors Using Hyperbolic Metamaterials

Kandammathe Valiyaveedu Sreekanth,* Yunus Alapan, Mohamed ElKabbash, Amy M. Wen, Efe Ilker, Michael Hinczewski, Umut A. Gurkan, Nicole F. Steinmetz, and Giuseppe Strangi*

1. Introduction

Surface plasmon polariton (SPP) based sensors have emerged as a promising technology for the development of next-generation biological and chemical sensors for high-throughput, label-free, and multianalyte sensing applications.[1–5] SPPs are collective charge density oscillations occurring at the interface between a metal and a dielectric that have been widely used for the investigation of change in refractive index near the metal–dielectric interface. This is because the electric field associated with these oscillations is very intense at this interface and decays exponentially in both interfacing media. Based on the momentum matching condition required for the excitation of SPP on a thin metal film, the SPP sensors are clas-

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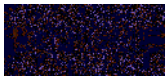

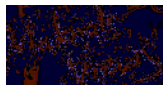

coupling.[6–9] In both con gurations, two major interrogation schemes have been employed for acquiring the information by measuring the re ected signal: spectral scan and angular scan interrogation.[9–12] In spectral interrogation, the resonant wave-

was patterned on top of the spacer layer using electron-beam lithography. Using a two-layer photoresist coating, deep sub-wavelength gratings were realized. Then, a thin gold layer of 20 nm thickness was directly deposited on top of the sample using thermal evaporation to realize a gold diffraction grating.

The metal diffraction grating was chosen to be gold because of the easier functionalization processes with biomolecules by using thiol-based surface chemistry.[33] Figure 1b shows a scanning electron microscope (SEM) image of the 2D gold diffraction grating on the HMM with an average period and hole diameter of 500 and 160 nm, respectively.

We then used effective medium theory (EMT) to determine the uniaxial dielectric tensor components of the fabricated HMM,[25] $\varepsilon_{ll} = \varepsilon_x = \frac{t_m\varepsilon_m + t_d\varepsilon_d}{t_m + t_d}$ and $\varepsilon_{\perp} = \varepsilon_z = \frac{\varepsilon_m\varepsilon_d(t_m + t_d)}{t_m\varepsilon_d + t_d\varepsilon_m}$, where ($\varepsilon_d$, t_d) and (ε_m, t_m) are the dielectric permittivity and thickness of Al_2O_3 and gold, respectively. EMT-derived permittivity components (Figure S1, Supporting Information) of the fabricated gold-Al_2O_3 multilayer con rmed a Type II HMM with hyperbolic dispersion at $\lambda \geq$ 520 nm (i.e., $\varepsilon_{ll} < 0$ and $\varepsilon_{\perp} > 0$). Re ectance spectra at oblique incidence were obtained using variable angle high-resolution spectroscopic ellipsometry. The re ectance spectrum of the GC-HMM obtained using p-polarized light is shown in Figure 1c. In the hyperbolic region ($\lambda >$ 520 nm), four re ectance dips with high quality factor resonance were obtained, which represents the highly con ned bulk plasmon polaritons.[30,31] These modes can be experimentally probed by studying the re ectance spectra as a function of incident angle at a particular wavelength in each BPP mode band. As shown in Figure 2, the excitation of BPP modes in the four BPP bands with an equal bandwidth (50 nm) was investigated: 2100 to 2050 nm, 1100 to 1050 nm, 750 to 700 nm, and 550 to 500 nm. A decrease in resonance angle was observed when the excitation wavelength in each BPP band was increased, which indicates that the modal index of BPP modes decreases with an increase in excitation wavelength.[21,32] In particular, the resonance angle variation was different in each BPP wavelength band, with maximum for the shorter wavelength band (550 to 500 nm) and minimum for the longer wavelength band (2100 to 2050 nm). This is due to the fact that the modal index varies more at shorter wavelengths compared to longer wavelengths (Figure S2, Supporting Information). Also, the modal dispersion plot for surface and bulk plasmon modes is given in the Supporting Information (Figure S3). Since the BPP modes showed different angular variation for each BPP wavelength band, it could be used to design a potential plasmonic biosensor with different angular sensitivities for each BPP wavelength band. Therefore, we fabricated a micro-

change caused by the capture of biotin at the sensor surface within the microfluidic channel. The performance of the sensor was monitored based on the resonant angular shift in the reflectance spectrum when different concentrations (100×10^{-12} M to 10×10^{-6} M) of biotin in phosphate buffered saline (PBS) were injected into the sensor microchannel. It should be noted that we have used a single injection procedure for all our measurements. We first recorded the reflectance spectrum of the sensor device by injecting PBS. We then injected different concentrations of biotin, and the corresponding reflectance spectra as a function of incident angle were recorded after a reaction time of 40 min. Before each injection of a new concentration of biotin, PBS was introduced into the microchannel to remove the unbound and weakly attached biotin molecules. Since the mode at 1250 nm showed maximum angular sensitivity, we used that particular mode to study the biomolecular binding on the streptavidin-functionalized sensor surface, as shown in **Figure 4**a. A positive angular shift was obtained when the biotin concentration was increased, which is due to the increase in refractive index by the capture of biomolecules. As shown in Figure 4b, a nonlinear variation in angular shift, as a function of the biotin concentration, was observed. In particular, a large angular shift of 1° was obtained for 100×10^{-12} M biotin, while the obtained angular shift for 10×10^{-6} M was only 3°. These results demonstrate that the presented device has the capability to detect low molecular weight biomolecules with a sensitivity of 100×10^{-12} M. We used biotin as a proof-of-concept—biotin is a model system for small molecule compounds such as other vitamins, cancer-specific proteins, hormones, therapeutics, or contaminants such as pesticides or toxins.

We then performed experiments to detect high molecular weight macromolecules Cowpea mosaic virus. CPMV was propagated in and isolated from plants using methods described by Wen et al.[35] Different concentrations (1×10^{-15} M to 1×10^{-9} M) of CPMV were prepared in PBS and then injected into the sensor microchannel. PBS was injected into the channel to remove unbound and weakly attached CPMV before each injection of a new solution. After 20 min reaction time, the reflectance spectra for the highly diluted CPMV solutions were recorded (Figure 4c). A large angular shift of 1° was obtained for 1×10^{-15} M CPMV, whereas 1×10^{-9} M CPMV provided an angular shift of 6°. In addition, the binding of CPMV to the sensor surface was investigated by recording the reflectance spectrum over time (Figure S6, Supporting Information). Also in this case, we observed a nonlinear variation in angular shift with concentration. The sensor performance for the mode at 1250 nm is shown for different concentrations of CPMV in Figure 4d. Since the molecular weight of CPMV is very large, the sensor saturates very quickly when the concentration was increased up to 1×10^{-9} M. We then determined the sensitivity of the angular shift to the number of adsorbed molecules on the sensor surface and the data points in Figure 4b,d were fitted using a double exponential function (Figure S7, Supporting Information). The sensitivity analysis indicates that for CPMV at 1×10^{-15} M concentrations there are at most eight particles on average adsorbed in the sensing region of the surface. Given the corresponding large angular shift of 1°, this suggests that our novel sensor system could possibly achieve single-molecule detection levels. Therefore, the BPP-based biosensor has potential to match the sensitivity of nucleic acid technology-based sensors, with the added advantage that biospecimen processing

with antibodies, which can selectively capture speci c proteins, viruses, or even entire cells.

5. Conclusions

In summary, a grating coupled-hyperbolic metamaterial biosensor with enhanced angular sensitivity was demonstrated. In contrast to existing biosensors based on SPPs, we achieved different extreme sensitivity modes with 14, 12, 6, and 4 times higher angular sensitivity by exciting the bulk plasmon polaritons of HMM at 1250, 850, 650, and 530 nm wavelengths, respectively. With our new biosensor, we demonstrated the angular detection of low molecular weight biomolecules such as biotin in highly diluted solutions using a standard streptavidin-biotin af nity model. In addition, the angular detection of high molecular weight biomolecules such as CPMV, a model virus particle, at concentrations as low as 1×10^{-15} M was demonstrated. Since angular interrogation scheme offers higher measurement precision as compared to spectral interrogation, this miniaturized sensing scheme can be used for the development of next-generation sensors for biomedical, environmental, and chemical applications.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

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