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Y. A. W. de Kort, W. A. IJsselsteijn, I. M. L. C. Vogels,
M. P. J. Aarts, A. D. Tenner, & K. C. H. J. Smolders (Eds.)

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Volume Editors

Yvonne de Kort, PhD
Wijnand IJsselsteijn, PhD
Karin Smolders, MSc
Eindhoven University of Technology
IE&IS, Human-Technology Interaction
PO Box 513, 5600 MB Eindhoven, The Netherlands
E-mail: {y.a.w.d.kort, w.a.ijsselsteijn, k.c.h.j.smolders}@tue.nl

Ingrid Vogels, PhD
Visual Experiences Group
Philips Research
High Tech Campus 34, WB 3.029
5656 AE Eindhoven, The Netherlands
E-mail: ingrid.m.vogels@philips.com

Mariëlle Aarts, MSc
Eindhoven University of Technology
Department of Architecture Building and Planning
PO Box 513, VRT 6.34
5600 MB Eindhoven, The Netherlands
E-mail: M.P.J.Aarts@tue.nl

Ariadne Tenner, PhD
Independent consultant
Veldhoven, The Netherlands
E-mail: ariadne.tenner@onsmail.nl

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Flexible Light Sources for Health and Well-being

Margreet de Kok, Herman Schoo,
Marc Koetse, Ton van Mol
Holst Centre / TNO
Hightech Campus 31
5605 KN Eindhoven
The Netherlands

ABSTRACT
The availability of flexible and conformable light sources will enable applications that demand minimal distance between light source and body, such as therapeutic use of light and monitoring of personal health by wearable optical sensors. Organic light emitting diodes are large area, low voltage, thin light sources which can be processed on flexible substrates like foil in Roll-to-Roll process technology and are therefore potentially low cost. All of these characteristics render OLED highly suitable for applications on the body. The Holst Centre is developing Systems in Foil, both OLEDs and sensors. Lighting applications in health and well-being will benefit from both functionalities.

Keywords
OLED, foil, sensors, light source

INTRODUCTION
In recent years it has become clear that there are many positive effects of light on health and well-being of people. Blue light for example can be used for phototherapy of dermatological diseases like psoriasis and neonatal jaundice [1] or even skin rejuvenation [2, 3]. Acne [4] and seasonal affective disorder (SAD) [5] can be successfully treated with light. Pain can be relieved with light treatment [6].

For extra-clinical treatment and home-use wearable or at least conformable light sources are optimal. Not only the comfort of patients will be increased, also a serious cost reduction will be possible with these conformable light sources as the burden on clinical personnel and space will become less for cases which do not need continuous supervision.

With these wearable light sources, phototreatment can not only be used in healthcare, but mainly in well-being applications such as pain relief, anti-wrinkle measures and perfusion enhancement.

Integration of inorganic LEDs in textile is developed at the moment. Organic LEDs are an interesting alternative light source, because of their intrinsic large area and the possibility to be produced in a cost effective Roll-to-Roll processing. Furthermore they operate at a low voltage and will be energy efficient. When in contact to the skin, permeability to ensure regulated humidity at the skin-device interface and sterile contact areas are prerequisites and to be taken into account when developing these systems for health and well-being.

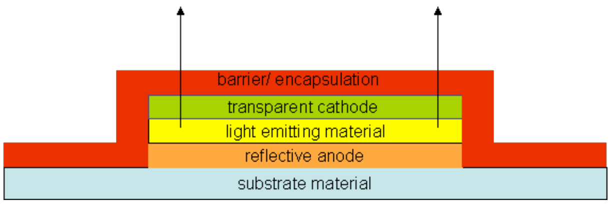
Light can also be of use for well-being and health in optical sensors. The condition of the body in blood perfusion, saturation, skin tone, etc. can be determined when light sources are combined with photodetectors. Both functionalities can be achieved in foil based structures with organic active materials.

These conformable light sources with integrated sensors are yet to be developed for commercialisation. The research at Holst Centre is focusing on enabling technology for these applications. In an open innovation environment, together with industrial and academic partners We develop technology platforms for autonomous wireless transducer solutions and systems-in-foil. The Systems-in-Foil Program Line targets to develop new device architectures, technologies and production processes for foil based electronic devices that will revolutionize the electronics industry. It will enable new ultra-light, ultra-thin, flexible, easy-to-wear electronic products such as lighting and signage devices, reusable and disposable sensor devices, foldable solar and battery panels and displays. The research addresses batch-wise and web-based processing (Roll-to-Roll), encompassing processes like printing, vacuum deposition, lithography, lamination and interconnection, which will enable the manufacturing of these devices in large sizes and quantities at low costs.

We will hereafter highlight the components that can be realised in foil and which are of most interest in light for health and well-being: organic light emitting diodes, organic photo diode, and the assembled sensors.

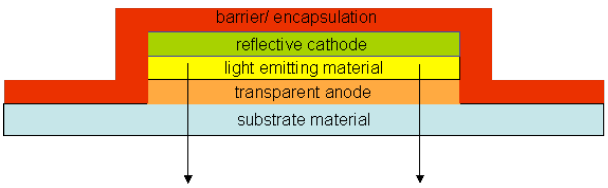
Organic Light-emitting diodes
An organic light-emitting diode (OLED) consists of an active material, sandwiched between two electrodes. At least one of the electrodes must be transparent. By applying a voltage, charge carriers are injected into the light-emitting material. Upon recombination excitons are formed and by subsequent radiative decay photons are emitted from the device. With a transparent cathode a top emissive OLED is formed, whereas a reflective cathode with transparent anode results in a bottom emissive OLED is the result (see figure 1). With two transparent electrodes a transparent OLED can be constructed. By this concept light sources

can be realised which are transparent in the off state, i.e. they can be integrated imperceptibly. Transparent windows generating light in the evening is one example of an envisioned application.



a) top emissive OLED

b)



b) bottom emissive OLED

Figure 1 OLED cross-section

In the foil based OLED program of Holst Centre, challenges like design-layout are studied. Flexibility, encapsulation and related lifetime are being optimised.

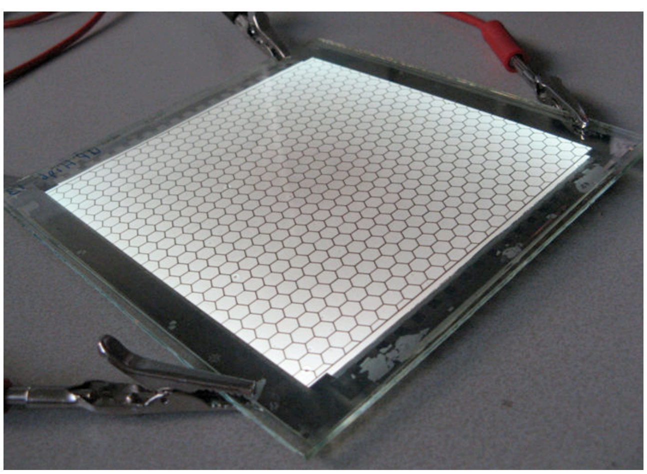


Figure 2 Bottom emissive ITO-less OLED (144 cm²) with inkjet printed metal grid

Recent advances in the Holst Centre include improved light homogeneity and increased reliability. In case a transparent anode is applied, a transparent conductive oxide (TCO) like indium tin oxide (ITO) can be used. ITO is however relatively expensive and shows only limited compatibility with roll to roll processes. Such high speed, low cost processing is considered to be required to meet the low cost per area for these light sources. Moreover in order to fully exploit the large area, the TCO must be conductive enough to ensure a homogeneous current distribution over the complete area. At Holst Centre a combination of printed Ag shunt lines and PEDOT:PSS, (poly (3,4-ethylene dioxy thiophene) : poly (styrene sulphonic acid), was recently presented to be a viable alternative[7].

Because of the reactive materials used in OLED devices encapsulation is key to ensure long lifetimes and reliable systems. If water is allowed to come into contact with the cathode, it is oxidised and the electron injection blocked which becomes visible as dark areas in the OLED. This can be prevented by thin film barrier technology. Holst Centre has developed a barrier with a water vapour transmission rate through the barrier well below 10^{-5} g/m^{2}day under ambient conditions without visible defects in a Ca-mirror test for 67 days. These barriers are capable of bending to radii of 20 mm which allows roll to roll processing [8].

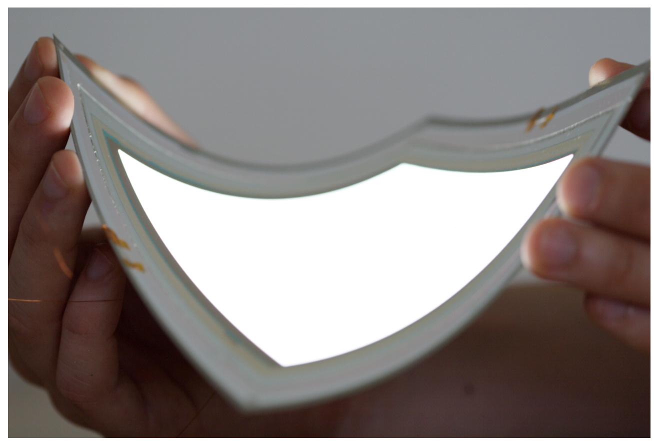


Figure 3 Top emissive OLED (100 cm^2) on metal foil thin film encapsulated

Nowadays energy consumption is a very important topic and therefore the increase of efficiency of OLEDs is given a lot of attention. Reported record value for white light by Konica Minolta is 64 lm/W at 1000 cd/m^2 and a lifetime (50 % luminance decrease) of 10.000 hrs. Kido et al. also reported similar high power efficiency for white OLEDs: 63 lm/W and 64 cd/A at a luminance of 100 cd/m^2 at the MRS meeting in 2006. For blue light emitting devices, record value is 50 lm/W [10], for green OLEDs 130 lm/W [9].

Small molecule OLEDs show currently higher efficiencies and lifetimes than polymer OLEDs. However there is also progress in the latter field: a luminous efficacy of 25 lm/W, 39 cd/A was reported for a white OLED based on a host polymer with phosphorescent dye [11]. A fluorescent, blended polymer system showed 16 lm/W and an external quantum efficiency of 6 % [12]. For roll to roll process technology solution processing has substantial advantages and solution processing of small molecules is therefore gaining interest.

Current challenges for OLEDs on foil lie in increasing efficacy, flexibility, lifetime and reliability.

For on the body applications a new generation of OLEDs are needed to adjust to the shape of the underlying person.

The next generation of conformable OLEDs demand to be truly stretchable and not only bendable. This will cost considerable efforts. Up till now several stretchable electronic circuits have been published including conductive wiring into a stretchable matrix of PDMS or thermoplastic polyurethane [13-15]. The method to render the light generating area stretchable is still to be explored.

Sensors in Foil

The system in foil approach as taken by the Holst Centre will allow to add more functionalities like sensors to the light generating foil. With sensors, feed-back of the effectiveness of the phototherapy can be derived and active control of the phototherapy will become possible. Sensor based functionalities might be focusing on the perfusion of tissue, saturation of the tissue or biological processes associated with healing of f.e. burn wounds. Abnormalities in the healing process, like infection, should at best be detected at a very early stage allowing for the proper action to be taken to minimize negative effects and promote the healing process. These biological processes can be detected by several physical parameters: colour of the skin (e.g. red for infection), a temperature or chemical substances in the wound fluids as indicators. These parameters can be monitored by dedicated sensors of which several examples will be discussed.

The in plane optical sensor (IPOS) platform, developed at the Holst Centre as a platform for many application areas (Figure 4). It can serve as a chemical sensor but can also be used for direct optical measurements of physiological parameters on the skin.

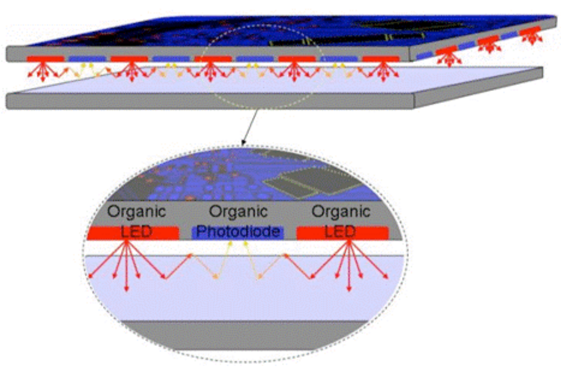


Figure 4 Principle of In plane Optical Sensor (IPOS)

An artist impression of such an application, a smart bandage, is given in Figure 5. The fact that printing can be used as a processing technology allows for the construction of arrays of optical elements, which may be advantageous for large wound areas or a combination of functions.

Figure 5 Artist impression of a smart bandage, integrating organic electronic devices with a wound dressing.

A foil based sensor capable of measuring the perfusion of the microvascular tissue in the wound area by means of photoplethysmography has already been shown [16] (PPG). Traditionally, a PPG is recorded with a pulse-oximeter giving additional information on the blood oxygen saturation (SpO_2). This is achieved by measuring the change in absorption due to the pulse (blood volume) at two wavelengths, typically in the red and near infrared from which the ratio between haemoglobin and oxyhaemoglobin is derived (Figure 6). For perfusion only one wavelength is sufficient, thereby simplifying design and manufacturing of the sensor device.

Apart from being a platform for use in many application areas, the IPOS may also be seen as a platform for the development and testing of manufacturing technologies. These include printing technologies for the active materials, barrier development for encapsulation of the devices, lamination and interconnection technologies for the final device assembly, printing of conductive structures, and lithography on foil.

For our sensor application we aim at an array of organic photo detectors (OPD) and a compatible array of organic light emitting diodes (OLED). The OPDs are based on a blend of poly(3-hexylthiophene) (P3HT, Merck Chemicals Ltd) and [6,6]-phenyl C61butyric acid methyl ester (PCBM, Solenne BV). This blend is used for photovoltaic research but has also been well studied for use in photodetectors [17, 18]. This blend has an optical band gap of 650 nm which is able to detect part of the red light.

The light emitting polymers (LEP) used for this study are a red emitting [19] and yellow emitting [20] material, obtained from Merck OLED Materials GmbH. The emission of the yellow material (λ_{max} 575 nm) has a strong emission shoulder in the red spectral region and has a strong overlap with the diode spectrum.

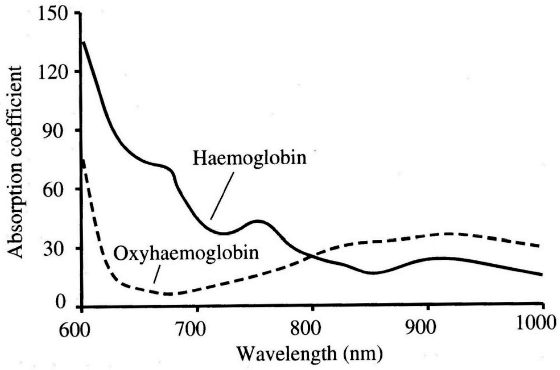


Figure 6 Absorption of haemoglobin and oxy-haemoglobin in the part of the spectrum that is useful for pulse-oximetry

In order to increase the flexibility with regard to the fabrication and design of these devices, we decided to fabricate the detectors and LEDs on separate foils. For example, this allows comparison of various printing and coating technologies and substrates, poly(ethylenenaphthalene) (PEN) and glass. A further implication of this approach is that one is essentially free to choose the order of functional foils in the final device.

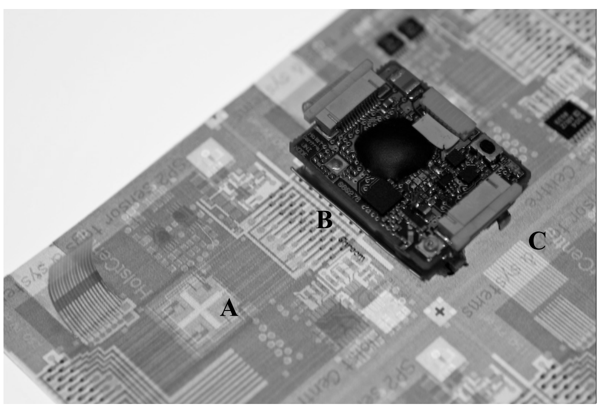


Figure 7 Mock up version of the sensor node showing the optical array (A), interconnection (B) and wireless node (C).

For the device discussed in this study we opted for a three foil assembly containing an OLED foil with an OPD foil laminated on it. The device is finished with a flexible circuit board containing noise filters (band pass 0.5 – 17 Hz), logarithmic amplifier and DC/DC converter for the power supply of the OLEDs. Data collection and OLED driving is controlled with a microprocessor embedded on a multifunctional wireless node [21]. The interconnection between the foils is achieved using a propriety lamination and interconnection technology. Figure 7 shows a mock up version used for testing the interconnection technology and of the attached node as well as a picture of the OLED and OPD foils.

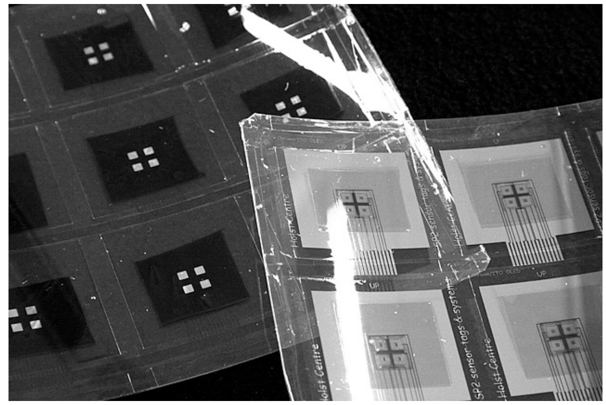


Figure 8 OLED (foreground) and OPD (background) foils.

The functional foils were processed in a batch wise manner. A typical work flow for both the OLED and OPD foils involves: lamination of a substrate to a carrier; deposition of the barrier; deposition and patterning of the anode and shunt lines; deposition of PEDOT:PSS and the active layer by means of spin coating or inkjet printing; evaporation of the cathode; and, finally thin film encapsulation [85]. Both foils can be made either bottom or top emissive (receptive) giving a large design freedom. Although processing on glass is a well know procedure, our devices were designed to be double sided. In such a device light leakage through the substrate can be avoided by using a top emissive OLED on one side of the substrate en a bottom receptive OPD on the other. This light will be emitted close to the skin and only reflected and scattered light will pass through a single substrate. Classical encapsulation with a metal lid is therefore not useful and a transparent thin film encapsulation was used. A further advantage of using a double sided device is that spin-coating can be used without the risk of contamination of the devices. A schematic overview of the architecture of both device designs is given in figure 4. The photodiodes with an area of 1 mm^2 are placed behind the LEDs (8 mm^2) and receive the reflected light via an opening in the middle.

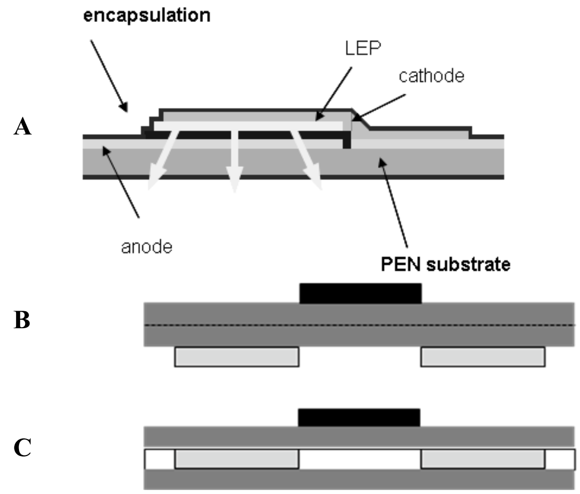


Figure 9 Schematic overview of the buildup (A) and cross section (B,C) of the devices. B shows the architecture in case a top emissive OLED (light gray) is used in combination with a bottom receptive OPD (black). The dotted line in the middle may be seen as the adhesive in case foil is the substrate. C depicts the combination of both bottom emitting and receptive devices. For clarity, the size of the actual devices are strongly exaggerated.

Top-emissive OLEDs

Bottom-emissive OLEDs are considered Lambertian emitters. As indicated before this might reduce the amount of light that is able to penetrate the skin because of "leakage" into the substrate. Clearly, the loss of light is strongly dependent on the thickness of the substrate. For the devices on glass and the first generation on foil (Figure 9, B) we therefore chose to use top emissive OLEDs. Here the light has to pass only through the thin film encapsulation, minimising this loss.

The red emitting material shows a peak emission at 670nm and has a large overlap with the absorption spectrum of the photodiode blend (Figure 10). Also the emission overlaps with the prime wavelength for the PPG measurement, 650 nm. This makes this material particularly useful for the saturation sensor (Figure 6).

The bottom-emissive devices had a maximum efficiency of 1.5 Cd/A. In top emissive devices this was reduced to a maximum efficiency of 0.45 cd/A at 8V and 320 cd/m^2 as measured with a luminance meter. The area of the LED was 8 mm^2, implying a current of 5.6 mA. These results show that the luminescence of top emissive devices is much lower than their bottom emissive counterpart. This is partly due to the reduced transparency of the cathode (60-70%) and partly due to the angle dependent emission by a cavity effect. Figure 10 shows a conoscopic measurement (Eldim, EZ-Contrast L160D) of a top-emission device driven at 9 V. Clearly the light is not emitted in a Lambertian mode but has a maximum between 40° and 50°. Control of this angle dependence, would give an interesting tool for light incoupling.

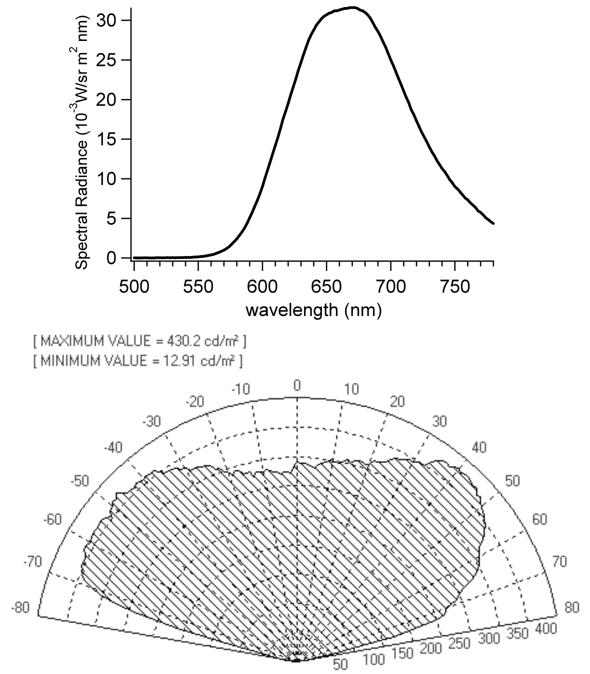


Figure 10 Emission spectrum of the red emitting OLED (top) and Cross section of a conoscopic image in Luminance (Cd/m^2) vs. the viewing angle (bottom).

Organic photodiodes

The photodiodes used in this study were bottom-receptive and had an area of 1 mm^2. The active layer was spin coated from a chlorine free solvent. No separate thermal annealing was required. A TEM picture (Figure 11) of an inkjet printed layer, using the same solvent, showed similar features as reported in literature [23]. This indicates that the morphology is very similar to that of films obtained by chlorinated solvents such as ortho dichloro benzene. This was corroborated with IV-measurements of both spin coated and inkjet printed devices using LiF/Al as cathode. Both showed high short circuit currents (Jsc) of 10 and 8.7 mA/cm^2, respectively (Figure 11) under approximately 100 mW/cm^2 white light illumination. The open circuit voltage (Voc) was 0.58 V with a fill factor (FF) of 0.56 for the spin coated device. The printed device showed a slightly lower Voc of 0.54 V but a dramatically lower FF. We attribute this to the inhomogeneity of the printed layer. Optimisation of the printing process including ink formulation is currently being carried out.

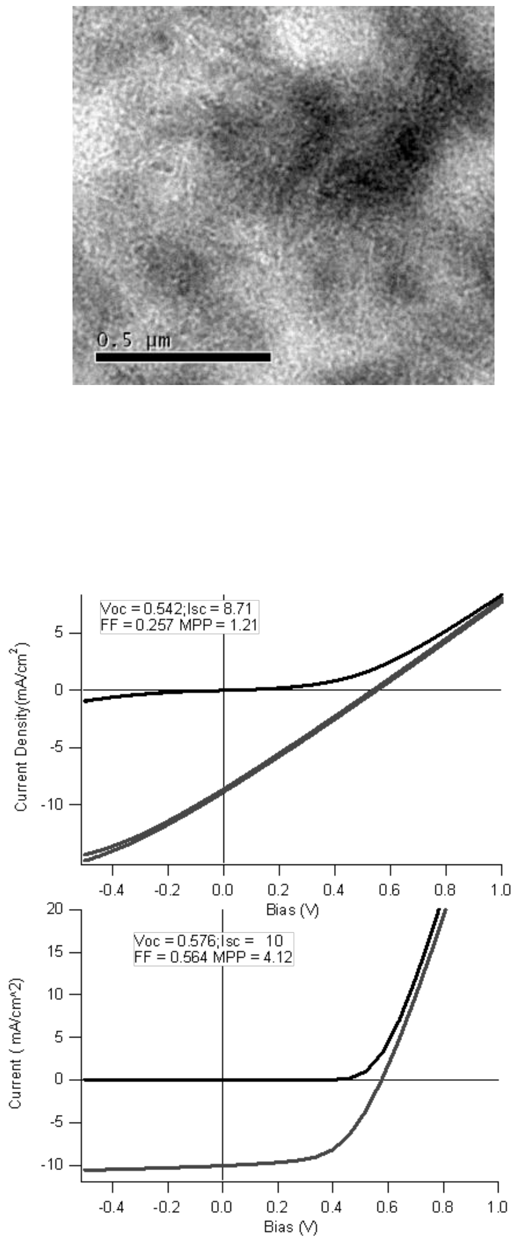


Figure 11 TEM image of an inkjet printed film of P3HT/PCBM blend (top) and IV curves of a spin coated device (bottom) and inkjet printed device (middle).

The noise levels of our devices can be relatively easily estimated from the shunt resistance (R_{sh}) measured in the dark. The measured thermal noise currents are in the order of 20 fA/Hz1/2. This is slightly higher than Si diodes. However, these levels are well below the measured currents, during operation (nA).

Photoplethysmography with OLEDs and OPDs

For the proof of principle we made use of a double side glass based device using two red emitting OLEDs and one photodiode. The OLEDs were driven at 9V. The measured photocurrents were filtered (band pass 0.5-17 Hz) and the AC component was amplified using a logarithmic amplifier (AD8304). The measurement was performed on the right index finger of a test person. A simultaneous measurement with a commercial pulse-oximeter (Nelcor N200) on the middle finger served as control. The resulting photoplethysmograms are shown in Figure 12.

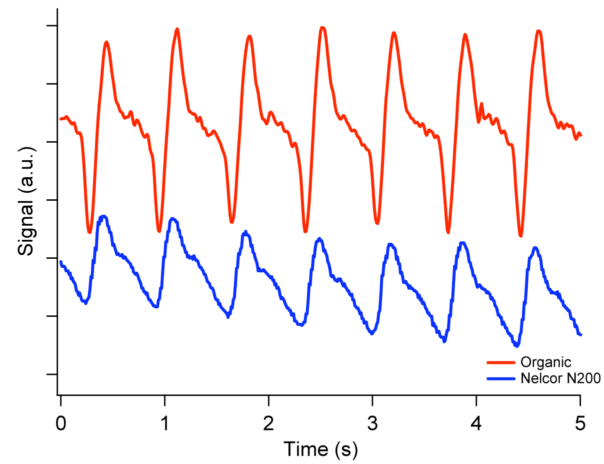


Figure 12 Photoplethysmogram showing the pulse of a test person measured with the organic device (top) and a commercial pulse-oximeter (bottom). The signals have been shifted on the y axis for clarity.

The signal of the organic device and the control match perfectly showing that our sensor can be used for measuring the pulse. The PPG is typical for a measurement on a finger [16]. The commercial device delivers a strongly smoothed signal, whereas the signal of the organic device was only slightly smoothed. The total signal measured by the photodiode was 10 µA, the relevant modulation (AC signal) was approximately 50 nA, estimated from the amplifier characteristics [24].

CONCLUSIONS

Homogeneous flexible light sources can be produced at large areas by OLEDs comprising thin film encapsulation and shunting lines for homogeneous current density. Recent advances at the Holst Centre have shown that areas of more than 100 cm^2 become feasible and compatible with Roll-to-Roll processing, thereby allowing low cost production methods. Holst Centre's strategy furthermore focuses on sensors incorporated in foils. Both functional components are ultimately suited to serve in devices for health and well-being in phototherapeutic applications like dermatological disorders and pain relief. Future developments are expected to increase conformability, area and efficacy of light sources.

Sensors based on organic electronic devices allow for such applications but can also be used for direct measurements of physiological processes on the body. We have shown a functional prototype comprising an in plane optical sensor node containing OLEDs and OPDs on foil that can be integrated with existing electronics using lamination technologies. More specifically the sensor was designed for photoplethysmography, measuring the perfusion in the skin. The sensor produced a signal from measurement on finger extremity which was comparable to a commercial pulse-oximeter, This shows that organic optoelectronic devices can well be used for the direct measurement of physiological parameters such as the perfusion in skin.

We foresee a bright future for light sources and associated sensor systems based on organic electronics in the application fields of health and well-being.

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