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A flexible organic reflectance oximeter array

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Transmission-mode pulse oximetry, the optical method for determining oxygen saturation in blood, is limited to tissues that can be transilluminated, such as the earlobes and the fingers. The existing sensor configuration provides only single-point measurements, lacking 2D oxygenation mapping capability. Here, we demonstrate a flexible and printed sensor array composed of organic light-emitting diodes and organic photodiodes, which senses reflected light from tissue to determine the oxygen saturation. We use the reflectance oximeter array beyond the conventional sensing locations. The sensor is implemented to measure oxygen saturation on the forehead with 1.1% mean error and to create 2D oxygenation maps of adult forearms under pressure-cuff–induced ischemia. In addition, we present mathematical models to determine oxygenation in the presence and absence of a pulsatile arterial blood signal. The mechanical flexibility, 2D oxygenation mapping capability, and the ability to place the sensor in various locations make the reflectance oximeter array promising for medical sensing applications such as monitoring of real-time chronic medical conditions as well as postsurgery recovery management of tissues, organs, and wounds.

flexible electronics | wearable sensors | bioelectronics | organic electronics | oximetry

Hemoglobin, a protein molecule in the blood, transports oxygen from the lungs to the body’s tissues. Oximeters determine oxygen saturation (\(\text{SO}_2\)) in tissues by optically quantifying the concentration of oxyhemoglobin (\(\text{HbO}_2\)) and deoxyhemoglobin (\(\text{Hb}\)) (1). Pulse oximetry, the most ubiquitous noninvasive method of oximetry, performs this ratiometric optical measurement on pulsatile arterial blood via photoplethysmography (PPG) at two different wavelengths (2). Pulse oximeters use optoelectronic sensors composed of light-emitting diodes (LEDs) and photodiodes (PDs) and operate at red and near-infrared (NIR) wavelengths, where the molar absorptivities of \(\text{HbO}_2\) and \(\text{Hb}\) are significantly different. While both transmitted and reflected light can be used for pulse oximetry, in transmission-mode pulse oximetry (\(S_p\text{O}_2\)), the LEDs shine through the tissue and the transmitted light is collected using the PD on the opposite side (SI Appendix, Fig. S1)—this restricts sensing locations to only tissues that can be transilluminated, such as the earlobes and the fingers, and the feet for neonates. On the contrary, reflection-mode pulse oximetry (\(S_r\text{O}_2\)) uses LEDs and PDs on the same side of the tissue (SI Appendix, Fig. S1), which allows for diverse sensing locations such as the forehead, forearm, abdomen, and leg. Additionally, \(S_r\text{O}_2\) provides 2D oxygenation mapping capability with an array of sensors, whereas only single-point measurements can be performed with \(S_p\text{O}_2\).

Recent progress in flexible and stretchable sensors has made them extremely promising for medical sensing and diagnostics because they enhance the signal-to-noise ratio (SNR) by establishing a conformal sensor–skin interface (3–11). Consequently, novel flexible sensors using organic and inorganic optoelectronics for transmission and reflection-mode pulse oximetry show a higher SNR due to a reduction in ambient noise (12–18).

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Conflict of interest statement: A provisional patent application has been filed based on the technology described in this work.

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Significance

The optical method to determine oxygen saturation in blood is limited to only tissues that can be transilluminated. The status quo provides a single-point measurement and lacks 2D oxygenation mapping capability. We use organic printed optoelectronics in a flexible array configuration that senses reflected light from tissue. Our reflectance oximeter is used beyond conventional sensing locations and accurately measures oxygen saturation on the forehead. In a full system implementation, coupled with a mathematical model, we create 2D oxygenation maps of adult forearms under pressure-cuff–induced ischemia. Our skin-like flexible sensor system has the potential to transform oxygenation monitoring of tissues, wounds, skin grafts, and transplanted organs.
Results

Analytical Models for Reflectance Oximetry. Oximeters use the property that the molar extinction coefficients of Hb and HbO₂ vary appreciably over the visible and NIR spectrum. If two regions in the spectrum are chosen so that in one region, Hb has a higher absorptivity than HbO₂, and in the other region, Hb has a lower absorptivity than HbO₂, a ratiometric measurement can be performed to obtain the concentration of HbO₂ and Hb. The oxygen saturation, SO₂ is the concentration of HbO₂ divided by the sum of the concentrations of HbO₂ (C_HbO₂) and Hb (C_Hb): SO₂ = C_HbO₂ / (C_HbO₂ + C_Hb). In Fig. 1D, three regions are shown: (i) green (ε_Hb/ε_HbO₂ < 2), (ii) red (ε_Hb/ε_HbO₂ > 6), and (iii) NIR (ε_Hb/ε_HbO₂ < 3). Therefore, the combinations of red and green or red and NIR can be used for oximetry because of the contrast in molar extinction coefficients. Since the PPG signal magnitude for NIR is higher than the visible spectrum because light attenuation in tissue for NIR is much less than the visible spectrum, we use red and NIR OLEDs. In addition, the optical flux requirement for oximetry is less stringent for NIR than for visible colors—the NIR OLEDs used in the ROA provide 0.2 mW of flux, compared with the 0.9 mW of flux of the red OLEDs at the operating condition of 10 mA cm⁻² (SI Appendix, Fig. S3).

The operation of noninvasive reflectance oximetry can be grouped into two modes: (i) reflection-mode pulse oximetry...
(SpO₂), when a pulsatile PPG signal is present, and (ii) reflectance oximetry, when a pulsatile PPG signal is absent. If a pulsatile PPG signal is present, a modified Beer–Lambert's law can be used to model the light propagation in tissue as shown in Fig. 1B and Eq. 1.

\[
I(\lambda) = I_0(\lambda)e^{-\mu_a(\lambda) d \text{DPF}(\lambda)},
\]

where \(I(\lambda)\) is the measured diffused reflected light intensity, \(I_0(\lambda)\) is the incident light intensity, \(\mu_a(\lambda)\) is the absorption coefficient of the sensed tissue, \(d\) is the distance between the light emitter and detector, and DPF(\(\lambda\)) is the differential pathlength factor (DPF), which accounts for the multiple scattering of light in tissue. The light attenuation in pulsatile arterial blood can be used to calculate \(S_pO_2\) in accordance with Beer–Lambert's law and an empirical correction according to Eq. 2 (the complete derivation is provided in SI Appendix):

\[
S_pO_2(R_{os}) = \frac{\varepsilon_{\lambda_1,Hb} - \varepsilon_{\lambda_2,Hb}R_{os}}{(\varepsilon_{\lambda_1,Hb,O2} - \varepsilon_{\lambda_2,Hb,O2} - \varepsilon_{\lambda_2,Hb})R_{os}}.
\]

Here, \(\varepsilon_{\lambda,Hb,O2}\) and \(\varepsilon_{\lambda,Hb}\) are the molar extinction coefficients of oxyhemoglobin and deoxyhemoglobin at each wavelength. \(R_{os} = \frac{\text{DPF}(\lambda)}{\text{DPF}(0)}\) where \(R_{os} = \frac{\Delta I_{\lambda_1,Hb,O2}}{\Delta I_{\lambda_2,Hb,O2}}\) is the ratio of pulsatile (ac) to stationary (dc) signals at the two wavelengths.

In the case of low perfusion or in the absence of a pulsatile arterial blood signal, pulse oximetry in both transmission and reflection modes cannot be performed. In these scenarios, Eq. 1 can be rewritten to measure the time-averaging light intensity attenuation, \(\Delta I(\lambda)\) in blood and tissue. Here, \(\Delta \mu_o\) expresses the change in absorption during the measurement:

\[
\Delta I(\lambda) = I_0(\lambda)e^{-\Delta \mu_o(\lambda) d \text{DPF}(\lambda)}.
\]

\(\Delta \mu_o(\lambda)\) can be represented as the sum of the molar extinction coefficients multiplied by the concentrations of \(HbO_2\) and \(Hb\):

\[
\Delta \mu_o(\lambda) = \varepsilon_{HbO_2}(\lambda) \cdot \Delta C_{Hb,O2} + \varepsilon_{Hb}(\lambda) \cdot \Delta C_{Hb}.
\]

Since there are two wavelength channels, a system of linear equations can be established using Eqs. 3 and 4:

\[
\begin{bmatrix}
\varepsilon_{HbO_2}(\lambda_1) & \varepsilon_{Hb}(\lambda_1) \\
\varepsilon_{HbO_2}(\lambda_2) & \varepsilon_{Hb}(\lambda_2)
\end{bmatrix}
\begin{bmatrix}
\Delta C_{Hb,O2} \\
\Delta C_{Hb}
\end{bmatrix}
= \begin{bmatrix}
\ln \frac{I_{\text{os}}(\lambda_1)}{I_{\text{os}}(\lambda_2)} \\
\ln \frac{I_{\text{os}}(\lambda_1)}{I_{\text{os}}(\lambda_2)}
\end{bmatrix}
\]

In Eq. 5, the molar extinction coefficients and DPF(\(\lambda\)) can be obtained from the literature (1, 24). Since change in the concentration of \(HbO_2\) (\(\Delta C_{Hb,O2}\)) and \(Hb\) (\(\Delta C_{Hb}\)) can be calculated, the change in oxygen saturation (\(\Delta SO_2\)) can be determined for the transient measurement. The complete derivation of Eq. 5 is provided in SI Appendix.

Reflectance Oximeter Design and Placement on the Body. Emitter–detector spacing (\(d\)) is an important design parameter for reflectance oximetry. To find the optimal \(d\), we use a reflection-mode sensor board and measured the effect of \(d\) on PPG ac and dc signals at the eight locations on the body as depicted in SI Appendix, Fig. S4 A and B. The schematic of the sensor, containing three rings of four PDs spaced at 0.5 cm, 0.8 cm, and 1.1 cm away from the red and NIR LEDs at the center, is shown in SI Appendix, Fig. S4C. Both ac and dc signal magnitude drops exponentially with increasing \(d\). SI Appendix, Fig. S4 D and E shows ac and dc signals for \(d = 0.5\) cm, 0.8 cm, and 1.1 cm recorded on the wrist. When placed at \(d < 0.5\) cm, the dc signal saturates the PD. This issue can be mitigated by putting an optical barrier between the LED and the PD to reduce direct coupling of light from the LED to the PD. While \(d = 0.5\) cm provided us the best SNR, \(d\) can be different for other sensor designs. For a single-pixel reflectance sensor, a minimum optical flux should be maintained to resolve the pulsatile PPG signal. We measured that this minimum flux is \(\approx 0.2\) mW for NIR and \(\approx 0.6\) mW for red light. Once the minimum optical flux is ensured, the emitter–detector spacing (\(d\)) can be reduced so that the light detected from the arteries (signal) is distinguishable from the light scattered by the skin surface (noise). Overall, the optical flux output of the LEDs, external quantum efficiency (EQE) of the PD, and active area of the LEDs and the PD influence the optimum \(d\) for a reflection-mode sensor. All of the mentioned strategies for reducing \(d\) can be used for scaling down the pixel-to-pixel spacing of the ROA, which will reduce the array size while providing an adequate signal level to quantify the light absorption in blood and tissue.

An approach similar to obtaining the optimal \(d\) is used to find the optimal sensing location for \(SpO_2\)—we place the reflection-mode sensor at eight different locations on the body as depicted in SI Appendix, Fig. S4 A and B. SI Appendix, Fig. S4A shows the pulsatile (ac) signal magnitude for red (\(Red_o\)) and NIR (\(NIR_{ac}\)) channels with an emitter–detector spacing, \(d = 0.5\) cm. A high ac signal and a low dc signal are desirable for PPG measurements. The forehead provides the strongest peak-peak ac signal current, 20 nA for red and 60 nA for NIR, making it the most suitable location for \(SpO_2\). The signal strength drops roughly by half on the wrists. Although we observed a clear degradation of the ac signal on the ribcage and the legs, heart rate and oxygenation values could be extracted from the measured signal. Similar to that for the ac signal, the forehead provides the highest dc signal, while the ribcage demonstrates the lowest dc signal magnitude. SI Appendix, Fig. S5 provides the full dataset of the ac and dc measurements at the eight sensing locations for five subjects. The reflectance sensor is mounted on the skin, using an adhesive foam dressing. The sensor mounting photos are shown in SI Appendix, Fig. S6.

OLED and OPD Array Fabrication and Characterization. We printed the organic optoelectronics of the ROA on separate substrates and then assembled them to form the sensor array. With 0.7 \(\times\) 0.7 cm active area for both OLEDs and OPDs and 0.5 cm spacing between the OLEDs and OPDs, the dimension of the complete ROA is 4.3 cm in both length and width. The OLED arrays are fabricated on top of polyethylene naphthalate (PEN) substrates with patterned indium tin oxide (ITO) for contacts. A surface energy patterning (SEP) step is then performed that creates hydrophilic regions where poly(3,4-ethylenedioxythiophene)-poly(styrenesulfonate) (PEDOT:PSS) is blade coated, which is discussed in detail by Han et al. (15, 25) (Fig. 2A, Left). The interlayer and the emission layer are deposited using subsequent blade-coating steps (Fig. 2B, Left). Then, the dielectric and the silver traces are printed using a screen printer (Fig. 2C, Left). The purpose of printing the dielectric is to prevent shorts between the underlying ITO strips and the silver traces. Finally, thermal evaporation is used to deposit calcium/aluminum to finish the fabrication of OLED arrays (Fig. 2D, Left). Each OLED pixel is encapsulated with UV curable epoxy and a plastic film. The OLED device stack is shown in Fig. 2G. The same process steps apply for both red and NIR OLEDs; only the active materials are different.

The OPD array is fabricated on a planarized PEN substrate. A PEDOT:PSS anode is blade coated using the SEP technique as shown in Fig. 2A, Right. The SEP process for OPDs is
previously described by Pierre et al. (26, 27). A patterned anode is necessary because, without patterning, a large parasitic capacitance is formed between the PEDOT:PSS layer and the body, which obscures the signal in noise. The active layer is then blade coated (Fig. 2B, Right). Next, silver traces are screen printed to connect the anodes and cathodes of each pixel to external circuitry as shown in Fig. 2C, Right. Finally, an aluminum cathode is evaporated to complete the device stack, which is shown in Fig. 2H.

The OPD and OLED arrays are shown in Fig. 3A and B, respectively. The OPD array comprises eight OPD pixels, where each OPD row contains two OPD pixels. Brown markers from darker to lighter shades are used to label rows 1–4 of the OPD array. The same markers are used to present the performance characteristics of the OPD pixels. As for the 2 × 2 red and NIR OLED arrays, rows 1 and 3 contain the four red OLED pixels, and rows 2 and 4 contain the four NIR OLED pixels. The ROA is formed by stacking the OLED and OPD arrays. The arrays are assembled such that emitter–detector spacing of 0.5 cm is maintained.

The performance parameters of the OPD array are shown in Fig. 3C–E. The shade of brown lines indicates the row position of the pixels in the array as shown in Fig. 3A. An average EQE of 30% is observed across the absorption spectrum (Fig. 3D) with dark currents of a few nanoamperes per square centimeter (Fig. 3C). The cutoff frequency is measured at over 5 kHz for OPDs as shown in Fig. 3E. Since the operation frequency of the pulse oximeters is generally less than 1 kHz, this bandwidth is sufficient for oximetry. The linear dynamic response of the OPDs is shown in SI Appendix, Fig. S7.

The OLEDs show turn-on voltages at around 3 V as designated in the J-V characteristics in Fig. 3F. The OLEDs are operated at 10 mA cm−2 for oximetry, where the red OLEDs provide 0.9 mW of flux, while the NIR OLEDs provide 0.2 mW of flux. The EQE values at operating conditions are ~8–10% for red OLEDs and ~2–3% for NIR OLEDs (Fig. 3G). The OLEDs demonstrate a change in performance parameters, depending on the row position due to the decrease in active layer thickness in the blade-coating direction; this variability can be mitigated by continuously feeding ink in front of the blade coater (15, 28). The variability in the OLED and OPD performance can be accommodated by taking a calibration measurement before using the array for oximetry. The emission spectrum of the OLEDs is shown in Fig. 3H, where the red OLED has a peak emission at 612 nm and the NIR OLED has a peak emission at 725 nm.

**System Setup and Single-Pixel Reflection-Mode Pulse Oximetry.** The full system implementation requires addressing individual pixels of the oximeter. Therefore, the hardware and software for the ROA are designed to support both single-pixel and array
measurements (Fig. 4A). The printed ROA is interfaced with the control electronics using flexible flat cable (FFC) connectors. Each pixel of the ROA is composed of one red and one NIR OLED and two OPDs. Signals from the red and NIR channels are read out sequentially using the two OPDs, and the average of the OPDs is used for signal processing. Using this format, the 4 × 4 device (OLEDs and OPDs) array provides 3 × 3 readout pixels. The pixels are selected using analog switches. An analog front end (AFE) sequentially drives the OLEDs and reads out the OPD signal. The AFE is controlled by a microcontroller. Software control of the AFE allows flexibility in choosing OLED driving parameters and also gives access to the variable OPD gain circuitry. A photograph of the control electronics is shown in SI Appendix, Fig. S8, and the photographs of the software’s graphical user interface (GUI) are shown in SI Appendix, Fig. S9.

To test the reflectance oximeter in the single-pixel mode, we used a setup where oxygenation of a volunteer can be changed by varying the oxygen concentration of the inhaled air (Fig. 4B). An altitude simulator is used to change the oxygen concentration of the air the volunteer breathes in via a facemask. Depending on the oxygen concentration of the air, the volunteer’s oxygen saturation is measured. This setup allows for the calibration of the oximeter in different environments and under varying oxygen concentrations.
oxygenation changes. This change in oxygenation is then picked up by a commercial finger probe sensor and using the reflection-mode sensor on the forehead. Calculated oxygen saturation using the commercial probe (SpO₂) and the reflectance probe (SpO₂) is shown in Fig. 4 C and D. We varied the inhaled oxygen concentration (O₂%) from 21% to 15% over a period of 8 min. During the first 30 s, a baseline oxygen concentration of 21% was set and then reduced to 17.5% at t = 30 s; after keeping O₂% at 17.5% for 120 s, O₂% was further reduced to 15% at t = 150 s and was kept at that level for 150 s. Then O₂% was brought back to the baseline of 21%.

For the transmission-mode probe, oxygen saturation (SpO₂) drops from 96% to 90.5% and then comes back up to 94.5%. For the reflection-mode probe on the forehead, oxygen saturation (SpO₂) changes from 98% to 90.4% and then comes back up to 93.5%. We observed 1.1% mean error between SpO₂ and SpO₂ over the period of 8 min. The PPG signals for both transmission and reflection-mode probes at 240 s < t < 245 s are shown in Fig. 4 E and F. PPG signal peaks and calculated heart rate from the PPG peaks show almost identical results for both SpO₂ and SpO₂. Here, an error of 0.85% is seen between SpO₂ and SpO₂, which falls

---

**Fig. 4.** System design for reflectance oximetry and single-pixel reflection-mode pulse oximetry (SpO₂) results. (A) Reflectance oximeter system design. Each pixel of the ROA (one red and one NIR OLED and two OPDs) is connected to an AFE using analog switches, for both single-pixel and array operation. The AFE drives the OLEDs and reads out the OPD signal. The AFE is controlled using an Arduino Due microcontroller. The data are then collected using a universal serial bus (USB) interface and processed using custom software. (B) Setup for changing oxygen saturation of human volunteers. An altitude simulator varies the oxygen content of the air the volunteer breathes in via a facemask. The SpO₂ is recorded using a commercial probe on the finger and the reflectance oximeter on the forehead. (C and D) Results from the commercial transmission-mode finger probe oximeter (SpO₂) and the reflectance oximeter (SpO₂), where the oxygen concentration is changed from 21% to 15%. Shown are the oxygen concentration of the air (C and D, Top, blue trace) and calculated oxygen saturation using SpO₂ and SpO₂ (C and D, Bottom, purple trace). (E and F) Zoomed-in data for the SpO₂ in C and SpO₂ in D during 240 s < t < 245 s show the red channel, NIR channel, PPG peaks, heart rate, R₂, and SpO₂.
within the 1% to 2% error margin that is inherent to pulse oximetry.

The pulse arrival times at the forehead and the fingers are different; the delay is on the order of 50 ms (29), which may slightly affect the pulse oxygenation calculations. Therefore, for a more direct comparison between the transmission- and reflection-mode pulse oximetry, we collected pulse oximetry data in both transmission and reflection mode from the fingers of the same hand as shown in SI Appendix, Fig. S10. In this experiment, the printed reflectance probe is placed under one finger and the commercial transmission-mode finger probe is worn on another finger. The commercial and reflectance finger probes provide almost identical SpO₂ variation with different concentrations of the inhaled oxygen. We observed a mean error of 0.41% between the commercial transmission-mode finger probe and the printed reflectance oximeter.

To investigate the temperature effects of the reflectance sensor, we operated the OLEDs of the device at different drive conditions and recorded the corresponding temperatures of the sensor on a volunteer’s forearm (SI Appendix, Fig. S11). We observed a negligible change in the temperature—the sensor temperature remained within 32 ± 0.5 °C under the different OLED drive voltages from 0 V to 9 V. Moreover, to study the effect of an external pressure on the reflectance sensor, we collected reflection-mode signal with and without a 0.7-kPa external pressure (SI Appendix, Fig. S12A). With the external pressure, the signal baseline changes for both red and NIR channels, and the PPG ac signal magnitude improves for the red channel (SI Appendix, Fig. S12B) compared with the PPG ac signal without an external pressure (SI Appendix, Fig. S12C). However, for pulse oximetry calculations, the ratio of pulsatile (ac) to stationary (dc) signals at the two wavelengths is used, which remains almost the same: R’(t) = 0.65 with external pressure, and R’(0) = 0.67 without external pressure.

**In Vivo 2D Oxygen Saturation Monitoring.** The pulse oximetry model is applicable when there is a pulsatile arterial blood signal. In the absence of a pulsatile arterial blood signal, we use the modified model (Eq. 5 and SI Appendix) to monitor local changes in tissue oxygenation of a volunteer’s arm under normal and ischemic conditions. By restricting blood supply to the arm with a pressure cuff, we induce temporary ischemia to the arm by inflating the pressure cuff to 50 mmHg over the systolic pressure. We use the ROA to monitor the change in oxygen saturation (ΔSO₂) under normal conditions and under pressure-cuff–induced ischemia with the ROA. The measurement is shown in Fig. 5A, where the ROA is used to measure ΔSO₂ on the forearm, while the pressure cuff is used to control blood supply to the arm, subsequently changing ΔSO₂ of the sensed tissue. Under normal conditions, there is a pulsatile arterial blood signal (SI Appendix, Fig. S13), which can be used with the pulse oximetry model. However, when the blood supply is restricted, only reflectance oximetry can be performed to measure ΔSO₂.

The 4 × 4 OLED and OPD devices provide 3 × 3 oximeter pixels. These pixels are indexed as pixels 1–9, (Px1–Px9) and are shown in Fig. 5B. A raster scan from Px1 to Px9 is used to collect data from the tissue. We used raster scanning speeds ranging from 100 ms-Px⁻¹ to 1 s-Px⁻¹. Within this range, we did not observe analog-to-digital converter (ADC) channel leaking. For each pixel, samples are collected at 500 Hz, and then the collected data are averaged. At a sampling rate faster than 50 ms-Px⁻¹, we observed ADC channel leaking. The limiting factors to achieve a faster scanning speed are the 3-dB cutoff of the OPDs, the resistance and capacitance associated with each channel, and the settling time of the analog switches. After collecting data from all 9 pixels, 2D contour maps of red and NIR channels and ΔSO₂ are created. For the in vivo 2D oxygen saturation monitoring test, data during the first 30 s are collected under normal conditions and are considered the baseline. The pressure cuff is then used to induce ischemia; therefore the signal amplitude in the red and NIR signal channels gradually decreases. Once the pressure is released, the signals overshoot, going over the baseline (Fig. 5C). Fig. 5D shows the 2D maps of red and NIR signal channels and ΔSO₂ during the test under normal conditions (t = 0 s), under ischemia (t = 60, 120 s), and after releasing the pressure cuff (t = 180, 240, 300 s). Since this is a transient measurement, ΔSO₂ remains at the baseline (ΔSO₂ = 0%) at t = 0 s, under ischemia ΔSO₂ drops to −9.3% at t = 150 s, and after releasing the pressure cuff ΔSO₂ increases to +8.4% at t = 180 s and comes back to the baseline +0.7% at t = 300 s. The 2D contour maps at every 30-s interval are provided in SI Appendix, Fig. S14. We also monitored how different durations of ischemia affect ΔSO₂. SI Appendix, Fig. S15 provides ΔSO₂ monitoring for 1 min, 2 min, and 2.5 min of ischemia, and ΔSO₂ drops to −6%, −9.5%, and −11.3%, respectively, during these experiments. The results obtained in these tests agree with the studies reported in the literature on pressure-cuff–induced ischemia (20, 24, 30, 31).

In the 2D oxygenation mapping experiments, we monitored tissue oxygenation of the forearm with and without pressure-cuff–induced ischemia. When blood supply to the arm is occluded using the pressure cuff, oxygenated blood cannot circulate to the forearm, which results in a drop in tissue oxygenation. We recorded this change in oxygenation using the ROA. With the status quo, i.e., transmission-mode pulse oximetry, this change in oxygenation cannot be observed, because when blood circulation is cut off, the pulsatile arterial blood signal disappears, which is essential to calculate the pulse oxygenation using transmission-mode pulse oximetry. The ROA can measure the change in SO₂. Moreover, with the 2D mapping capability, the ROA can monitor oxygenation of an area rather than a single point, which is promising for monitoring oxygenation of tissues, wounds, and newly transplanted organs. In addition, the ROA can be integrated to a multimodal near-infrared spectroscopy (NIRS) system, where the ROA is interfaced to printed electromyography (EMG) or electrocardiography (ECG) electrodes on a flexible substrate to provide a lightweight, comfortable, and wearable sensor platform for muscle assessment during a person’s normal activities and exercise (32–34).

**Discussion**

Existing techniques for measuring oxygen concentration in blood heavily rely on noninvasive transmission-mode pulse oximetry (SpO₂), which presents two fundamental limitations: (i) Sensing locations are limited to only tissues that can be transilluminated and (ii) only single-point measurements can be performed with SpO₂ due to the sensor configuration. Here, we presented a flexible and printed electronic system realized by printing and integrating arrays of organic optoelectronics for measuring oxygen saturation in the reflection mode. Two different modes of oximeter operations are discussed: (i) reflection-mode pulse oximetry (SpO₂), when pulsatile PPG signal is present, and (ii) reflectance oximetry, when pulsatile PPG signal is absent. Additionally, we explored the sensor design and placement of the sensor on the body. The forehead provided the strongest pulsatile signal. By using the reflectance oximeter, we monitored oxygen saturation of a volunteer on the forehead and successfully measured SpO₂ with a mean error of 1.1%. The use of an altitude simulator to change the oxygen concentration verified the efficacy of the reflectance oximeter under both hypoxia and normal conditions. Finally, in the case of a medical shock, low blood perfusion, or locations on the body with a low PPG signal, we demonstrate a method to determine ΔSO₂ in the absence of a pulsatile.
**Fig. 5.** In vivo 2D oxygen saturation monitoring with the ROA. (A) The ROA is placed on a volunteer's forearm to monitor the change in oxygen saturation ($\Delta SO_2$). Blood supply to the forearm is controlled by a pressure cuff. The $4 \times 4$ devices of the ROA provide $3 \times 3$ oximeter pixels. (B) Oximeter pixel switching during the array operation. Each pixel is composed of one red and one NIR OLED and two OPDs. A raster scan from pixel 1 (Px1) to pixel 9 (Px9) is used to collect data from the tissue. (C) $\Delta SO_2$ for pressure-cuff–induced ischemia for a recording of 300 s. Red, NIR, and $\Delta SO_2$ data are shown as red, black, and purple dotted lines (dotted lines represent the means of the nine oximeter pixels, and error bars represent the SD of the data). Using the pressure cuff, blood supply to the forearm is occluded and restored. In the first 30 s, a baseline reading with no ischemia is taken. The pressure cuff is then inflated to 50 mmHg over the systolic pressure at $t = 30 < t < 150$ s and released at $t = 150$ s. $\Delta SO_2$ varies from 0% under normal conditions to $-9.3\%$ ($t = 150$ s) under ischemia and to $+8.4\%$ ($t = 180$ s) immediately after releasing the pressure cuff. (D) Two-dimensional contour maps of red, NIR, and $\Delta SO_2$ under normal conditions ($t = 0$ s), under ischemia ($t = 60, 120$ s), and after releasing the pressure cuff ($t = 180, 240, 300$ s).
The OLED arrays were fabricated and characterized in the OLED Arrays section. The OPD array was fabricated and characterized in the OPD Array section. The fabrication and characterization of the OLED arrays were performed using blade coating techniques. The substrate was first plasma treated in a nitrogen-filled chamber overnight. The substrate was patterned, exposing the active area of the OLEDs. Then ink was blade coated with photoresist to selectively etch off the hydrophobic layer. The substrate was then coated with a 100 nm OLED emissive layer, 10 nm calcium, and 100 nm aluminum. Each pixel was encapsulated by face sealing using a UV curable epoxy (Delo 100 nm). After the dielectric and the silver traces were screen printed, the samples were transferred to a thermal evaporator in a glovebox to evaporate calcium (99.5%; Strem Chemicals) and aluminum (99.999%; ACI Alloys) to finish the OLED stack. The final device stack consisted of 70 nm PEDOT:PSS, 100 nm OLED emissive layer, 10 nm calcium, and 100 nm aluminum. Each pixel was encapsulated by face sealing using a UV curable epoxy (Delo KatioBond LP612) and plastic film (PQA1) on top. After all of the layers were blade coated, the samples were taken out of the glovebox for screen printing. The reflectance oximetry experiments performed on human subjects were carried out with informed consent under the approval of the University of California, Berkeley Institutional Review Board, protocol ID 2014-03-6081.

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