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Micro-ball lens based optical biosensors

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A new optical biosensor is proposed and demonstrated using a new microfiber based structure so called micro-ball lens to sense glucose and uric acid concentration in distilled water. The micro-ball lens is formed at the cleaved tip of the microfiber coupler using an arcing technique. In the proposed sensor, the lens acts as a probe while a reflector is used as a target, is immersed into various concentrations of the glucose and uric acid solutions. It is observed that the intensity of the output power decreases linearly as the glucose concentration is varied from 0 % to 20 % with a sensitivity of 0.27 dB / % and a linearity of more than 98%. On the other hand, as the solution concentration of the uric acid varies from 0 to 500 ppm, the received reflected light intensity of the sensor decreases linearly from -58.1 dBm to -67.8 dBm with a sensitivity of 0.02 dB/ppm and a slope linearity of more than 99 %. The stability, high sensitivity and simplicity of the micro-ball lens probe make this new glucose and uric acid sensors are attractive alternative to other optical based sensing techniques.

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1. Introduction

Microfibers have attracted considerable interests in recent years, as they exhibit a number of exciting properties such as large evanescent field, strong confinement, easy configurability and high robustness [1-4]. These properties are advantageous for a wide range of applications including high-sensitivity optical sensors, nonlinear optics, atom trapping, micro/nano-scale photonic devices and for evanescent coupling to planar waveguides or microcavities. Likewise, biosensors have wide applications, including toxin and pathogen detection in food and water, as well as in biomarker detection for medical diagnostics [5].

The biosensor is defined as an analytical device that couples biological recognition element with a transducer to enable rapid, accurate and sensitive detection of target analytes. Meanwhile, fiber-optic biosensors (FOBs) are analytical devices in which a fiber optic device serves as a transduction element. FOBs have been widely investigated because of their potential sensitivity, fast detection, biocompatibility and adaptability to a wide variety of assay conditions [5-6]. Sensing in the FOBs is based on two fundamental concepts, namely spectroscopy and evanescent wavefield (EWF) interaction. The first concept is based fluorescence and absorption, which the sensing is determined by optical spectrum measurement. The second concept is based on the partial overlap of the evanescent guided electromagnetic wave with a medium whose refractive index (RI) is measured. In a standard optical fiber, the intensity of the EWF decays to almost zero at the outer surface of the cladding. Thus, light propagating in these fibers is insensitive to the surroundings. With the introduction of the low-loss microfiber, light can be guided along the tapered section for optical sensing within the visible and near infrared spectral ranges. Due to their extremely small diameters of microfibers, the amount of the penetration depth and intensity of the EWF can be significantly enhanced [7]. This will make it highly sensitive to the index change in the surrounding medium. As many optical biosensors are essentially refractive index sensors, the microfiber sensor can be utilized as a biosensor for numerous applications in health care, environmental protection, food safety, and petrochemical industries.

On the other hand, research interest in miniature integrated-optics whispering-gallery mode (WGM) resonant cavities has grown dramatically in recent years [8-9]. This has been due to the potential applications of such devices in optical communication including filtering, multiplexing, and switching. The technologies involved are mostly related to ring and disk-type WGM resonators fabricated from silica or silicon-based materials by traditional chemical vapor deposition (CVD) and photolithographic methods [10]. Since the size of the devices is relatively small (<100 μm diameter) the mass-production efficiency can be augmented, if a large number of such devices can be fitted on a single substrate wafer. Recently, optical WGM resonators with high quality factors (Q) have attracted attention especially in detection and sensing applications [11-12]. This is mainly attributed to the introduction of a relatively new micro-optical cavity type — the silica micro-ball lens. The micro-ball lens can be regarded as three-dimensional WGM resonators, which are 50–500 μm in diameter, and are often fabricated by simply melting the tip of an optical fiber. In a dielectric micro-ball lens, an optical WGM is formed by light propagating around the equator, spatially confined to a narrow beam near the ball lens's surface by total internal reflection. The extremely low WGM losses of fused-silica micro-ball lens allow them to be used as high-Q micro-resonators [13]. The Q-factors of these devices are reported to be so much higher compared to that of ring resonators, which translates directly into an ultra-narrow

resonant optical line width that can be employed in a variety of extremely precise measurement tasks.

In this paper, glucose and uric acid biosensors are demonstrated using an optical sensor based on micro-ball lens probe. The micro-ball lens structure is fabricated on the cleaved tip of the microfiber coupler (MFC). MFC is a multi-port device with stable and low-loss single-mode operation over a wide spectral range including short wavelengths.

2. Fabrication of Micro-ball lens structure

An MFC structure is made by laterally fusing and tapering two optical fibers using an experimental setup as shown in Fig. 1. In this experiment, a standard SMF was used to make a low noise MFC with the aid of the well-established flame-brushing technique. In the fabrication process, two fibers are brought into close proximity after the protective plastic jacket is removed. Then, both fibers are twisted at two different locations to make ovelapping contact. Then, while heated by a torch, the fibers are fused and stretched. The longitudinal profile of the conical transition tapers was achieved by reliable control of the hot zone and precise movement of the translation stages. During the tapering process, the coupling ratio is being monitored in real time by using a 1550 nm light source and power meters. The heating and pulling processes are stopped at the moment of achieving 50/50 coupling ratio. The diameter of the MFC waist was about 5µm and the lengths of tapered region and uniform waist were 70 mm and 40 mm, respectively.

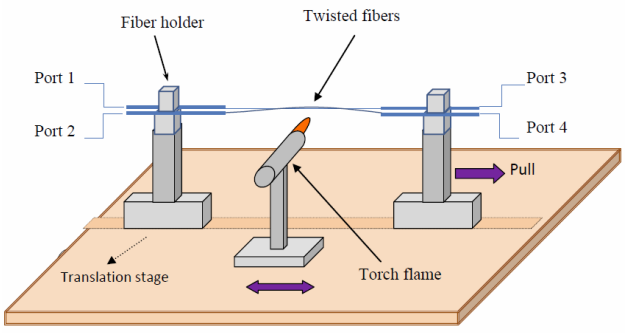

Fig. 1: Fabrication setup for microfiber coupler

The fabricated MFC is then used to fabricate a micro-ball lens by an arcing technique using a fusion splicing machine. The MFC is cut at the center of the minimum waist region to form micro-ball lens at the tip. The fabrication process involves two stages; loading and arcing processes. In the loading process, we open the windshield of the splicer, the fiber holders and the fiber clamps and load the prepared MFC tip into the left side of the holder with the tapered region in the V-groove. It is important to make sure that the MFC is properly aligned in the V-groove before closing the fiber holder. The fiber clamp is closed afterwards to hold the fiber on the V-groove. Once the fiber is loaded correctly, the windshield is closed and finally the formation of the micro-ball lens starts with the arching process. After setting the arc power, the cleaning arc power offset and the cleaning time, the fabrication of the micro-ball lens commences. The MFC tip absorbs the arc discharging heat and melts instantaneously. Due to the surface tension, the melting part of the MFC starts to form a spherical shaped tip gradually during solidification. As the spherical tip grows bigger, the effect of gravity force grows, pulling the tip towards the gravity field. This causes a drop of the spherical tip and increases the offset distance between the centre of the sphere and the axis of the fiber stylus. Fig. 2 show the microscope image of the micro-ball lens on the tip end of the MFC.

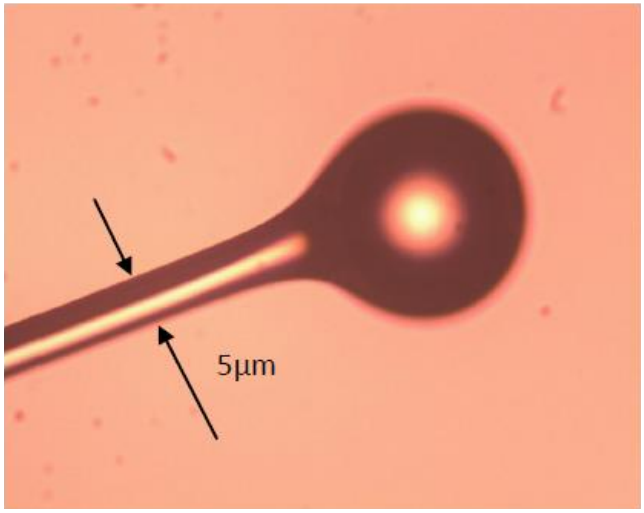

Fig. 2: Microscope image for the micro-ball lens fabricated at the cleaved tip of MFC

3. Micro-ball lens based glucose sensor

Monitoring glucose level in food is crucial for diabetic patients to prevent the disease from becoming proliferative. Enzymatic methods have been traditionally used to measure glucose concentration and provide point sample results [14-15]. An ideal glucose level sensor should be non-invasive, easy to use and reliable such that it provides predictable response to changing glucose concentrations. Taking these requirements into considerations, optical fiber sensors are an attractive option due to their non-invasive nature, immunity to electromagnetic interference, high sensitivity, compact size and possibility to be integrated into continuous measurement systems. In the previous section, microfiber coupler sensor has been demonstrated for glucose concentration measurement.

In this section, a new glucose sensor is proposed using a micro-ball lens fabricated at the cleaved top of an MFC as the sensing probe and 1550 nm light as a laser source. The proposed sensor uses 1550 nm laser source since the near infra-red (NIR) radiation has low absorption coefficient leading to a larger penetration depth in human tissue [16-17]. Furthermore, it could provide enough

absorption features for glucose that are sufficiently higher and more dominant than the other light intensity attenuation aspects occurring in human tissues such as scattering and absorption by water, skin pigments, blood and fat [17]. The probe is made at the minimum waist region of the MFC, which is cleaved to form a micro-ball lens using an arcing technique. This sensor provides high sensitivity, fast and predictable responses and the capability to be embedded into continuous measurement systems.

In the experiment, the previously developed MFC structure with 50/50 coupling ratio is used to fabricate micro-ball lens. Fig. 3 shows the experimental setup for the proposed micro-ball lens based glucose sensor. A petri dish with a mirror attached to its bottom was used as a container of liquid with various glucose concentrations. The coupled microfiber was fixed perpendicular to the mirror surface and the micro-ball lens was located at a distance of 1.3 mm from the center of the mirror throughout the experiment. One of the coupler ports was connected to an ASE light source operating in 1550 nm region. The other port was connected to an OSA as shown in the figure. The micro-ball lens was first immersed in de-ionized water to measure the output voltage of a 0 % glucose concentration, followed by liquid with glucose concentrations from 4 % to 20 %. The measurements were carried out for glucose solutions (D (+) - Glucose (C6H12O6 g/mol), John Collin Corporation, United Kingdom) with concentrations of 2, 4, 6, 8 and 10g per 50 ml of de-ionized water. During the experiment, the errors caused by temperature were taken to be negligible and the temperature was kept constant at 25°C.

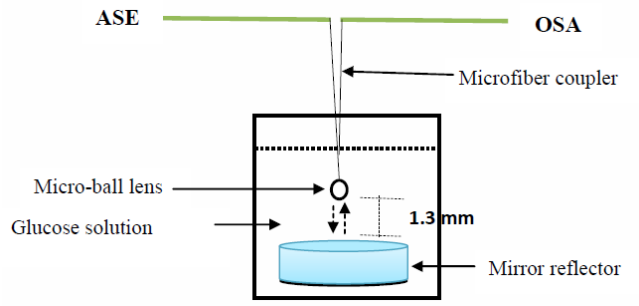

Fig. 3. The schematic diagram of the proposed glucose sensor.

Fig. 4 shows the output spectra of the micro-ball lens based sensor before and after the gap between the probe and mirror is filled with water. It can be seen that there is a significant decrease in the maximum reflected power as the medium is changed from air to water. When the ASE light is injected into the micro-ball lens through the input port of the MFC, reflection occurs at two places namely at the inner surface of the lens and the surface of the reflector. Reflection at lens inner surface is mainly from the Fresnel reflection, which is caused by the high index contrast between the silica glass and air or water. Both reflected beams interfere inside the micro-ball lens before it is transferred to the output port of the coupler. The interference fringes are then produced from the Fabry-Perot cavity. However, the fringes are almost disappeared as the gap between the lensed fiber probe and reflector is filled with water as shown in Fig. 4. This is attributed to the significant decrease in the reflection at the ball lens wall due to the significant decrease in the index contrast between two media. Subsequently, the intensity of the overall output spectrum is also reduced as shown in Fig. 4.

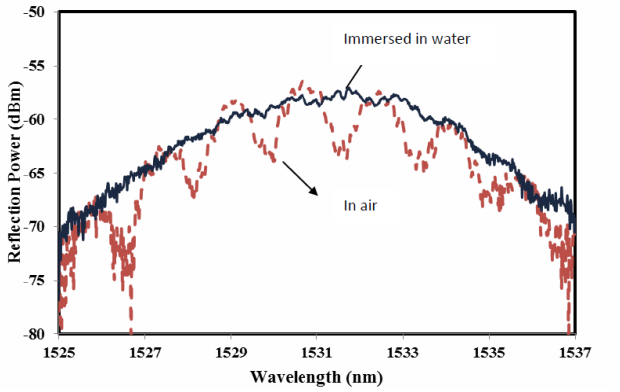

Fig. 4: Output spectra obtained as the micro-ball lens is immersed in air and water

Fig. 5 shows the peak output power of the micro-ball lens when immersed in different concentration of glucose solution. The results show that there is a linearly inverse relationship between the signals received in the receiving fiber as a function of the concentration of the glucose. From the experimental results, it can be concluded that an increase in glucose concentration can be detected by the decrease in the output power. The variation or modulation of the received light intensity is due to the change of the refractive index (RI) of the solution, which increases as the concentration of glucose increases. The reduction in index contrast between the two media decreases the back reflection at the lensed fiber inner surface and thus reduces the overall intensity received by the receiving fiber. It is observed that the output intensity of the sensor decreases linearly from -59.2 dBm to -65.0 dBm as the concentration of glucose is increased from 0 to 20 %. The sensitivity is obtained at 0.27 dB / % and the slope shows a good linearity of more than 98 %.

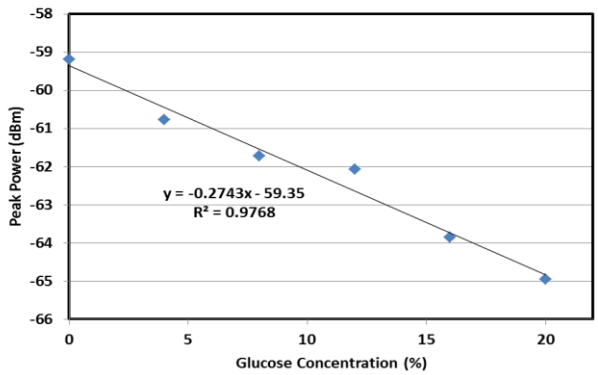

Fig. 5: Peak output power of the micro-ball lensed fiber for different glucose concentration

4. Micro-ball lens based uric acid sensor

Uric acid is a heterocyclic compound of carbon, nitrogen, oxygen and hydrogen. It is a product of the metabolic breakdown of purine nucleotides. Monitoring of uric acid is essential because abnormal levels of uric acid lead to several diseases like gout, Lesh-Nyhan syndrome, renal failure, hyperuricaemia, and physiological disorders, while patients with Wilson's disease are observed to have low uric acid level [18]. High level of serum uric acid is also considered as a risk factor for myocardial infarction and stroke [19]. Therefore, the need for uric acid biosensors is tremendously increasing [20-21]. Amperometric principles are the most common uric acid biosensor used for the detection of oxygen consumption, chemi-luminescense and fluoride ions. In practical, this detection method requires the electrode to be held at approximately 0.7V where other biological electro active molecules react with the surface of the electrode. In comparison with amperometric biosensor, potentiometer can reduce the interferences but a limitation of ion sensitive electrodes (ISEs) can only detect charged molecules [21]. In this section, a new micro-ball lens based sensor is proposed and demonstrate for measurement of uric-acid concentration.

In the experiment, the previously developed micro-ball lens is used for measurement of the uric acid concentration using a similar setup as Fig. 3. A petri dish with a mirror attached to its bottom was used as a container of liquid with various uric acid concentrations ranging from 0 to 500 ppm. The micro-ball lens was located at a distance of 1.3 mm from the center of the mirror throughout the experiment. We use an ASE light source operating in 1550 nm region as a light source. Prior the experiment, the refractive index of the uric acid solution was measured by using METTLER Toledo RE40D refractometer. It is obtained that the refractive index of the solution increases from 1.3330 to 1.3336 as the concentration of uric acid increases from 0 to 500 ppm. During the experiment, the temperature was kept constant at 25^{o}C.

Fig. 6 shows the variation of the reflected light from the mirror against the concentration of uric acid solution. The reflected light intensity is observed to be linearly decreases as the concentration of the uric acid solution increases. The received reflected light intensity of the sensor decreases linearly from -58.1 dBm to -67.8 dBm as the concentration of urid acid is increased from 0 to 500 ppm. This is attributed to the change of the refractive index of the solution, which increases from 1.3330 to 1.3336 as the concentration of uric acid increases from 0 to 500 ppm. The index increment reduces the index contrast between the liquid and silica ball and thus decreases the back reflection at the inner surface of the ball. This reduces the overall reflected light intensity. The sensitivity of the sensor is obtained at 0.02 dB/ppm with a slope linearity of more than 99 %. Overall, the sensor is observed to be sufficiently stable and repeatable. These results show that the proposed sensor is applicable and useful for the detection of biomolecular concentration such as uric acid. The sensor also has the ability to provide real time measurement. Somehow since the output is critically depends on the distance, further research is needed to study how to obtain an ideal distance and maintain it as a fixed value.

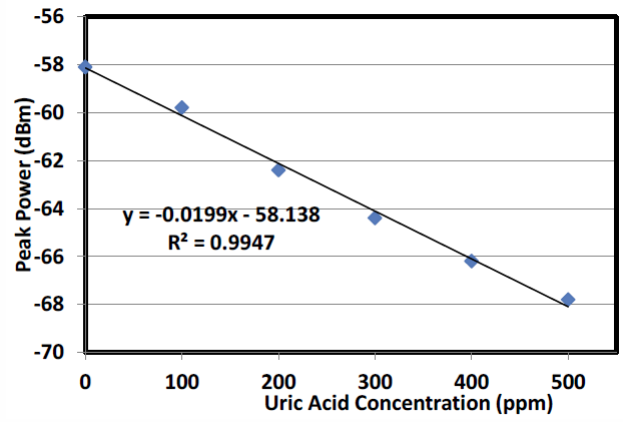

Fig. 6: Peak output power of the micro-ball lensed fiber for different uric acid concentration

5. Conclusion

We have successfully demonstrated new microfiber based structure so called micro-ball lens, which is then used for biosensors to sense glucose and uric acid concentration in distilled water. In the proposed biosensors, micro-ball lens which is formed at the cleaved tip of the MFC using an arcing technique, acts as a probe while a reflector acts as a target, is immersed into various concentrations of the glucose and uric acid solutions. It is observed that the intensity of the output power decreases linearly as the glucose concentration is varied from 0 % to 20 % with a sensitivity of 0.27 dB / % and a linearity of more than 98%. This new glucose sensor is an attractive alternative to other optical based glucose sensing techniques. On the other hand, as the solution concentration of the uric acid varies from 0 to 500 ppm, the received reflected light intensity of the sensor decreases linearly from -58.1 dBm to -67.8 dBm. The sensitivity of the micro-ball lens based uric acid sensor is obtained at 0.02 dB/ppm with a slope linearity of more than 99 %. The stability, high sensitivity and simplicity of the sensor make it suitable for various biosensors for pharmaceutical and biomedical applications.

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