New compact biochemical-sensor platform for industrial and environmental security

Ruth Shinar and Joseph Shinar

A new platform of photoluminescent sensors excited by an array of organic-light-emitting-device (OLED) pixels is being developed for a broad range of medical, environmental, and industrial applications.

Chemical and biological sensors are essential for many applications including industrial and environmental security. Photoluminescence (PL)-based sensors are advantageous due to their high detection sensitivity and specificity. These sensors are typically composed of a sensing component, whose PL is affected by the analyte, a light source that excites the PL, and a photodetector (PD). The structural integration of these components is highly desirable for generating compact sensors. Yet current light sources, e.g., lasers and inorganic LEDs, are often bulky or require intricate integration procedures.

The growing demand for low-cost sensors is driving efforts to develop sensor platforms using simplified fabrication procedures and miniaturization. The PL-based platforms typically employ an LED as the excitation source, which means that couplers and/or an intricate design for sensor miniaturization and configuration are required for multianalyte detection. However, OLEDs can also be used as excitation sources. With $\sim 100 \mu m^2$ to $> 100 mm^2$-sized pixels, they can be integrated on a substrate with the sensing component, and eventually with the PD as well. This uniquely-simple approach results in small devices that can be miniaturized to microarrays for multianalyte analysis.

Small-molecular OLEDs—consisting of organic layers typically sandwiched between an indium tin oxide (ITO) anode and a metal cathode—are easily fabricated by thermal vacuum evaporation. A forward bias generates the electroluminescence (EL). In the basic OLED-based sensor platform (see Figure 1), the OLED (thickness $< 0.5 \mu m$) and a $\mu m$-thick sensing film are fabricated on opposite sides of a common substrate, resulting in a very compact module. The viability of this structure comes from the intrinsic advantages of OLEDs as low-voltage bright, miniaturizable, flexible, and efficient blue-to-red light sources.

Figure 1. A structurally integrated sensor: the photodetector collects the photoluminescence between the gaps of the OLED pixels.

Figure 2. OLED-based glucose sensor. The sensor film is on the center $2 \times 2 mm^2$ pixels: their yellowish color is a combination of the green EL and red PL.

We have recently explored how the new platform can be used for monitoring various analytes. Figure 2 shows a glucose sensor in operation. Of the six pixels shown, the middle two are integrated with a sensing film containing an oxygen-sensitive dye and an enzyme that catalyzes the reaction between glucose and oxygen. When this reaction occurs, oxygen is consumed, resulting in increases in the PL intensity and lifetime that are proportional to the glucose concentration. Recent efforts have been aimed towards the realization of this platform for the detection of hydrazine and anthrax lethal factor (LF)—needed for industrial and environmental security—and for multianalyte detection in blood serum.

Continued on next page
Hydrazine, a highly-toxic volatile compound, is used as a monopropellant in NASA space shuttles and as a precursor in the synthesis of some polymers, plasticizers, and pesticides: the recommended eight-hour exposure limit is 10 ppb. The sensor is based on the known reaction between anthracene-2,3-dicarboxaldehyde (ADA) and hydrazine, which generates a product that is luminescent when excited by a blue OLED: the PL is proportional to the hydrazine level. By optimizing the OLED voltage and pulse width (see Figure 3), the limit of detection (LOD) was 60 ppb in ~1 min or ~1 ppb in 1 h, i.e., far exceeding the recommended limit.

The LF sensor under development is based on a fluorescence-resonance-energy-transfer (FRET) assay, where the anthrax-secreted LF enzyme cleaves certain labeled peptides at a specific site. The cleaving separates the FRET donor-acceptor pair, resulting in an increase in the previously-quenched donor PL. The donor/acceptor pair we used initially was a rhodamine-based dye/dark quencher. Preliminary results at 37 °C—using a green OLED—demonstrated a ~100% increase in the PL, which reached its maximal value after an incubation time of ~15 min (see Figure 4). We expect that the PL change will increase significantly if the EL background is decreased and the Stokes shift of the fluorescent dye is increased.

In conclusion, the use of an ultra-thin OLED excitation source in PL-based structurally-integrated sensors will enable the realization of small, inexpensive, field-deployable sensors. The utility of this platform has been demonstrated for analytes in medical, environmental, and security monitoring. As examples, oxygen and glucose sensors perform comparably or better than those excited by other sources. The LOD for hydrazine gas far exceeds the recommended requirements. The cleaving of a rhodamine-based, FRET-labeled substrate by anthrax LF yielded an OLED-excited PL that increased by a factor of ~2. Luminescent dyes with a large Stokes shift and reduction of the EL background will increase the PL change. Current efforts focus also on developing OLED-based (micro)arrays, where individually addressable pixels monitor different analytes.

**Author Information**

**Ruth Shinar**  
Microelectronics Research Center  
Iowa State University  
Ames, IA

**Joseph Shinar**  
Ames Laboratory and Physics Department  
Iowa State University  
Ames, IA

**References**


© 2006 SPIE — The International Society for Optical Engineering