Published in final edited form as:

IEEE Sens J. 2010 December ; 10(12): 1824–1832. doi:10.1109/JSEN.2010.2047497.

CMOS Imaging of Pin-Printed Xerogel-Based Luminescent Sensor Microarrays

Lei Yao [Student Member, IEEE],

Department of Electrical and Computer Engineering, McGill University, Montreal, QB, H3A2A7 Canada

Ka Yi Yung,

Department of Chemistry, University at Buffalo, The State University of New York, Buffalo, NY 14260 USA

Rifat Khan,

Department of Electrical and Computer Engineering, McGill University, Montreal, QB, H3A2A7 Canada

Vamsy P. Chodavarapu [Member, IEEE], and

Department of Electrical and Computer Engineering, McGill University, Montreal, QB, H3A2A7 Canada

Frank V. Bright

Department of Chemistry, University at Buffalo, The State University of New York, Buffalo, NY 14260 USA

Vamsy P. Chodavarapu: vamsy.chodavarapu@mcgill.ca

Abstract

We present the design and implementation of a luminescence-based miniaturized multisensor system using pin-printed xerogel materials which act as host media for chemical recognition elements. We developed a CMOS imager integrated circuit (IC) to image the luminescence response of the xerogel-based sensor array. The imager IC uses a 26×20 (520 elements) array of active pixel sensors and each active pixel includes a high-gain phototransistor to convert the detected optical signals into electrical currents. The imager includes a correlated double sampling circuit and pixel address/digital control circuit; the image data is read-out as coded serial signal. The sensor system uses a light-emitting diode (LED) to excite the target analyte responsive luminophores doped within discrete xerogel-based sensor elements. As a prototype, we developed a 4×4 (16 elements) array of oxygen (O_2) sensors. Each group of 4 sensor elements in the array (arranged in a row) is designed to provide a different and specific sensitivity to the target gaseous O_2 concentration. This property of multiple sensitivities is achieved by using a strategic mix of two oxygen sensitive luminophores ($[Ru(dpp)_3]^{2+}$ and $[Ru(bpy)_3]^{2+}$) in each pin-printed xerogel sensor element. The CMOS imager consumes an average power of 8 mW operating at 1 kHz sampling frequency driven at 5 V. The developed prototype system demonstrates a low cost and miniaturized luminescence multisensor system.

Index Terms

CMOS imager; luminescence; multi sensor system; optical sensors; oxygen sensors; pin-printing; sensor microarrays; xerogel

I. Introduction

Sensor microarray technologies have created tremendous opportunities in many application areas including drug discovery, genomics, proteomics, and point-of-care diagnostics [1]–[5]. Sensor microarrays consist of an ordered ensemble of microscale elements (spots) that each contains a specific immobilized recognition element on a largely planar surface and allow one to multiplex and demultiplex biological or chemical information from multicomponent samples. To date, we find a large number of sensor microarray related technologies including those that exploit cantilevers [6], electrochemistry [7], surface plasmon resonance [8], piezoelectric effect [9], and surface acoustic waves [10]. We prefer luminescence-based sensors because they offer several advantages including fast response, they do not poison sample media, and they require no additional reagents [11]. Luminescence sensing is also extensively used for DNA-chip and Gene-chip technologies in genomic research and diagnostics [12]. Typically, in these microarray systems, complementary metal-oxide semiconductor (CMOS) or charge coupled device (CCD) imagers are used to convert the optical signals into electrical signals. CMOS imagers are more preferable because of monolithic integration of photodetection elements and signal processing circuitry leading to low-cost miniaturized systems [3].

Previously, many different approaches have been used to fabricate sensor microarrays including pin [13], ink-jet, and screen printing [14], [15], and photolithography [16]. Micro-pin printing methods use one or more metallic pins to contact print/spot liquid onto a planar surface (e.g., a microscope slide). Pin-printing allows low-cost, accurate, and high-throughput immobilization of a variety of recognition materials. We have also explored the use of pin-printing methods to fabricate reusable xerogel-based chemical sensor arrays [13]. Xerogels are sol-gel derived materials that have been extensively used as immobilization media for sequestering a variety of recognition elements including fluorophores, enzymes, and modified nanoparticles [17]–[20]. Xerogels are porous glasses and their appeal for biorecognition elements derives from their production at room temperature, thermal stability, tunable pore dimensions and distributions, biocompatibility, adjustable pH, and a broad optical transparency window [21].

Sensor microarrays would provide simultaneous multiple independent sensor responses. This would allow incorporation of sensor redundancy and improving the reliability of sensor systems, in general, purpose applications. In the current research, we use the pin-printing technique to develop a 4×4 array of xerogel sensor elements that are responsive to ambient gaseous oxygen (O_2) concentration as a simple prototype implementation. The sensor array is excited by an LED ($\lambda_{\text{peak}} = 470 \text{ nm}$) and the subsequent emission ($\lambda_{\text{peak}} = 610 \text{ nm}$) from the sensors is detected by a CMOS imager. The developed CMOS imager uses a 26×20 (520 elements) array of active pixel sensors and each active pixel includes a high-gain phototransistor as the photodetector. Fig. 1 shows a simplified block diagram of the CMOS imager-based multisensor system. The oxygen sensor described here is based on co-encapsulating the luminophores tris(4,7-diphenyl-1,10-phenanthroline) ruthenium(II) ($[\text{Ru}(\text{dpp})_3]^{2+}$) and *tris*(2,2'-bipyridyl) dichlororuthenium(II)hexahydrate ($[\text{Ru}(\text{bpy})_3]^{2+}$), both of which are well known to be responsive to gaseous oxygen, in the xerogel matrices. Assuming all the luminophore molecules in the xerogel-thin-film are equally accessible to the O_2 molecules, one can write the quencher-dependent response as (1) [22], [23]

$$\frac{I_0}{I} = \left[\frac{f_1}{(1+k_q\tau_{0,1}[Q])} + \frac{f_2}{(1+k_q\tau_{0,2}[Q])} \right]^{-1} \quad (1)$$

where f_1 , and f_2 denote that fraction of the total emission (in the absence of quencher) that arises from luminophore 1 and 2, respectively; $\tau_{0,1}$ and $\tau_{0,2}$ are the excited-state luminescence lifetimes from luminophores 1 and 2, respectively, in the absence of quencher; k_q is the bimolecular quenching constant between the luminophores and the quencher molecule; and I_0 and I are the luminescence intensity in the absence and presence of quencher (O_2), respectively. [Note: In (1), we assume that k_q , the bimolecular quenching constant, is luminophore-independent because the luminophores are structurally similar and thus would be distributed similarly within the xerogel host matrix]. Inspection of (1) shows that one can continuously tune the sensor response profile (i.e., I_0/I versus $[O_2]$) within a single analyte-permeable host matrix by creating sensor elements that contain different fractions of luminophore 1 and 2.

Thus, in the current research, we use the strategy of mixing a combination of different luminophores (e.g., $[Ru(dpp)_3]^{2+}$ and $[Ru(bpy)_3]^{2+}$) encapsulated in each xerogel sensor element. The luminophores are selected such that they show the same emission spectrum but each shows different τ_0 [24]. In Section II, we provide the details of the CMOS imager IC. Section III describes the fabrication of xerogel sensor microarray. Finally, Section IV describes the experimental results and discussion.

II. CMOS Imager Integrated Circuit

Vo-Dinh *et al.* [25] first reported a CMOS phototransistor-based imager to implement a microsystem for DNA microarrays. There are several subsequent reports using CMOS imager ICs for luminescence monitoring which are related to DNA microarrays [26], [27], modeling and simulation [28], and lifetime measurements [29]. We find no major reports of CMOS imager systems for use in luminescence monitoring of discrete biochemical sensor microarrays. There are however previous reports on the use of discrete CMOS photodetectors for luminescence monitoring in biochemical sensors [30]–[33]. Fig. 2 shows the circuit architecture of the developed CMOS imager IC. The CMOS imager essentially consists of three blocks: (i) active pixel array; (ii) digital control circuit and X-Y address circuit; and (iii) analog signal processing circuit. The imager uses an active pixel sensor (APS) array based on standard circuit designs [34]. To operate the APS array, three digital control signals are required: RST , S_x and S_y . The output signal of pixel array is read out serially at $V_{arrayout}$ by the signal processing circuit. The signal processing circuitry mainly includes the correlated doubling sampling (CDS) circuit and integrator circuit. The CDS block is used to minimize Fixed pattern noise (FPN) and the integrator is used to integrate the ramp signal into DC form for further image processing. Three digital signals are needed to operate the signal processing circuit: SH , RST , and the complementary signal of SH . All required digital signals in imager IC are generated on-chip by the digital control circuit/ X-Y address circuit and are synchronized with the external clock signal CLK . Fig. 3 shows the microphotograph of the fabricated $2.2\text{ mm} \times 2.2\text{ mm}$ CMOS imager IC in AMI $1.5\text{ }\mu\text{m}$ process available through MOSIS (www.mosis.org).

A. Active Pixel Sensor (APS) Array

The CMOS imager which forms the essential part of many commercial digital cameras typically consists of an array of individually accessible photosensitive pixels. The developed imager IC includes 26×20 (520 elements) array of APSs. Each APS pixel uses a vertical p-n phototransistor as the photodetector. Phototransistors can produce current levels that are

several times larger than a comparably sized photodiodes needed for LED-based sensors as they produce very weak luminescence signals. The vertical p-n-p phototransistor can only be used in an emitter-follower configuration and is formed by the p-active (emitter)/n-well (base)/p-substrate (collector). Each APS element [30], [35] contains the phototransistor and four MOSFETs (M_{rs} , M_{rd} , M_x , and M_y), as shown in Fig. 1. M_{rs} is the reset transistor and M_{rd} acts as a source follower. M_x and M_y are addressing transistors that perform the serial pixel readout. The serial readout data is then processed by the analog signal processing circuit. Typical waveforms of the digital control signals for the imager are shown in Fig. 2.

During imager operation, there are three periods in a single scan circle: RESET, SAMPLING, and HOLD. First, in RESET period, the signal RST is activated (5 V) to charge the emitter terminal of the phototransistor with the impinged optical signals to V_{dd} (5 V). During SAMPLING period, RST is deactivated (0 V) and the signal is sampled through M_x and M_y . The voltage drop between RESET and SAMPLING period is $V_{arrayout}$ which can be calculated according to (2), where I_{pt} is the current generated by the phototransistor when excited by the luminescence, T_{sample} is the time of the SAMPLING period, C_p is the parasitic capacitance existing in the phototransistor. Finally, in HOLD period, the signal is processed by signal processing circuit while the rest of the pixel array is in idle mode

$$\Delta V_{arrayout} = \frac{I_{pt} \cdot T_{sample}}{C_{pt}}. \quad (2)$$

From Fig. 2, the size of each active pixel is 54 by 58 μm with a fill factor of 28.7%. The W/L ratios for the transistors in the active pixel are: M_{rs} 4.8/1.6, M_{rd} 8/1.6, M_x 9.6/1.6, M_y 9.6/1.6 (all dimensions are in μm). The size of phototransistor is 30 μm by 30 μm . Based on the AMIS 1.5 μm process parameter, the parasitic capacitor C_p can be estimated as 90 fF. The typical value of the photo current generated by phototransistor under luminescence ranges between few hundred pico-amperes to several tens of nano-amperes. As an example, with a moderate sampling frequency of 1 kHz, $\Delta V_{arrayout}$ can be calculated as 556 mV when $I_{pt} = 100$ pA [using (2)], which can be easily handled by the signal processing circuit.

B. Signal Processing Circuitry

As shown in Fig. 2, the serial output of the pixel array is read out at $V_{arrayout}$ by the signal processing circuitry which consists of a CDS circuit [34] and an integrator circuit. This circuit component operates as following: When the signal RST is active (5 V), a selected pixel enters its RESET period, the voltage in the sample and hold capacitor (C_{sh}) is clamped at V_{clp} . At the same time the integrator is isolated from CDS circuit and works as a simple voltage follower. The output voltage of integrator during RESET period is V_{clp} . When the RST signal goes low (0 V), the current pixel enters the SAMPLING period, the output voltage of the pixel array $V_{arrayout}$ drops proportional to the power of the optical signal. Then, C_{sh} samples the drop in voltage valve and then the voltage is integrated into a DC value by the integrator circuit. Since CDS only samples the voltage drop and the DC level of each pixel is isolated by C_{in} , FPN noise which is caused by pixel mismatch can be greatly suppressed. The output voltage of CDS block, V_{sh} , can be expressed by (3). The output voltage V_{out} during the HOLD period can be expressed by (4). Substituting the value of ΔV_{pixout} obtained from (4) into (2), the relationship between ΔV_{out} and I_{pt} can be derived as presented by (5). Thus, the sensitivity, S , of CMOS imager IC can be calculated in (6)

$$V_{sh} = V_{clp} - \Delta V_{arrayout} \cdot \frac{C_{in}}{C_{in} + C_{sh}} \quad (3)$$

$$V_{out} = V_{clp} + \Delta V_{arrayout} \cdot \frac{C_{in}}{C_{in} + C_{sh}} \cdot \frac{C_{sh}}{C_{fb}} \quad (4)$$

$$V_{out} = V_{clp} + \frac{C_{in} C_{sh} T_{sample}}{C_{fb} C_{pt} (C_{in} + C_{sh})} \cdot (I_{pt} + I_{dark}) \quad (5)$$

$$S = \frac{\Delta V_{out}}{\Delta I_{pt}} = \frac{C_{in} C_{sh} T_{sample}}{C_{fb} C_{pt} (C_{in} + C_{sh})} \quad (6)$$

The CDS circuit block consists of three (3) switches and two (2) capacitors, and the integrator consists of one switch, one capacitor and an operational amplifier (OPA). The OPA used in integrator is a high-gain high-swing rail-to-rail folded-cas-code op-amp and its circuit architecture is shown in Fig. 4. The W/L ratio of all switches is 4.8 $\mu\text{m}/1.6 \mu\text{m}$. Compromising among chip area, detection sensitivity and noise issues, we choose the value of capacitors as follows: $C_{in} = 4 \text{ pF}$, $C_{sh} = 1 \text{ pF}$ and $C_{fb} = 0.5 \text{ pF}$. With $C_p = 90 \text{ fF}$ and $T_{sample} = 0.1 \text{ ms}$, the detection sensitivity S can be calculated as 1.78 V/nA (output voltage/generated photocurrent). For a good output dynamic range, V_{clp} has to be decided carefully. First, as seen from (3) and (4), V_{clp} can neither be set too high (V_{sh} will be saturated) nor too low (V_{out} will be saturated). The calculated ideal point for V_{clp} from simulations is 1.67 V (based on the ratio of C_{sh} to C_{sh}). However, considering the dynamic range of OPA and mismatch issue of the capacitors, we set V_{clp} as 2 V to obtain a good output dynamic range.

The simulation response of the imager IC is shown in Fig. 5. Here, the phototransistor is modeled by an ideal current source connected with a capacitor in parallel. In the simulation, the photocurrent is varied from 0.2 to 3.2 nA with a step of 0.2 nA. This range of photocurrents was selected based on our previous experience working with these CMOS optical biochemical sensors [35].

C. Digital Control Circuit

The digital control circuit generates all digital signals shown in Fig. 2. An external clock signal is used to generate and synchronize all digital signals. In subsequent prototypes, the clock signal can be generated on-chip. The signal, SH , and the complementary signal of SH are generated by a nonoverlapping clock generator as shown in Fig. 6. RST is generated by a frequency divider. Sx and Sy for each pixel are generated by X - Y address circuit. The X - Y address is generated by a series of digital shifters using the SH signal.

D. CMOS Imager System Characterization

Here, we detail the CMOS imager IC experimental characterization. In luminescence monitoring applications the phototransistor (or in general the photodetector device) plays a critical role. Previously, we performed spectral responsivity characterization of a larger sized array of similarly designed phototransistor. We noticed a high responsivity (Ampere/Watt) over a wide range of the visible wavelength spectrum which is well suited for luminescence sensor applications. For characterizing the current CMOS imager, we used an orange LED ($\lambda_{peak} = 595 \text{ nm}$) to illuminate the CMOS imager uniformly. This wavelength was chosen to match the peak luminescence of our oxygen sensors which is in the range of 590–615 nm. An external clock generated by a Data Acquisition System (National Instruments) is used and its frequency is varied from 200 Hz to 10 kHz (equivalent to integration or sampling time, T_{sample} , varying from 50 μs to 2.5 ms). The optical power of

the LED source is measure by a power meter placed near to the imager. Fig. 7 shows the relationship between incident optical power and sampling/exposure time T_{sample} . Fixed pattern noise (FPN) and dark current are two key performance parameters for the CMOS imager. We kept the CMOS imager IC in complete dark conditions at room temperature (24 °C) and observe the imager IC output which is shown in Fig. 8. We also varied the sampling time T_{sample} (50 μs to 2.5 ms) to study the variation of FPN with different exposure times. We notice that the measured FPN noise is approximately 80 mV and its variation is insignificant with different sampling times. The output referred dark current shows a proportional increase with sampling time T_{sample} . The output referred dark current varied from 500 to 830 mV when the sampling time T_{sample} was varied from 50 μs to 2.5 ms. We notice that the output referred dark current is comparatively high in the current imager and it is mainly due to the use of high-gain phototransistor in the APS array.

There is a tradeoff between detecting low luminescence signals and faster image readout from imager. Detecting weak luminescence signals requires longer integration or sampling times. Using our previous knowledge on the approximate power luminescence signals with LED excitation, we selected 500 μs as the sampling time. This sampling time provides a good compromise between achieving high detection sensitivity and faster sampling time and this relationship is shown with measured response in the inset in Fig. 7.

III. Pin-Printed Xerogel Sensor Array

A. Chemical Reagents

The following reagents were used: tris(4,7'-diphenyl-1,10'-phenanthroline) ruthenium(II) chloride pentahydrate ($[\text{Ru}(\text{dpp})_3]\text{Cl}_2 \cdot 5\text{H}_2\text{O}$) and tris(2,2'-bipyridyl)ruthenium(II) chloride hexahydrate ($[\text{Ru}(\text{bpy})_3]\text{Cl}_2 \cdot 6\text{H}_2\text{O}$) (GFS Chemicals, Inc.); tetraethylorthosilane (TEOS) (Sigma-Aldrich); *n*-octyltriethoxysilane (C8-TEOS) (Gelest Inc.); HCl (Fisher Scientific Co.); and EtOH (Quantum Chemical Corp.). Corning # 2947 microscope slides (Fisher Scientific Co.) were used as the xerogel substrates. The $[\text{Ru}(\text{dpp})_3]^{2+}$ and $[\text{Ru}(\text{bpy})_3]^{2+}$ were purified as described in the literature [36]. All other reagents were used as received without further purification. Deionized water was prepared to a specific resistivity of at least 18 M Ω -cm by using a Barnstead NANOpure® II system.

B. Sol Stock Solution Preparation

Class II xerogels were produced by creating a sol solution composed of TEOS (1.448 mL, 6.5 mmol) and C8-TEOS (2.052 mL, 6.5 mmol), EtOH (2.52 mL, 44 mmol) and HCl (0.8 mL of 0.1 N HCl, 0.08 mmole). This mixture was capped and magnetically stirred under ambient conditions for 1 h.

C. Luminophore-Doped Sol Solution Preparation

It proved beneficial to have the sensor elements within the array exhibit nearly equal emission intensities in the absence of quencher. In this way, we could record a single image at each O₂ concentration and avoid multiple images at varying gains or integration times and the concomitant backgrounds. Table I reports the $[\text{Ru}(\text{bpy})_3]^{2+}/[\text{Ru}(\text{dpp})_3]^{2+}$ -doped sol compositions that we prepared to yield the unique response profiles and achieve the aforementioned goal. These luminophore-doped sol solutions were capped, mixed with a touch mixer for 2 min, and stored in the dark without stirring under ambient conditions. These sol solutions were then allowed to hydrolyze and condense for 24 h. Sensor elements were formed from these sols.

D. Metal Mask

A metal mask is used to improve the contrast for the detected luminescence by reducing the background excitation signal. The mask was fabricated on the microscope slide with sputtered metal layers of titanium (adhesion layer, 500 nm thick) and gold (masking layer, 750 nm thick). The metal layers were photolithographically patterned and etched with 1:2:10 I₂:KI:H₂O and 20:1:1 H₂O:HF:H₂O₂ to define transparent circular openings for printing the xerogel sensors, as shown in Fig. 9.

E. Sensor Array Fabrication

We formed the xerogel-based sensor elements by pin-printing with a 600 μm diameter solid tungsten pin onto the surface of the masked substrates [13], [37]. The substrates were cleaned by an EtOH rinse, 1–2 h soak in 1 M NaOH, deionized water wash, 1–2 h soak in 1 M HCl, deionized water wash, and oven dried at 80 °C for > 3 h. The relative humidity within the print chamber was maintained at $30 \pm 5\%$, the temperature was 20 °C–22 °C, the pin velocity was 15 mm/s, pin acceleration was 15 mm/s, and the pin-to-surface contact time was 7 ms. All sensor arrays were aged under ambient conditions in the dark for at least seven (7) days prior to assessment and testing.

IV. Experimental Results and Discussion

The experimental setup for characterizing the CMOS sensor system is shown in Fig. 10. A laboratory DC power supply (INSTEK: PST-3202) and a function generator (TABOR: WW2572) are used to supply the CMOS imager IC and power the blue LED ($\lambda_{\text{peak}} = 470$ nm) which is used as excitation source. The blue LED is located in a test chamber along with the 4×4 O₂ sensor array. The O₂ concentration in the test chamber controlled by a custom built flow-meter, consisting of a matched pair of air flow controllers connected to O₂ and N₂ gas cylinders. A long-pass optical filter (cutoff, $\lambda = 590$ nm) is placed between the chamber and the imager IC to remove the excitation optical signal and allow only the luminescence to fall on the imager. A bi-convex optical lens (Diameter = 25.4 mm, Focal length = 25.4 mm) is used to focus the luminescence signal from the sensor array onto the CMOS imager. A data acquisition card (NI: DAQPad-6259) connected to a personal computer is used to process the CMOS imager output signal and perform the image processing. The imager sampling frequency is set at 1 kHz ($T_{\text{sample}} = 500 \mu\text{s}$) in all the subsequent experiment testing.

Fig. 11 shows the processed luminescence images of the 4×4 sensor array when the O₂ concentration is varied from 0% to 100% in steps of 20%. All four sensor elements arranged in a row are designed to provide identical sensitivity to O₂ concentrations. We notice that some sensor elements in the array (e.g., D-1) do not provide significant luminescence. This is due to the fabrication error in aligning the transparent circles in the gold-coated glass mask and the pin of the pin-printer system. This can be easily overcome in future by preparing a separate reference transparent circle in the gold-coated glass mask with respect to which the rest of the array can be fabricated. The developed sensors are responsive over the complete range of oxygen concentrations. We observed an average system response time on the order of 2 s. These graphs provide us important information related to the fixed pattern noise (FPN) and the average amount of background signal leaking to the photodetector due to the excitation source or ambient noise source.

Fig. 12 shows the peak recorded voltage value from each sensor element in the array for a single experiment run for different O₂ concentrations. As expected the sensors provide a different sensitivity to O₂ concentrations. The average of the measured sensitivity of the different sensors is as follows, sensor type A (1.9 ± 0.47 mV/%O₂ concentration), sensor

type B (2.7 ± 0.2 mV/%O₂ concentration), sensor type C (3.1 ± 0.21 mV/%O₂ concentration), sensor type D (3.6 ± 0.45 mV/%O₂ concentration).

Fig. 13 presents the Stern–Volmer response of the sensor array. Here, the background signal is subtracted from the voltage values for different O₂ concentrations. The graph plots the ratio of the voltage value when no oxygen is present to the voltage value for different O₂ concentrations. The data is averaged for the four (4) sensors of the same type in the array and for multiple experiment runs. The Stern–Volmer characteristics match the response obtained from our previous studies [24], [31]. Finally, our previous studies in these sensor materials reveal that the sensors can be used for continuous monitoring, they are reversible, they provide fast response time (on the order of few seconds), and can remain stable over extended period of time depending on the amount of usage [38].

V. Conclusion

We present the design and implementation of a luminescence multisensor system using pin-printed xerogel thin-films housing oxygen sensitive luminophore and CMOS imager. The imager uses standard APS configuration with a high-gain phototransistor as the photosensitive element. The developed imager consists of 26×20 (520 elements) array of active pixel sensors using which we image the luminescence of a 4×4 array of xerogel sensor elements. We demonstrate the concept of multiple detection sensitivities by using a strategic mix of two oxygen sensitive luminophores ([Ru(dpp)₃]²⁺ and [Ru(bpy)₃]²⁺) in each pin-printed xerogel spot. In the future, similar multisensor systems can be incorporated with abilities to suppress environmental variations such as temperature and humidity the affect the calibration of the sensor elements. This work shows a good perspective to create robust multisensor detection systems with low-cost and miniaturized configuration.

Acknowledgments

This work was supported in part by the Canada Natural Sciences and Engineering Research Council (NSERC), Canada Foundation for Innovation (CFI), Le Fonds québécois de la recherche sur la nature et les technologies (FQRNT), and United States National Science Foundation (NSF).

REFERENCES

1. Nagl S, Wolfbeis OS. Optical multiple chemical sensing: Status and current challenges. *Analyst*. 2007; vol. 132:507–511. [PubMed: 17525805]
2. Bally M, Halter M, Voros J, Grandin HM. Optical microarray biosensing techniques. *Surface and Interface Anal*. 2006 Nov.vol. 38:1442–1458.
3. Golden JP, Ligler FS. A comparison of imaging methods for use in an array biosensor. *Biosens. Bioelectron*. 2002 Sep.vol. 17:719–725. [PubMed: 12191919]
4. Cho EJ, Bright FV. Optical sensor array and integrated light source. *Anal. Chem*. 2001 Jul.vol. 73:3289–3293. [PubMed: 11476227]
5. Ligler FS, Sapsford KE, Golden JP, Shriver-Lake LC, Taitt CR, Dyer MA, Barone S, Myatt CJ. The array biosensor: Portable, automated systems. *Anal. Sci*. 2007 Jan.vol. 23:5–10. [PubMed: 17213615]
6. Baller MK, Lang HP, Fritz J, Gerber C, Gimzewski JK, Drechsler U, Rothuizen H, Despont M, Vettiger P, Battiston FM, Ramseyer JP, Fornaro P, Meyer E, Guntherodt HJ. A cantilever array-based artificial nose. *Ultramicroscopy*. 2000 Feb.vol. 82:1–9. [PubMed: 10741645]
7. Stefan RI, van Staden JF, Aboul-Enein HY. Electrochemical sensor arrays. *Critical Reviews in Analytical Chemistry*. 1999; vol. 29:133–153.
8. Wong CL, Ho HP, Suen YK, Kong SK, Chen QL, Yuan W, Wu SY. Real-time protein biosensor arrays based on surface plasmon resonance differential phase imaging. *Biosens. Bioelectron*. 2008 Dec.vol. 24:606–612. [PubMed: 18644712]

9. Barko G, Abonyi J, Hlavay J. Application of fuzzy clustering and piezoelectric chemical sensor array for investigation on organic compounds. *Analytica Chimica Acta*. 1999 Oct.vol. 398:219–226.
10. Grate JW. Acoustic wave microsensor arrays for vapor sensing. *Chem. Rev.* 2000 Jul.vol. 100:2627–2647. [PubMed: 11749298]
11. McDonagh C, Burke CS, MacCraith BD. Optical chemical sensors. *Chem. Rev.* 2008 Feb.vol. 108:400–422. [PubMed: 18229950]
12. Heller MJ. DNA microarray technology: Devices, systems, and applications. *Annu. Rev. Biomed. Eng.* 2002; vol. 4:129–153. [PubMed: 12117754]
13. Cho EJ, Tao ZY, Tehan EC, Bright FV. Multianalyte pin-printed biosensor arrays based on protein-doped xerogels. *Anal. Chem.* 2002 Dec.vol. 74:6177–6184. [PubMed: 12510736]
14. Calvert P. Inkjet printing for materials and devices. *Chem. Mater.* 2001 Oct.vol. 13:3299–3305.
15. Zhao WA, van den Berg A. Lab on paper. *Lab on a Chip*. 2008; vol. 8:1988–1991. [PubMed: 19023461]
16. Wallraff G, Labadie J, Brock P, DiPietro R, Nguyen T, Huynh T, Hinsberg W, McGall G. DNA sequencing on a chip. *Chemtech*. 1997 Feb.vol. 27:22–32.
17. Tao ZY, Tehan EC, Bukowski RM, Tang Y, Shughart EL, Holthoff WG, Cartwright AN, Titus AH, Bright FV. Templated xerogels as platforms for biomolecule-less biomolecule sensors. *Analytica Chimica Acta*. 2006; vol. 564:59–65. [PubMed: 17723362]
18. Bukowski RM, Chodavarapu VP, Titus AH, Cartwright AN, Bright VF. Phase fluorometric glucose biosensor using oxygen as transducer and enzyme-doped xerogels. *Electron. Lett.* 2007; vol. 43:202–204.
19. MacCraith BD, McDonagh C, Mcevoy AK, Butler T, O'Keeffe G, Murphy V. Optical chemical sensors based on sol-gel materials: Recent advances and critical issues. *J. Sol-Gel Sci. Technol.* 1997; vol. 8:1053–1061.
20. Jeronimo PCA, Araujo AN, Montenegro MCBSM. Optical sensors and biosensors based on sol-gel films. *Talanta*. 2007; vol. 72:13–27. [PubMed: 19071577]
21. Dave BC, Dunn B, Valentine JS, Zink JJ. Sol-gel encapsulation methods for biosensors. *Anal. Chem.* 1994; vol. 66:A1120–A1127.
22. Lakowicz, JR. *Principles of Fluorescence Spectroscopy*. 2 ed.. New York: Kluwer Academic; 1999.
23. Demas JN, DeGraff BA, Coleman PB. Oxygen sensors based on luminescence quenching. *Anal. Chem.* 1999 Dec.vol. 71:793a–800a.
24. Tang Y, Tao ZY, Bukowski RM, Tehan EC, Karri S, Titus AH, Bright FV. Tailored xerogel-based sensor arrays and artificial neural networks yield improved O₂ detection accuracy and precision. *Analyst*. 2006; vol. 131:1129–1136. [PubMed: 17003861]
25. Vo-Dinh T, Alarie JP, Isola N, Landis D, Wintenberg AL, Ericson MN. DNA biochip using a phototransistor integrated circuit. *Anal. Chem.* 1999 Jan.vol. 71:358–363. [PubMed: 9949727]
26. Eltoukhy H, Salama K, El Gamal A. A 0.18- μ m CMOS bioluminescence detection lab-on-chip. *IEEE J. Solid-State Circuits*. 2006 Mar.vol. 41:651–662.
27. Gaspar MA, Patounakis G, Schwartz D, Shepard K, Levicky R. Fabrication of a CMOS biochip for DNA diagnostics. *Abstracts of Papers of the Amer. Chem. Soc.* 2005 Aug.vol. 230:U1114–U1114.
28. Salama K, Eltoukhy H, Hassibi A, El Gamal A. Modeling and simulation of luminescence detection platforms. *Biosens. Bioelectron.* 2004 Jun.vol. 19:1377–1386. [PubMed: 15093208]
29. Patounakis G, Shepard KL, Levicky R. Active CMOS array sensor for time-resolved fluorescence detection. *IEEE J. Solid-State Circuits*. 2006 Nov.vol. 41:2521–2530.
30. Davenport M, Titus AH, Tehan EC, Tao ZY, Tang Y, Bukowski RM, Bright FV. Chemical sensing systems using xerogel-based sensor elements and CMOS photodetectors. *IEEE Sensors J.* 2004 Apr.vol. 4(no. 4):180–188.
31. Chodavarapu V, Bukowski RM, Titus AH, Cartwright AN, Bright FV. CMOS integrated luminescence oxygen multi-sensor system. *Electron. Lett.* 2007 Jun.vol. 43:688–689.
32. Faramarzpour N, Deen MJ, Shirani S, Fang Q, Liu LWC, de Souza Campos F, Swart JW. CMOS-based active pixel for low-light-level detection: Analysis and measurements. *IEEE Trans. Electron Devices*. 2007 Dec.vol. 54:3229–3237.

33. Faramarzpour N, El-Desouki MM, Deen MJ, Shirani S, Fang QY. CMOS photodetector systems for low-level light applications. *J. Mater. Sci.-Mater. Electron.* 2009 Jan.vol. 20:87–93.
34. Nakamura, J. *Imager Sensors and Signal Processing for Digital Still Cameras.* New York: Taylor & Francis Group; 2006.
35. Chodavarapu VP, Shubin DO, Bukowski RM, Titus AH, Cartwright AN, Bright FV. CMOS-based phase fluorometric oxygen sensor system. *IEEE Trans. Circuits and Syst. I (Regular Papers).* 2007 Jan.vol. 54:111–118.
36. Lin CT, Bottcher W, Chou M, Creutz C, Sutin N. Mechanism of quenching of emission of substituted polypyridineruthenium(II) complexes by iron(III), chromium(III), and europium(III) ions. *J. Amer. Chem. Soc.* 1976; vol. 98:6536–6544.
37. Cho EJ, Tao ZY, Tang Y, Tehan EC, Bright FV, Hicks WL, Gardella JA, Hard R. Tools to rapidly produce and screen biodegradable polymer and sol-gel-derived xerogel formulations. *Appl. Spectroscopy.* 2002 Nov.vol. 56:1385–1389.
38. Tang Y, Tehan EC, Tao ZY, Bright FV. Sol-gel-derived sensor materials that yield linear calibration plots, high sensitivity, and long-term stability. *Anal. Chem.* 2003; vol. 75:2407–2413. [PubMed: 12918984]

Biographies

Lei Yao (S'09) received the B.S. degree in applied physics from Science and Technology of China, Hefei, China, in 2004, and the M.Eng. degree in micro-electronics and solid-state electronics from Shanghai Institute of Microsystems and Information Technology, Chinese Academy of Sciences, Shanghai, China, in 2007. He is currently working towards the Ph.D. degree in electrical and computer engineering at McGill University, Montreal, QC, Canada.

His research interests are mixed signal VLSI design, microfluidic MEMS devices and system integration for biomedical sensor systems.

Ka Yi Yung received the B.S. degree in chemistry from Rochester Institute of Technology, Rochester, NY, in 2007. She is currently working towards the Ph.D. degree in chemistry at the State University of New York at Buffalo (UB), Amherst.

She is a Research Assistant in analytical chemistry and her major research interests include the development of chemical and biochemical sensor arrays based on sol-gel processing technology and ionic liquids.

Rifat Khan received the B.Eng degree in electrical and computer engineering in 2009 from McGill University, Montreal, Quebec, Canada.

He is currently a Research Assistant in the Sensor Microsystems Laboratory, Department of Electrical and Computer Engineering, McGill University. He is currently working on system integration for CMOS-based biochemical sensor system.

Vamsy P. Chodavarapu (S'04–M'06) received the B.Eng. degree in instrumentation engineering from Osmania University, Hyderabad, India, in 2001, and the M.S. and Ph.D. degrees in electrical engineering from University at Buffalo, The State University of New York, in 2003 and 2006, respectively.

In 2006, he joined the Department of Electrical and Computer Engineering, McGill University, Montreal, QC, Canada, as an Assistant Professor, where he directs the Sensor Microsystems Laboratory. His specific research interests are in the areas of CMOS sensor microsystems, biological/chemical sensors, mixed-signal VLSI design, nanomaterials, and MEMS/microfluidics. His research is funded by various government and private sources.

Dr. Chodavarapu is a member of SPIE.

Frank V. Bright received the B.S. degree from the University of Redlands, Redlands, CA, in 1982 and the Ph.D. degree from Oklahoma State University, Stillwater, in 1985.

He was a Postdoctoral Fellow at Indiana University from 1985 to 1987. He began his independent academic career as an Assistant Professor in the Department of Chemistry, University of Buffalo (UB), The State University of New York (SUNY), in 1987. He is currently a SUNY and UB Distinguished Professor of Chemistry, A. Conger Goodyear Chair in Chemistry, and Department Chairperson. He is the author/coauthor of more than 250 peer-reviewed publications and seven issued patents. His research centers on chemical sensors, antifouling materials, supercritical fluid science and technology, and chemical instrumentation.

Dr. Bright is a member of the American Chemical Society and the Society for Applied Spectroscopy. He has won numerous awards for research and teaching and he has served on numerous journal editorial boards.

Fig. 1.
Simplified block diagram of the CMOS imager-based multisensor system.

Fig. 2.
Schematic diagram of the CMOS Imager IC and typical waveform of key digital signals.

Fig. 3.
Microphotograph of the CMOS imager IC. The inset picture shows a single APS.

Fig. 4. Schematic of high-gain high-swing rail-to-rail folded-cascode op-amp used in integrator (W/L ratio of all unlabeled PMOS is $72/5.6$, W/L ratio of all unlabeled NMOS is $24/5.6$): (a) Bias circuit. (b) OPA core circuit.

Fig. 5. Simulation results of CMOS imager IC. The inset shows the ideal phototransistor model used for simulations.

Fig. 6. Nonoverlapping clock generator and control signal generator. Example of a nonoverlapping clock is also shown.

Fig. 7. Measured relationship between incident power and sampling time T_{sample} . Inset: Measured imager response with the sampling time, $T_{\text{sample}} = 500 \mu\text{s}$.

Fig. 8.
Oscilloscope view of CMOS imager output waveform ($T_{\text{sample}} = 1\text{ ms}$).

Fig. 9.
Microphotograph of the pin-printed oxygen sensor array.

Fig. 10.
Experimental setup for evaluating the multisensor imaging system.

Fig. 11.
Processed images from the CMOS IC with varying O_2 concentrations.

Fig. 12.
Measured sensitivity of different types of sensors in the array.

Fig. 13.
Measured Stern–Volmer response of the sensor array.

TABLE I

Description of Four Different Sensors

Sensor Row	Volume of sol solution (μL)	Volume of stock solution A (μL)	Volume of stock solution B (μL)
A	80	0	20
B	80	1.3	18.7
C	80	3.3	16.7
D	80	7.5	12.5