Chapter Eleven

The Potential of Polymer Photonics for Microflow Cytometry

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This chapter summarizes the current requirements of a microflow cytometer in terms of illumination source, optical detection and optical system. The state-of-the-art of the available organic photonic components is overviewed. The two parts are converged, and the real potential of organic photonics for microflow cytometry is investigated.

11.1 IMPORTANCE OF POLYMER PHOTONICS TO MICROFLOW CYTOMETRY

Flow cytometry has become a standard technique in cell biology and medicine. Commercially available flow cytometers have grown in complexity and performance, making use of multiple laser sources and an increasing number of detectors.[1]

A new disruptive technology, the bio-system-on-a-chip, holds the promise for new markets such as point-of-care and on-site analysis where portability and price are an issue. Microflow cytometers belong to the new optofluidics category that combines microfluidics and photonics. The main advantages of this approach are size reduction and the possibility of parallelization.

At its simplest, microfluidic flow cytometry chips consist of a microfluidic channel with a flowing liquid core. Detection is accomplished by focusing a laser into the channel and coupling out light (generally via microscope objective) to a photomultiplier tube (PMT), charge coupled device (CCD), or avalanche photo diode (APD). Fluidic control is accomplished via gravity fed systems, syringe pumps, or similar mechanisms.[2] A schematic example of such a system is found in Fig. 11.1. It shows the general architecture of the system including the four building blocks 'light source', 'optical system', 'detection system' and 'microfluidics',

The Microflow Cytometer by J S Kim & F S Ligler

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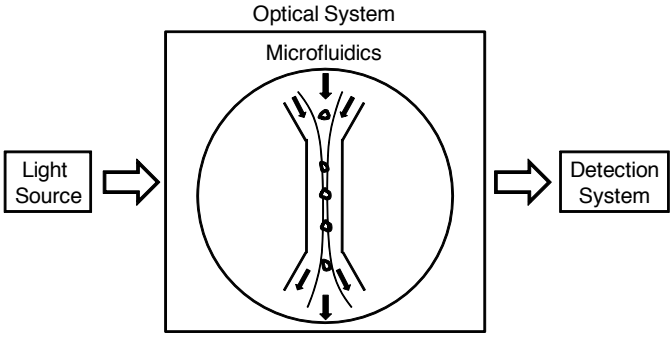

Figure 11.1. Microflow cytometers consist of the following basic building blocks: microfluidic system with a flow focusing unit, excitation source, optical detection system and optical system interfacing the different units. Color reference – pg. 344.

which also corresponds to the breakdown of this chapter. The requirements on the light source, the detector and the optical system will be detailed in the following sections.

Many microflow cytometers still rely on traditional bulky optics when it comes to focusing or extracting light from a small volume and make use of bulky external lasers and PMTs. In a further step, passive optical elements such as optical waveguides and microlenses for excitation and light extraction are added to the chip. As a result the optical alignment of light sources and detectors to the interrogation region in the flow channel is simplified. The issue related to the large footprint of these external sources and detectors still remains. In order to fulfill the promise of a low-cost lab-on-a-chip, new ways have to be found to integrate the light source and/or the detection with the microfluidics in a cost-effective manner.

Whether organic photonics in general, and polymer photonics in particular, will bring both light sources and detectors to where they belong, namely *on* the chip, is the question addressed in this chapter.

11.2 CURRENT STATE OF THE ART OF MICROFLOW CYTOMETRY

11.2.1 Requirements on the Light Source

Flow cytometry heavily relies on fluorescent probes — molecular tags that can be detected with appropriate excitation/emission conditions. Fluorescent probes can be used to detect receptors on cells, to determine the health and physiological state of a cell and even to measure gene expression in individual cells.[3] The main parameters are the excitation wavelength and the optical power of the source.

A huge range of probes suitable for biomedical analysis exists, and new ones are being developed constantly. While the absorption of fluorescent probes is rather broad band, on the order of 50 nm, most instruments rely exclusively on lasers as a source of excitation of these probes. Currently 488 nm is the standard

excitation wavelength in flow cytometry, low cost devices already use alternatives such as 532 and 635 nm laser diodes as well as violet laser diodes, which receive a great deal of attention as the possible next major biomedical laser source.[2] Even the most modern laboratory-scale, multi-laser flow cytometers typically provide no more than six discrete laser wavelengths, and most provide fewer. Coverage of the ultraviolet-to-infrared spectrum is therefore never complete, leaving large gaps in excitation capabilities. Several groups are therefore investigating the use of supercontinuum lasers in this field.[3]

In order to investigate one cell at a time and reduce the background noise, laser beams are tightly focused on the liquid core flow. A typical interrogation volume, *i.e.* the volume defined by the intersection of the core flow and the light cone, is roughly the size of a cell, i.e. in the order of a cube with 10 μm side length.[4] Standard microflow cytometry does not rely on pulsed laser sources in contrast to the related technique of 'scanning microflow cytology'. Concerning the optical power requirement, it is realistic to assume a HeNe laser with a power in the order of 10 mW.[5]

11.2.2 Requirements on the Detection System

Depending on the source and the intensity, a variety of detectors are commonly used in microflow cytometers: PMTs, APDs, CCD cameras, CMOS imaging arrays and PIN photodiodes.[1]

The detector should have external quantum efficiency close to 100% and a high — preferably single photon — sensitivity. The internal gain of a detector is the most important signal amplification step as it increases the signal with the smallest effect on noise.[1] In terms of typical signal power levels, one needs to distinguish between scattered light and fluorescence signal detection. Fluorescence signals can be smaller than 1 nW.[1] It is hard to define a precise figure of merit, but in order to compete with standard APDs, the minimum detectable power should be in the order of 5 pW.[5]

When it comes to speed, state-of-the-art flow cytometers, such as the BD FACSCanto II, can handle up to 10,000 events per second. Therefore the detector should be able to resolve signals of at least 100 kHz.

11.2.3 Requirements on the Optical System Integration

Initially, microfluidic chips were looked at as a mere replacement for the conventional flow cuvette of the cytometer, and all of the optical systems basically remained the same. However utilizing a bulk optical system significantly reduces the miniaturization benefits of the fluidics by requiring time-consuming optomechanical alignment steps, sometimes resulting in a "Chip-in-a-lab".

An intermediate solution is to integrate passive optical components to the microfluidic chip. This includes a large variety of optical fibers[6,7] and waveguides as well as on-chip lenses with the goal of transporting the photons from the light

source to the flow channel and from the flow channel to the detector. Ideally, these components are fabricated in the same process step as the microfluidic structures in order to provide self-aligned features suitable for mass-production.

Waveguides offer similar light confinement as optical fibers but in a more robustly integrated way, thereby avoiding the difficulties of alignment, epoxy-fixing or breakage. Other approaches, for instance from Bliss *et al.*[8], involve a combination of fibers and waveguides, interfacing the two by means of fiber-to-waveguide couplers.

The integration of the light collecting system directly on the chip is beneficial, since close-proximity detection can theoretically allow for lower loss and high-numerical aperture (NA) light collection due to the effectively 'immersed' optical system (no on-chip air gaps).[2] Although integration of optical waveguides on microfluidic chips provides several advantages in terms of alignment and total system size, it also comprises a few limitations. The integrated waveguides sources do not provide the very uniform, highly localized illumination compared to external interrogation sources collimated with high-NA microscope objectives. Similarly, light collection by fibers or waveguides does not provide the same localized NA light collection of traditional bench top flow cytometers.[2] In fact, most of the light collected by the fiber originates from locations other than that of the cell.[9]

Some of the drawbacks previously mentioned could be alleviated by adding on-chip lenses to the optical system, resulting in a more focused interrogation beam. This approach also allows an increased NA of the light collection from the cell, leading to higher signal/noise ratios and/or more sensitive detection. Wang *et al.*[10] demonstrated a microchip flow cytometer with integrated optical elements (waveguides, lens and fiber-to-waveguide couplers), all defined in a single layer of SU-8 polymer. In summary, to fully take advantage of the microfluidic approach, *the size of the optical system should scale down to the level of the fluidic system*. There are several technologies available to potentially enable such a reduction in size.

11.3 STATE-OF-THE-ART ORGANIC PHOTONICS

Organic optoelectronic molecules are a fascinating new class of materials. They combine the electrical properties of a semiconductor with the beneficial material properties of plastics.

Organic photonics devices are based on either small semiconducting molecules — mainly deposited by vacuum phase approach — or semiconducting polymers — deposited by liquid phase processes such as printing. Both approaches have specific advantages and disadvantages, which will be explored here. The main focus will be on printable polymer systems, which we believe promise large-volume, low-cost and customizable production.

In the small molecule approach, the basic building blocks (e.g. Alq3) have a low molecular weight and are essentially identical. In contrast, in the polymer approach, the building blocks are long, chain-like molecules with relatively high

molecular weights (a typical conjugated polymer has several hundred repeat units) and significant length and weight variations.

In the remainder of the text, the term organic light emitting diode (OLED) shall comprise both approaches. In order to highlight specific differences, the detailed terms small molecule organic light-emitting diode (SMOLED) and polymer light-emitting diode (PLED) will be employed.

11.3.1 State-of-the-art Organic Light Source

Organic electroluminescence is the electrically driven emission of light from non-crystalline *organic* materials, which was first observed and extensively studied in the 1960s.[11] In 1987, a team at Kodak introduced a double-layer OLED, which combined modern thin-film deposition techniques with suitable materials and structure to give moderately low bias voltages and attractive luminous efficiency.[12] Shortly afterwards, in 1990, a Cambridge research group lead by R. Friend announced a conducting PLED.[13] Since then, there has been increasing interest and research activity in this new field. Enormous progress has been made in color gamut, luminous efficiency and device reliability.

Since an electrical current is needed to stimulate the electroluminescent material, the semiconducting organic material in a light emitting diode has to be contacted with two electrodes. Furthermore, since the electrical conductivity of organic semiconductors is very low, one has to apply a large electric field across the semiconductor in order to pass the required amount of electrical current to stimulate light emission. A very thin ($< 0.2\ \mu$m) film of the organic semiconductor is thus normally sandwiched between two electrodes to form a light-emitting diode. Fig. 11.2 shows the simplified architecture of an OLED.

Glass or a plastic foil such as polyethylene terephthalate (PET) is used as substrate. A thin transparent anode, usually indium tin oxide (ITO), is deposited onto the substrate. The active organic semiconducting material is then applied, followed by the deposition of the cathode. Low work-function, hence reactive metals such as calcium and barium coated by a protection layer of aluminum, are commonly used as cathodes.

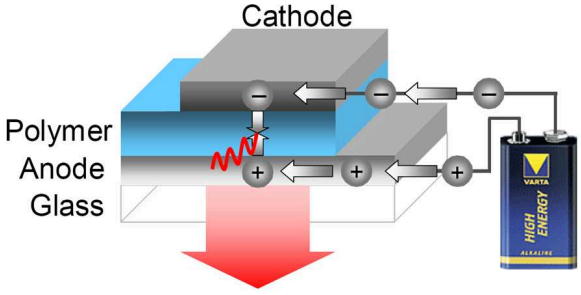

Figure 11.2. Schematic diagram of an organic light emitting diode (OLED). Color reference – pg. 344.

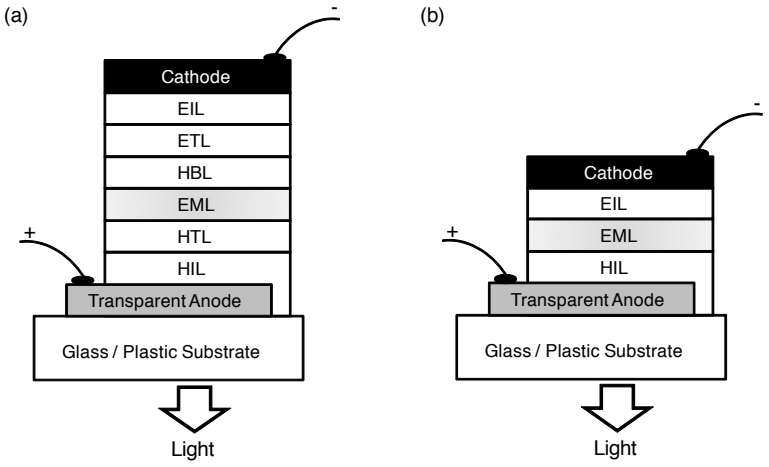

Figure 11.3. Detailed layer structure of (a) small molecule OLED (SMOLED) and (b) polymer light emitting diode (PLED). The stacks consist of a transparent anode (ITO) on a glass/plastic substrate, a hole injection layer (HIL), hole transport layer (HTL), emitting layer (EML), hole blocking layer (HBL), electron transport layer (ETL), electron injection layer (EIL) and a metal cathode.

When a positive voltage is applied to the anode, electrical charges are pushed through the organic thin film and current flows. Negative charge carriers (electrons) are injected into the device by the cathode and travel towards the anode, whereas positive charge carriers (holes) are injected by the anode and are driven towards the cathode. When positive and negative charge carriers meet in the organic thin film, they can "recombine" and emit light. This light escapes the diode usually through the transparent anode. In OLEDs the external quantum efficiency — ratio of the number of photons that escape the diode to the number of electrons that travel through the diode — is 1 to 10 %.

Layer structure, especially in the case of SMOLEDs can be considerably more sophisticated than the one depicted in Fig. 11.2. In order to assure a good charge injection, charge transport and radiative recombination in the emission layer (EML), additional layers such as hole injection layer (HIL), hole transport layer (HTL), hole blocking layer (HBL), electron transport layer (ETL) and electron injection layer (EIL) are added to the stack (11.3.2 Fig. 11.3).

As mentioned, one of the main differences between OLEDS based on small molecules versus polymers lies in processing. The most common way by which thin films of conjugated small molecules are applied is vacuum sublimation. The small-molecule material, which is normally a powder, is heated in vacuum to the point where molecules evaporate at a reasonable rate. A fraction of the molecules in the resulting vapor will fly in the direction of the sample and condense on its surface to form a thin film.

Vacuum deposited thin films of small molecules are normally polycrystalline and show a high degree of structural order. The sublimation process used to

deposit small molecules also ensures that the deposited films are very pure and of high quality. In contrast to small molecules, conjugated polymers are processed from solution. The most widely used deposition techniques are spin-coating, ink-jet printing and gravure printing. These methods produce rather poorly defined films with significant local material variations.

The polymer is first dissolved in a common organic solvent such as xylene. Then this solution is deposited onto the sample surface. As the solvent evaporates a thin amorphous polymer film remains. In the ink-jet method the organic material is dissolved in solvent and flies in a controlled manner out of the ink jet nozzle similar to inkjet printers used at home. The inkjet method applies the organic materials to the areas requiring pixels, precisely allowing for micro-level control of the formation of the film layers.

According to the review paper by Shinar *et al.*[14] the maximum reachable external quantum efficiencies for typical fluorescent SMOLED, fluorescent PLED and phosphorescent SMOLED are 6.5%, 10% and 26%, respectively. The highest efficiency OLEDs now exhibit electrical efficiencies above 20% and power efficiencies exceeding $50\,\mathrm{lmW^{-1}}$,[15] where 1 lm $\equiv$ 1.46 mW at 555 nm (the wavelength to which the human eye is most sensitive) at brightness L $\sim 150\,\mathrm{Cdm^{-2}}$, where 1 Cd $\equiv$ 1 lm $\mathrm{sr^{-1}}$.[14] For comparison, the value of 150 $\mathrm{Cdm^{-2}}$ corresponds to the standard brightness of an LCD computer monitor. Kido and coworkers have recently demonstrated $30 \times 30\,\mathrm{cm^2}$ white OLED (WOLED) panels with power efficiencies of 20 $\mathrm{lmW^{-1}}$ at L $> 1000\,\mathrm{Cdm^{-2}}$.[16] Schwarz *et al.*[17] have fabricated a white p-i-n OLED with luminous efficiency close to 16 $\mathrm{CdA^{-1}}$ for a brightness of 1000 $\mathrm{Cdm^{-2}}$.

For PLEDs the luminous efficiencies for red, green and blue emitters are 10, 15 and 9.9 $\mathrm{CdA^{-1}}$, respectively.[18]

In terms of stability of OLED devices, it is important to distinguish between the intrinsic lifetime of the electroluminescent polymer in inert atmosphere and the lifetime linked to degradation issues such as polymer and cathode oxidation—hydroxidation due to the oxygen and water in the atmosphere. The latter can be mitigated by proper encapsulation with glass/metal, UV-curable epoxies and the addition of getter materials, i.e. materials with the ability to bind oxygen and other gas traces, inside the sealed cavity.[19] Based on a recent review by Shinar *et al.*[14] red-to-green SMOLEDs and blue SMOLEDs feature continuous operating lifetimes >200,000 hours ($\sim$23 yr) and 100,000 hours ($\sim$11.5 yr), respectively, at 150 $\mathrm{Cdm^{-2}}$. Despite these impressive numbers, there are still serious lifetime issues to be solved, especially at high brightness ($> 1000\,\mathrm{Cdm^{-2}}$). For PLEDs the cited lifetimes are somewhat lower; however, they were taken at a higher initial brightness of 1000 $\mathrm{Cdm^{-2}}$. They are 89,000 hours ($\sim$10 yr), 79,000 hours ($\sim$9 yr) and 10,000 hours ($\sim$1 yr) for red, green and blue, respectively.[18] To our knowledge, the lifetimes of the PLED could be at least a factor 2 longer when operated only at 150 $\mathrm{Cdm^{-2}}$, depending on various factors such as duty cycles, peak luminance and pulsed driving.

One of the obstacles for the wide deployment of OLEDs in lab-on-a-chip applications that require narrow excitation lines, are the relatively broad emission spectra of OLEDs, with spectral widths on the order of 75 nm. A possible route to

shrink the linewidth is to fabricate the OLED on top of a distributed Bragg reflector (DBR)[20]. By modifying the thickness of the spacer layer between the DBR and the ITO, the emission wavelength can be tuned in the range of 495–625 nm.

A related topic is organic lasers.[21,22] They offer monochromatic light, are tunable over a wide spectral range and have low pumping thresholds. In many biosensing applications, small linewidth is desired in order to avoid spectral overlap between the excitation source and the marker emission and thus reach a higher signal-to-noise ratio. The quality of organic lasers has improved in such a way that they can be pumped by cost-effective inorganic violet diode lasers. Riedl *et al.*[22] have demonstrated an extremely compact organic thin-film DFB laser that is pumped with a 406 nm laser diode and tunable between 496 and 516 nm. Direct electrical excitation has long been considered the 'holy grail' for all types of semiconductor lasers, and organics are no exception.[23] Until electrically pumped organic lasers become reality, many challenges remain to be mastered.

An alternative to the just described thin-film organic laser might be the optofluidic distributed feedback (DFB) dye laser.[24] It consists of a microfluidic channel with an integrated DFB grating fabricated in polydimethylsiloxane (PDMS). A dye solution which acts as both the core of the optical waveguide and the gain medium can be introduced into the structure through the channel. These lasers are highly compact, widely tunable and robust. They can be pumped either by an external light source or on-chip laser diode.

11.3.2 State-of-the-art Organic Detection System

One of the most interesting structures for an organic photodiode (OPD) is the bulk heterojunction,[25,26] which results when electron donors and acceptors are dissolved in an appropriate solvent and deposited in solution. Bulk heterojunctions have proven to be a very successful concept to overcome the short exciton diffusion length of organic molecules.[27] However, they enhance the disorder and hamper the collection of photogenerated charge carriers. Therefore, creating bicontinuous and interpenetrating networks between the donor and acceptor phase is of key importance (Fig. 11.4). The use of high boiling point solvents and low spin speeds turned out to be a successful approach to increase the molecular order and thus the collection efficiency of charge carriers.[28]

State-of-the-art OPD-blends consist of poly(3-hexylthiophene):phenyl-C61-butyric acid methyl ester, P3HT:PCBM with weight ratio of 1:1. The layer thickness of the OPD is 240 nm and has been optimized for optimal external quantum efficiency, on/off ratio and lifetime.[29,28]

Although OPDs and organic photovoltaics (OPVs) are in many ways similar devices, they nevertheless fulfill a different set of requirements. For OPDs the spectral region of interest is often quite narrow, and the important parameters are the photocurrent and the dark current, with the dark current being only a fraction of the photocurrent. OPVs, on the other hand, must have high conversion efficiency over the entire solar spectrum, which reaches significantly into the near-IR.

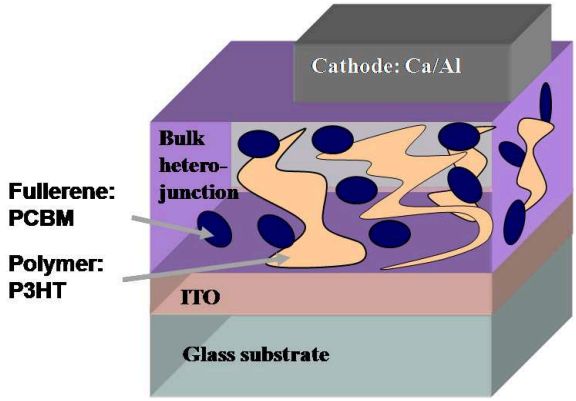

Figure 11.4. Schematic device architecture of a bulk heterojunction organic photodiode (OPD). Color reference – pg. 344.

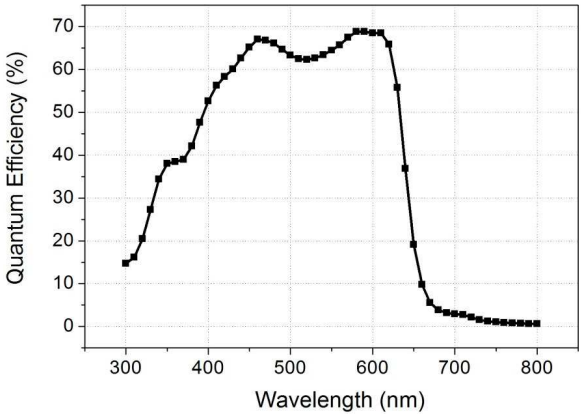

Figure 11.5. EQE at 0 V as a function of wavelength.

The external quantum efficiency (EQE) of such devices is above 60% for the range of 420-630 nm (Fig. 11.5).[29,30,31] Recent publications show a new class of low bandgap polymer with high sensitivity in the red / NIR.[32,33]

Typical frequency response — at 3dB attenuation — of bulk heterojunction OPDs biased at 0 V occurs at a frequency of several 100 kHz.[34,29] Although f_{3dB} increases with increasing reverse bias, the values for the P3HT:PCBM heterojunctions are more than three orders of magnitudes smaller than the ones of optimized small molecule photodiodes.[35] Further device optimization (reducing capacitance) and different driving schemes (e.g. under strong reverse bias) will lead to a somewhat improved response speed of P3HT:PCBM diodes, nevertheless these devices will be limited to frequencies of <1 MHz.

Punke *et al.*[34] have optimized their OPDs for speed by minimizing the pathlengths of the electrodes and by avoiding the parasitic capacitances formed by overlapping electrodes. They measured rise times as small as 1.6 ns and fall times

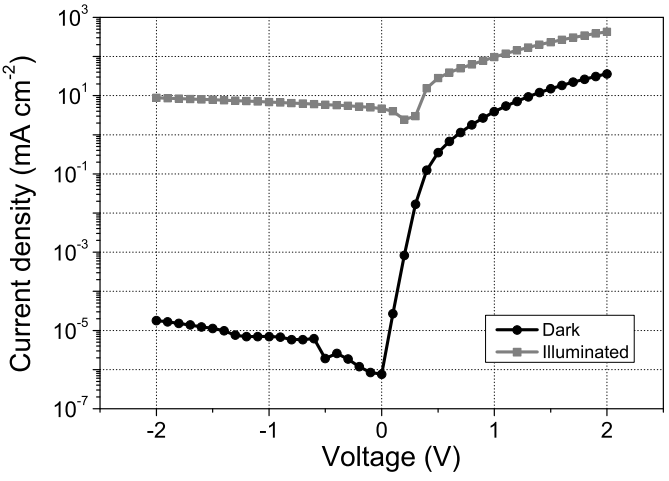

Figure 11.6. I(V) characteristics of a $2 \times 2\,\text{mm}^2$ PD measured in the dark and under ($40\,\text{mWcm}^{-2}$) illumination at 468 nm.

< 40 ns and derived a -3 dB cut-off frequency of $\sim$1 MHz at -5 V bias. State-of-the-art on/off current ratio of 10^6 at -1 V (Fig. 11.6) and dark current densities below 10 nAcm^{-2} at -1 V have been reported.[29]

Apart from spin-casting, ink-jet printed OPDs achieve very similar performance with EQE $> 50\%$ and on/off ratio $> 10^3$. From a processing and integration point of view, the inkjet technology is very advantageous since the polymers can be easily deposited on the chip with a positional accuracy of much less than 100 micrometers.Another figure of merit, especially when it comes to detecting small fluorescence signals, is detector noise. A low-noise figure is particularly important for the detection of weak side scattered light and fluorescence emission. It allows reducing detection levels and increasing dynamic range. For instance in OPDs fabricated in our lab, we obtained a specific detectivity of $7 \times 10^{12}\,\text{cm} \times \text{Hz}^{-1/2}\text{W}^{-1}$.[29]

Operating lifetime has also improved over the last few years.[36,29] As seen on Fig. 11.7, the photocurrent is stable for more than 1500 hours. After approximately 1800 hours the photocurrent starts to increase gradually. The on/off ratio decreases monotonically but is larger than 10^3 over the entire measuring period of 3000 hours (four months).

11.3.3 State-of-the-art Optical System Integration Using Organic Photonics

While there have been publications on organic photonics used in lab-on-a-chip applications,[2] to our knowledge there has not been a single peer-reviewed publication on a flow cytometer completely based on organic photonics. Knowing that the real potential of organics and flow cytometry lies in the miniaturization and integration, we would like to give the reader insight how far this field has advanced

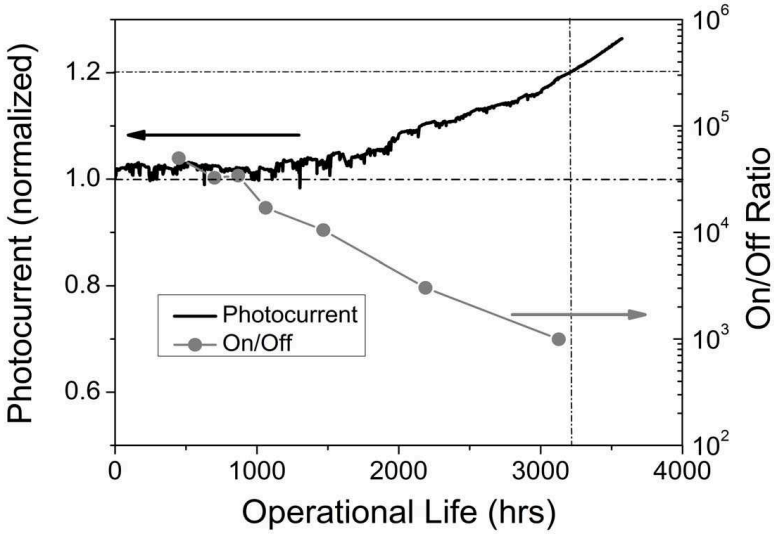

Figure 11.7. Stability of the PDs without PEDOT:PSS layer. The normalized photocurrent, measured at –1 V and wavelength illumination of 468 nm, is plotted as a function of operating time at room temperature in N_2 atmosphere.

in related application areas. Some of these areas might also be relevant for an integrated flow cytometer.

When it comes to integration of organic photonics on lab-on-a-chip two possible schemes are: (I) sandwich design and (II) waveguide design. The first scheme defines functional layers such as excitation, microfluidics and optical filtering layer and stacks them on top of each other. It has certain advantages, with respect to wafer-scale fabrication and parallelization of the detection. The second approach, which is the one our group pursues, builds everything around a waveguide structure. It enables better SNR and is considerably more flexible at the expense of lateral size, e.g. it is possible to increase the optical interaction length with the analyte and to integrate complex optical functions such as waveguides, dispersive elements and plasmonics.

In the following paragraphs different examples of both categories schemes are reviewed:

A simple yet powerful scheme of the stacked type for fluorescence-based assays has been proposed by Pais *et al.*[37,38] (Fig. 11.8). It is based on a high-sensitivity, cost-effective, cross-polarization scheme to filter out excitation light from a fluorescent dye emission spectrum. The cross-polarizers suppress the polarized excitation signal by 22 dB with respect to the randomly polarized fluorescent signal.

In order to detect oxygen and biological agents, a similar back detection geometry scheme was chosen by Shinar *et al.*[39,14] In this approach interdigitated OLEDs and PDs are placed on the same plane, separated from the sensing layer by an optical filter. The multi-analyte concentration is determined using a robust fluorescence decay technique with a pulsed OLED.

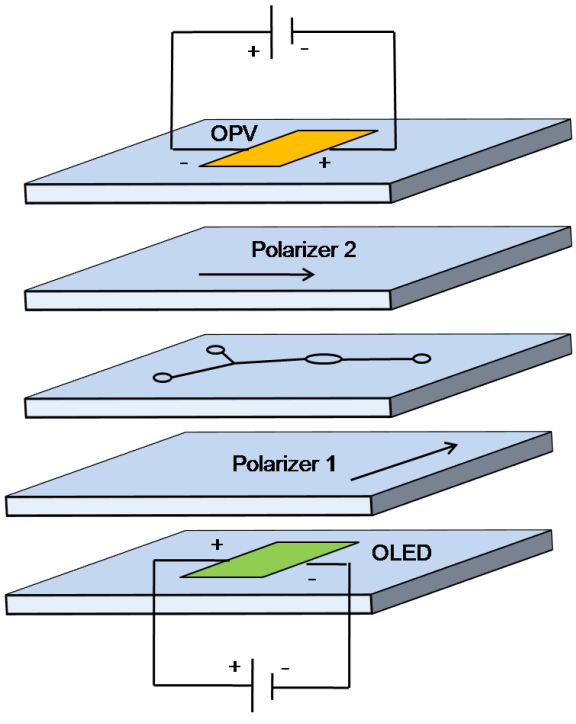

Figure 11.8. Schematics of integrated excitation/detection system. [Figure adapted from Banerjee *et al.*[38]] Color reference – pg. 345.

With a hybrid device using a silicon-based photodiode and an OLED, Shin *et al.*[40] have demonstrated an integrated fluorescence detector that achieved a limit of detection of 1 μM.

Hofman *et al.*[41] have used CuPc-C60-based thin film small molecule OPDs to successfully monitor chemiluminescence reactions. Their OPDs had an external quantum efficiency of ~30% in the 600–700 nm wavelength range and an active area of $2 \times 8\,\text{mm}^2$.

Another example uses an integrated PPV-based PLED excitation source for micro-scale fluorescence detection.[42] PLEDs also have been used as excitation source in micro-scale capillary electrophoresis.[43]

A company active in lab-on-a-chip is BioIdent.[44] They have developed a PhotonicLab^{TM} Platform consisting of the combination of printed opto-electronic components with microfluidic systems. The novel concept allows the integration of illumination and detection capabilities onto microfluidic-based devices using printing technologies (Fig. 11.9).

The ultrathin photodiodes with an overall thickness of only 300 to 500 nm show quantum efficiencies better than 0.5 and linear light-response over 6 orders of magnitude. The pixel size can range from 50 to over 1000 μm and inkjet fabrication allows tailoring the sensor layout to the needs of the specific application.[45] An

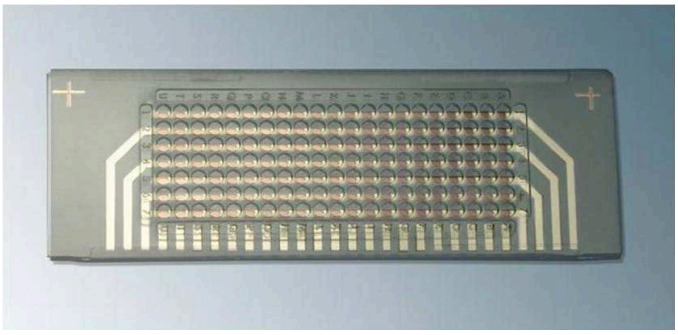

Figure 11.9. Disposable nanotiterplate with fully integrated optical readout system (© BIOIDENT Technologies, Inc.). Color reference – pg. 345.

equivalent OLED array can be generated using the same fabrication-procedure but with different organic materials. As a consequence, in principle any combination of light emitting and light detecting diodes can be printed on a variety of substrates for sample illumination and signal detection.[44]

A good showcase for the integration of polymer photonics in lab-on-a-chip applications is the European project SEMOFS (Surface Enhanced Micro Optical Fluidic System) which aims at the development of a fully integrated disposable biosensor. The detection is based on a surface plasmon (SP) scheme (Fig. 11.10): Light from an OLED is coupled into a single-mode waveguide and propagates towards the microfluidic flow cell containing the SP stack.

For a certain narrow wavelength range, the waveguide mode couples evanescently to a surface plasmon and is subsequently lost. This phase matching

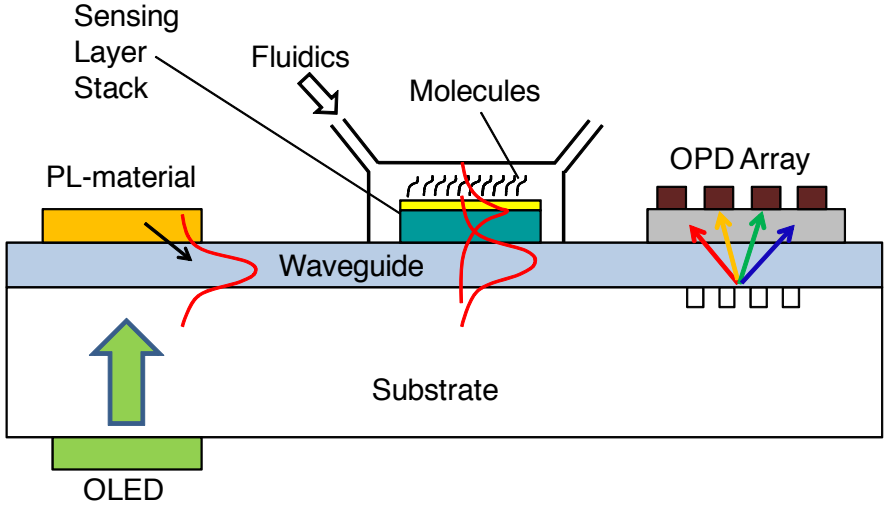

Figure 11.10. Schematics of microfluidics with integrated organic light source and organic mini-spectrometer. Color reference – pg. 346.

condition relates to a dip in the transmission spectrum which is detected by the organic mini-spectrometer. Since the dispersion of the SP mode is very sensitive to refractive index changes close to the metal surface, the adsorption of biomolecules on the functionalized surface of the plasmon stack translates into a shift of the phase matching condition and thus a shift in the transmission dip.

One of the key challenges for coupling light from an OLED into a waveguide is the fact that the waveguide has to be single-mode in order to have an unambiguous transmission dip. The OLED is a Lambertian emitter, i.e. it appears to have equal brightness independent of the observation angle. In contrast a single-mode waveguide allows propagation only from a small solid angle. The difficulty arises from the geometrically invariant radiance (power/area/steradian), *i.e.* independently of how tightly focused the photons are, the radiance basically remains the same. In order to get an intense focal spot, it is best to use a light source that has a high luminous intensity. Subsequently, two different schemes to couple the light from the OLED into a waveguide have been tested within this project. The first approach consists of depositing the OLED directly on top of the waveguide (Fig. 11.11(a)).

The approach is based on evanescent coupling of the waveguide mode and the dipoles inside the OLED. In this scheme our group has demonstrated coupling efficiencies as high as 3.2%.[46] There are two drawbacks in this geometry[47]: (a) The TM mode is suppressed due to the absorption of the metal cathode of the OLED itself, which is not acceptable for a surface plasmon resonance detection scheme; (b) the OLED contains optical loss layers such as the transparent anode (ITO based) and the metallic cathode. Thus, only light coupled into the waveguide within a

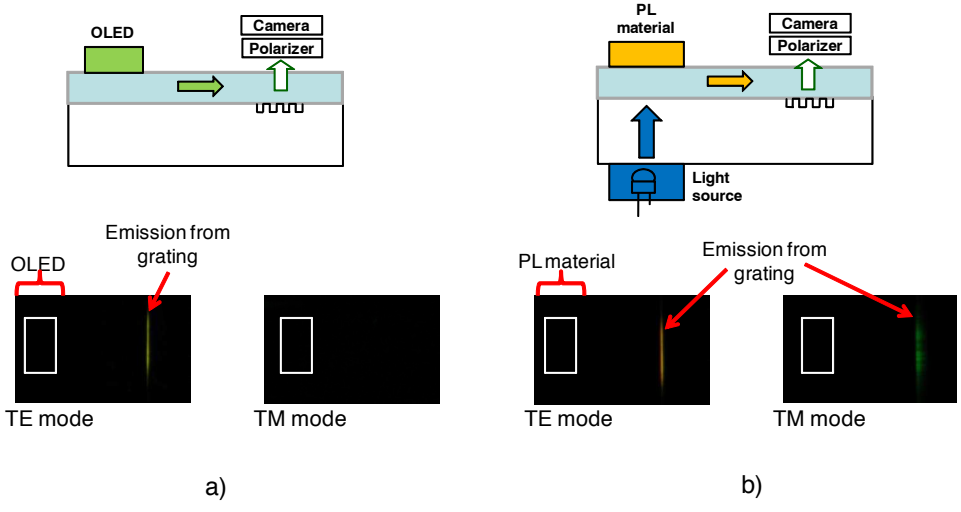

Figure 11.11. Two distinct schemes for coupling light from an OLED into single-mode waveguide. (a) Direct evanescent field coupling (TE mode only). (b) Indirect coupling by optically pumping a PL layer on top of waveguide. Color reference – pg. 346.

distance of 20 μm from the OLED edge propagates, the rest is absorbed by the electrodes.

In a second route, we use an indirect OLED-to-waveguide coupling in the way that the OLED emission was used to pump a PL-material layer located on top of the waveguide,[47,48] as depicted in Fig. 11.11b. In this configuration, loss layers (electrodes) of the OLED are kept at a safe distance from the waveguide. Thus, this architecture allows coupling of both TE and TM mode into the waveguide. Fig. 11.11b shows the extracted light at the out-coupling grating under TE or TM polarization. Another advantage of this second architecture is that the power coupled into the waveguide scales with the area of the illuminated PL-layer.A novelty in the device sketched in Fig. 11.10 is the organic mini-spectrometer that could be monolithically integrated on the biochip. It consists of a grating, etched in the glass substrate before depositing the waveguide, and a dense array of OPDs (Fig. 11.12).

The spectral resolution, $\Delta\lambda = \frac{g}{d}\Lambda$, is defined by the periodicity Λ, the length $g$of the grating, and as well by the distance $d$between the grating and the OPDs. Of course, the spectral resolution is directly related to the width of each individual OPD. Various masks were designed for spectral resolution ranging from 50 nm to 5 nm. Highest spectral resolution mask, as seen in Fig. 11.12(b), consists of 10 μm $\times$ 3.4 mm (width $\times$ length) OPD pixel with 5 μm spacing between two adjacent OPD pixels.

The guided light is diffracted by the grating and was detected by the organic spectrometer. In a first experiment, the PL-material has been pumped by a green OLED based on iridium complex emitter (Fig. 11.13).

The obtained spectral resolution was rather low ($\sim$10 nm) however proof of concept has been achieved. Replacing the OLED with an inorganic blue LED improved the spectral resolution of the organic spectrometer to below 10 nm FWHM (Fig. 11.13). Scaling up the PL-layer length from 2 mm to 6 mm will enhance the optical power in the waveguide by a factor of 5. Thus, a spectral resolution below 5 nm should be in principle achievable. Higher resolution organic spectrometers are currently under preparation.

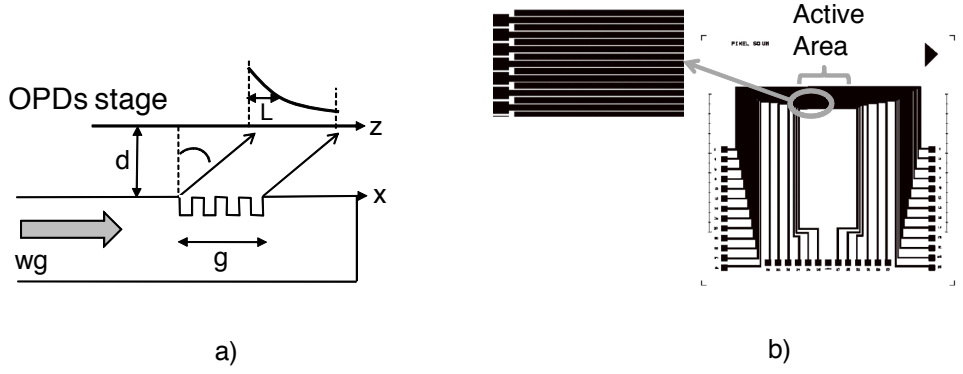

Figure 11.12. (a) Sketch of a simple model to calculate the spectral resolution. (b) ITO mask for organic spectrometer.

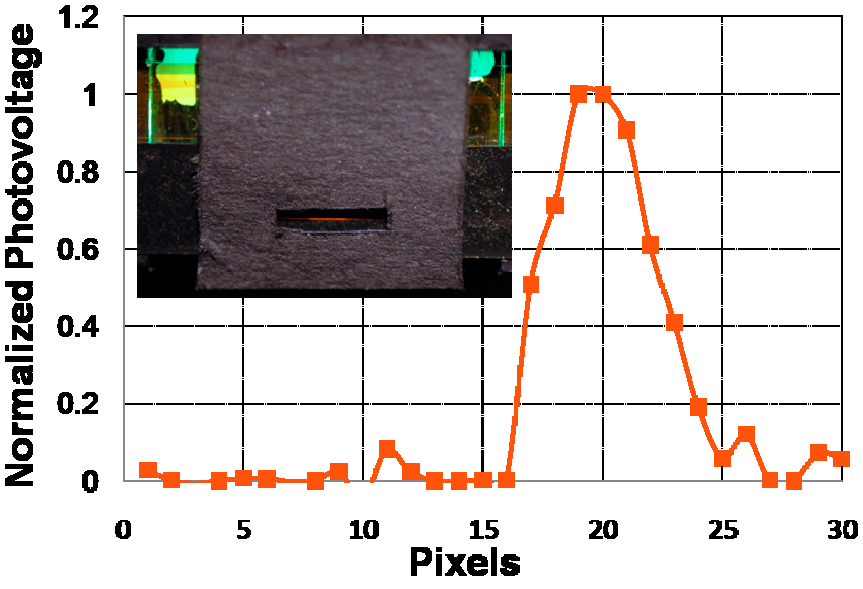

Figure 11.13. Organic spectrometer response when PL-material MEH-PPV pumped by green OLED. Inset: Photograph showing green OLED emission and red PL at output grating. Color reference – pg. 347.

The biochip developed within the SEMOFS project highlights the relatively simple deposition processes of organic material for integrating light source and light detection in a lab-on-a-chip device. However, it will be a complex task to converge all the different building blocks including the microfluidics and the surface chemistry to a fully integrated device. Figure 11.4 illustrates the end goal of this activity: an integrated organic opto-fluidic device.

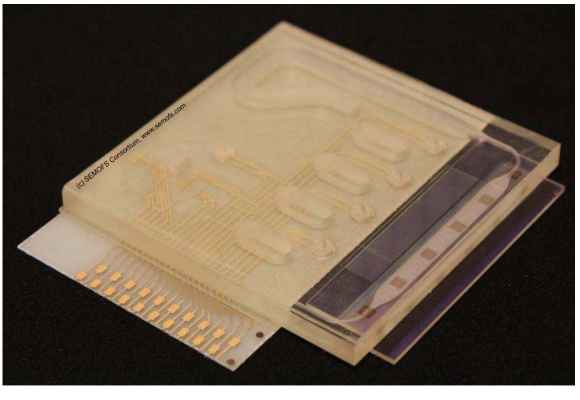

Figure 11.14. Top view of the SEMOFS chip with microfluidic components. Color reference – pg. 347.

11.4 OPPORTUNITIES AND CHALLENGES FOR THE APPLICATION OF ORGANIC PHOTONICS IN MICROFLOW CYTOMETRY

As mentioned earlier there are actually no publications that propose a flow cytometer entirely based on organic photonic components. A valid question is whether time is not ready yet or if there is some fundamental issue. A rough calculation shows that the OLED illumination is going to be the most challenged part. An OLED is by nature a large area emitter which emits in a half sphere. As such it is extremely difficult to focus it tightly without using very bulky optics. Assuming a conventional flow cytometer with a 10 mW laser focused on a flow channel on an area of $10 \times 10\ \mu m^2$ yields an intensity of $10^7\ mWcm^{-2}$. An OLED with a brightness of $1000\ cdm^{-2}$ on the other hand emits roughly $1\ mWcm^{-2}$ into a half space ($1\ lm = 1.46mW$ at 555 nm). There is a mismatch of 7 orders of magnitude in intensity, which is extremely challenging to compensate with external optics. A possible workaround is to make use of near-field light concentration methods such as high-quality factor optical resonators[49] or light-harvesting plasmonic structures.[50]

The detection side looks promising. As shown in the section on state-of-the-art of organic detection system solution bulk heterojunction OPDs provide external quantum efficiencies between 60% and 70% over the whole visible range. In terms of detector speed, state-of-the-art OPDs should be able to detect signals in the 100 kHz allowing for high-speed screening.[29] Speed-optimized OPDs have a frequency response up to 1 MHz.[34] In terms of noise figure, they reach a specific detectivity of $7 \times 10^{12}\ cm \times Hz^{-1/2}W^{-1}$, which is comparable to common inorganic silicon photodiodes like for example a Hamamatsu S2551 with a specific detectivity of $1.5 \times 10^{13}\ cm \times Hz^{-1/2}W^{-1}$. A classical method to externally amplify the detector without amplifying the inherent noise is the lock-in technique. Tung *et al.*[9] have applied it to standard PIN photodiodes but it should, in principle, also work for OPDs.

While attractive, the approach of solely replacing the external APD, PIN diode by an OPD would also mean not to realize the full potential of organic photonics. A big potential lies in the integration possibilities, i.e. the possibility to place a single OPD or an OPD array exactly where it is needed. This is especially true for liquid processed polymer organic photonics. The tool of choice in that context is inkjet printing which allows precise layer deposition with a spatial resolution better than 100 μm. We think it will prove very fruitful to investigate additional functionalities.

The ability to incorporate OPD in close proximity to the microfluidic channels would eliminate the need for complicated collection optics including bulk lenses, microscope objectives and fiber optics. The ability to easily fabricate OPD *arrays* opens up unprecedented possibilities in increasing the signal quality by performing multiplexed data collection and employing time-delay cross-correlation for signal enhancement.[51] Reducing the size of the optical system will also reduce the optical path length and therefore absorption losses, an important consideration in

fluorescence measurements.[2] OPD arrays can also perform imaging functions and thus replace bulky CMOS and CCD chips.

In the section on state-of-the-art optical system integration using organic photonics, we have seen a few very promising examples of system integration of OLEDs and OPDs for sensing applications. A highlight and potentially useful are polymer OPD arrays in conjunction with microptical elements such as diffraction gratings as in the case of the first organic mini-spectrometer.[47] Micro-optical elements such as gratings and microlens arrays may be fabricated by high-throughput low cost technologies, preferably in the same step as the patterning of the microfluidics channels.

In summary, we have seen that organic photonics are low-cost, environmentally friendly and thus potentially disposable. Because of the flexibility in processing and the ease of integration with various types of substrates including plastics, it could potentially allow the fabrication of very compact, lightweight and therefore portable devices. These features make organic photonics potentially attractive for application areas such as environmental sensors, personalized diagnostic tools deployed on large scales and screening tools for professional biologists. For certain applications, disposability might actually not just be a cost argument, but rather a requirement, e.g. in biological testing when working with pathogens.

Classical flow cytometry is a very mature field and hence the demands on the light source, the detector and the optical system are very high. Therefore it is reasonable to assume that microflow cytometry might not be the entry point of the emerging organic photonics technology. However we believe in the potential of organic photonics in related fields of diagnostics and screening, and that with increasing maturity, the technology will also penetrate microflow cytometry.

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