PROJECT FINAL REPORT

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1　Final publishable summary report

1.1　Executive summary

The bioEGOFET (Electrolyte-Gated Organic Field-Effect Biosensors) project aims at an electrical transduction of a bio-recognition event by combining the specificity of a defined bio-recognition element with the high sensitivity of the field-effect transduction principle. The recognition of the analyte is achieved through an antibody attached on top of a polymeric organic semiconductor, which constitutes the core of the device. The polymeric film is equipped with two conductive electrodes placed at a short distance from each other, which act as the source and drain electrodes of the transistor. Various strategies have been considered to immobilise the bio-recognition element on the semiconductor layer; in all cases, the required outcome was to maximise the recognition capability and minimise any non-specific binding. A unique feature of the device is to take advantage of the high inherent capacitance of the double-layer that forms at the interface between the semiconductor surface and the aqueous liquid used to carry the analyte in proximity to the sensitised semiconductor surface, thus achieving low-voltage operation. Furthermore, the electrolyte-gated architecture avoids the need for a dielectric layer between the semiconductor and the gate electrode, thus skipping one fabrication step. In this configuration, the gate electrode simply consists of a piece of metal immersed in the electrolytic solution. Implementation of the device on a flexible substrate was realized through low-cost printing-compatible techniques.

To reach its objectives, the project was divided into the following five work packages.

WP1 (electrolyte-gated OFET) aimed at setting a protocol for the fabrication of the electrolyte-gated organic transistor (EGOFET) that constitutes the transducer and amplifier of the electrical signal delivered by the sensor. The steps towards this goal were to select appropriate polymeric semiconductors, fully characterise the electrical properties of the transistors and test the device in terms of stability and response speed.

The goals of WP2 (bio-recognition elements) were first to select antigen-antibody systems for the specific bioEGOFET final application. Next, three different immobilisation strategies were developed for the anchoring of the bio-recognition molecules on the surface of the polymeric semiconductor. The highly specific biotin-avidin reaction was chosen to demonstrate the functionality of the modified surfaces. Finally, the bio-functionalised surfaces were characterised and the target molecule quantified for both the avidin-biotin system and the defined final application.

WP3 (biosensor implementation) aimed to produce the biosensors by combining the protocols for transistor fabrication and immobilisation of the bio-recognition element defined in WPs 1 and 2. The biosensor figures of merit (sensitivity, selectivity and detection limit) measured using the biotin-avidin test-bed were used to select the best performing architecture that was eventually implemented with the printing techniques.

The main objective of WP4 (modelling and testing) was to develop models for the electrolyte-gated sensors. Of particular importance was to identify the mechanism that allows the transduction of the bio-recognition event into an electrical signal.

WP5 aimed at proving that reliable sensors can be produced with good reproducibility at low-cost with printing techniques. This included the identification of the materials, the evaluation of their printability, and the optimisation of the structure of the device. The overall goal of this WP was to evaluate the feasibility of a printed sensor with the electrolyte-gated architecture.

1.2 Description of project context and objectives

Cost reduction has become a primary goal of healthcare politics at large. To survive and succeed in these challenging conditions, it is imperative for laboratories to find ways to adapt, and implement new strategies to help them save on costs. In today's competitive environment, revenues per diagnostic test are continually falling, requiring industries to increase their productivity. The high demand towards improved quality and error free services and the need to ensure patient satisfaction to retain their fidelity have prompted laboratories to adopt novel technologies such as automation and point-of-care in order to reduce their expenses. The acceptance of new technologies is expected to drastically improve among those who are currently reluctant to invest to achieve long term cost reduction goals.

The leading idea of the FP7-funded project BioEGOFET (Electrolyte-Gated Organic Field-Effect Biosensors) was to bring together organic electronics and biomedical testing to create an innovative single-use biosensor. The primary goal of the project was to bring a *proof of concept* for a sensor that can selectively detect an antigen present at a concentration as low as a fraction of a micromole per litre in a water solution. In addition, elements for the fabrication of the sensor by printing techniques on a flexible substrate have all been demonstrated, although a reliable, fully working printed device is still lacking.

In practice, the central element of the sensor is an organic semiconductor constituted by a commercially available conjugated polymer. The semiconductor layer is deposited on a substrate previously equipped with two conducting electrodes separated by a short distance. This configuration is liable to form a transistor structure that can convert the bio-recognition event into an electrical signal delivered by the sensor. Bio-recognition is achieved by attaching an antibody on top of the organic semiconductor.

Selecting the best technique to immobilize the bio-probe to maximize recognition capabilities and minimize non-specific binding was a key issue of the project. The objectives of the project listed three strategies do be taken in charge by three members of the consortium. In each of the strategy, the bio-recognition element is coupled:

1. To biotin-maleimide groups obtained by functionalization of the semiconductor surface,
2. Directly to the semiconductor polymer or to biotin groups in a co-polymer thereof,
3. To biotinylated phospholipids embedded in phospholipid bilayers formed by fusion of vesicles.

All of the three strategies led to sensors that reached the targeted detection limit. However, the project work plan anticipated to choose of one strategy for the realization of a final demonstrator fabricated by printing techniques. The choice was based on the compatibility of the structure with the fabrication technique. Another important criterion was the absence of non-specific bindings, which would be responsible for reducing the selectivity of the sensor.

To attain low-operating voltage and low-power consumption, the bioEGOFET sensor takes advantage of the high dielectric capacitance offered by the liquid medium used to bring the bio-marker to be detected up to the semiconductor surface. The validity of this concept has been confirmed by the work done during the project.

The project also demonstrated the implementation of the various parts of the device by printing techniques on flexible plastic substrates, thus providing elements for a future development of low-cost fabrication.

One of the main advantages of an electrical signal resides in the fact that multiple markers could be combined in a single device to detect several biomarkers in a single test.

Future applications of the device can be envisioned as low-cost, simple to use, disposable biosensors that could be used at the doctor's office or even at home. This would favour utilizations in places remote from complex bio-analysis facilities, for instance in developing countries. Low-cost fabrication by printing techniques on plastic of paper substrates will facilitate these applications.

1.3   Main S&T results / foregrounds

WP1. Electrolyte-gated OFET

The transducer principle includes replacing the insulator of the organic transistor by an aqueous electrolyte. In this configuration, the gate consists of a piece of metal immersed in the electrolyte that remains distant form both the organic semiconductor and the source and drain electrodes. The objectives of WP1 were to set up protocol for the fabrication of the electrolyte-gated organic field-effect transistor (EGOFET) that would afterwards constitute the transducer and amplifier of the sensor.

Task 1.1. Organic semiconductor selection

Poly(3-hexylthiophene) (P3HT) was used as our prime organic semiconductor. Layers are prepared from a solution, either by spin-coating or ink-jet printing. As P3HT revealed to be too hydrophobic, it has to be modified by attaching hydrophilic groups. The group chosen was carboxylic acid (COOH) that was also used to attach the appropriate functional group for bio-detection. Several P3HT-COOH copolymers were synthetized by UPD. During the course of the project, it appeared that the functional groups could strongly physisorb on pure P3HT. Accordingly, most of the sensors realized during the project used pure P3HT as the organic semiconductor. Occasionally, the P3HT surface was modified by a plasma treatment (see WP3).

Tasks 1.2 (realizing EGOFETs including different semiconductors) and 1.5 (device characterization of the EGOFETs)

Three different semiconductors were chosen for initial studies, primarily focusing at finding out which semiconductor material can be used as the transducer in the field-effect operation mode. This mode of operation, in which charge transport occurs along the outermost semiconductor surface, is desired since the targeted receptors in the biosensor will be located at the electrolyte-semiconductor interface.

To anchor and immobilise receptors on sensor electrodes and devices, COOH-groups are commonly added along the surface. The EGOFETs must operate in the pure field-effect mode, i.e., bulk doping of the semiconductor should be prevented. However, COOH-groups are known to make the interface between the electrolyte and semiconductor more prone to ion permeation. Further, COOH-groups inside the bulk of the semiconductor may enhance ion migration and mobility. Three semiconductors were chosen (P3HT, P3HT-COOH15 and P3CPT) including various COOH side-group concentrations, 0, 15% and 100% respectively. From impedance spectroscopy and transistor current-voltage characterisations, we found that only two candidates can be considered for true field-effect operation in an EGOFET sensor. The P3HT and the P3HT-COOH15 showed excellent field-effect transistor modulation, see Figure 1, 2 and 3 and Table 1, while the P3PCT material showed clear electrochemical current modulation.
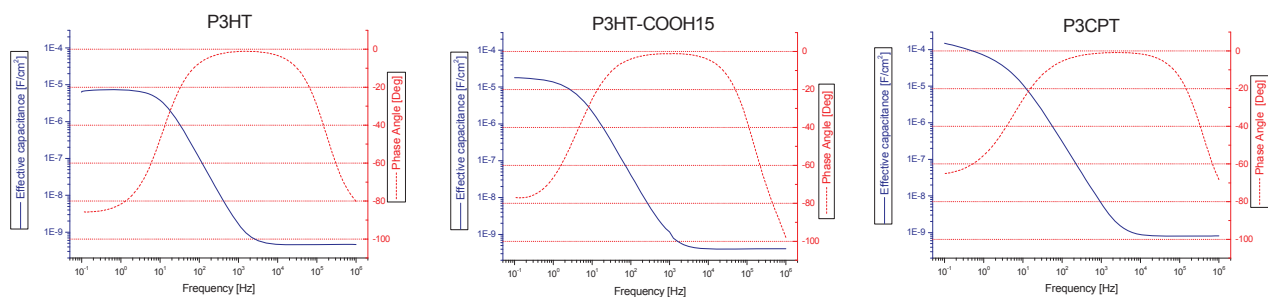
Figure 1.1. Impedance data for the three chosen semiconductors (P3HT, P3HT-COOH15 and P3CPT) in the structure source electrode-semiconductor-electrolyte-gate electrode. For P3HT and P3HT-COOH15, stable double layers are established at low frequencies, below 10 Hz. This is further supported by phase angle values close to 90°; the capacitance saturates at *ca*. 10 µF/cm^2. This does not apply to P3CPT, i.e., the material is expected to undergo bulk doping.
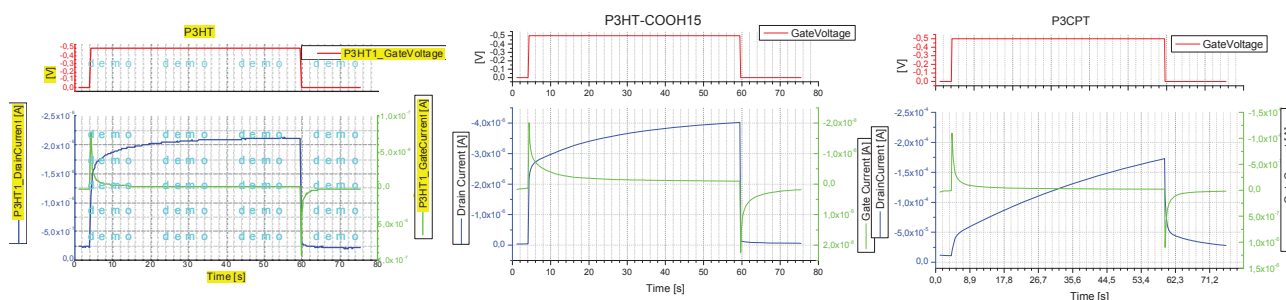

Figure 1.2. Response time of transistors with the three semiconductors. For P3HT and P3HT-COOH15 the response occurs at times lower than 1 s. For P3CPT the corresponding response time would be larger than 1 min.
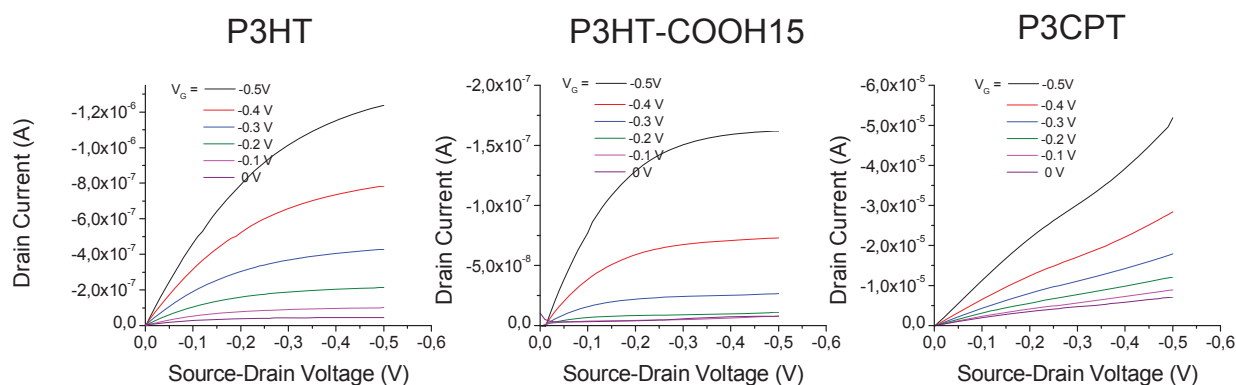

Figure 1.3. Typical output curves for the three polymers P3HT, P3HT-COOH15 and P3CPT. P3HT and P3HT-COO15 exhibit clear current saturation and typical drain current levels of field-effect operation. There is no clear saturation for P3CP and the drain current level is typical of bulk charge transport occurring in a doped organic semiconductor of a thickness of around 100 nm.

Table 1.1. Evaluation of the operating mode of EGOFETs made with the three polymers.

	P3HT	P3HT-COOH15	P3CPT
Phase Angle at low frequency (0.1 Hz)	-90°	-80°	-70°
I_D (VG = -0.5 V, sweep rate = 80 mV/s)	1.6 µA	0.11 µA	58 µA
I_D (VG = -0.5 V, sweep rate = 3 mV/s)	1.8 µA	0.11 µA	210 µA
Switch-on time	≤ 1 s	≤ 1 s	> 60 s
Field-effect operation in DI water	Yes	Yes	No

Milestone MS1. Selection of the EGOFET structure
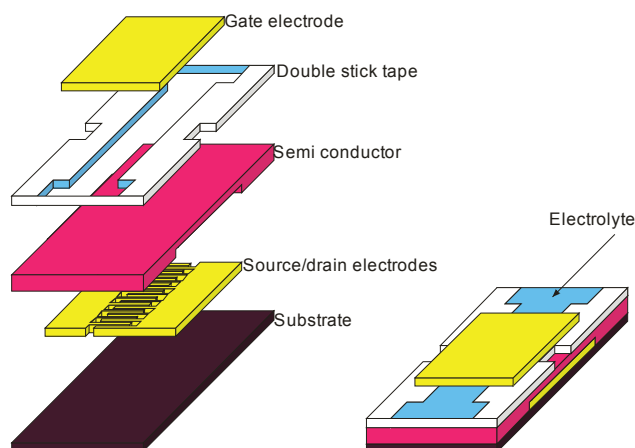
1. Rigid substrate (SiO_2 on Si-wafer).
2. Drain and source electrodes consisted of vacuum-deposited Ti (or Cr) (adhesion promoter) and then Au, patterned through standard photolithography. The channel dimensions decided on and acquired in this manner were L = 10 μm and W = 10 mm.
3. The semiconductor is spin coated onto the patterned substrates from solution.
4. A double stick tape, Mactac, received from Printed Electronics Arena (PEA, ACREO – contact person: Petronella Norberg) was evaluated as a simple space for the aqueous electrolyte between the gate electrode and transistor channel. The tape was applied manually on top of the chips with parts of the tape cut out to enable direct contact between the electrolyte and the transistor channel. Cutting was performed using a plotter with a plotter knife.
5. Flat gate electrodes were applied on top of the gasket tape structure to define the channel for aqueous sample and to define the gate-sample-channel configuration. The gate was applied in such a manner that the whole area above the transistor channel is covered, but leaving openings at both sides of the transistor to be used as wells where the aqueous solution can be applied or extracted. The space created between the transistor channel, flat gate and gasket tape has the approximate dimensions 0.5 × 2 × 0.1 mm^3.
6. This device configuration provides an easy route to add small amounts of electrolyte, which is guided into the gasket tape compartment entirely by capillary forces. Moreover, other electrolyte solutions can be added to the original electrolyte while probing the transistor.

Actually, the transistor fabrication process has been rationalized to exclude step 4 during normal testing of devices, which in most cases does not change anything in device performance. This also meant that the aqueous electrolyte was applied as a droplet directly onto the channel after which the gate electrode, held in place by a probe holder, was dipped or placed on top of the droplet.

Task 1.3. Optimising the semiconductor surface

The inclusion of a phospholipid layer (PL) on the organic semiconductor was one approach to immobilize the bio-recognition element, and also to improve the surface tension. EGOFETs including a zwitterionic PL film at the interface between the organic semiconductor and the ionic-gating solution were investigated. The PL layer was shown to efficiently act as a barrier against ion diffusion into the semiconductor bulk, thus allowing the PL–EGOFET to operate mostly in a capacitive mode when gated by a (phyiological) salt solution. The EGOFETs showed good electrical performance with a field-effect mobility of 6.4 × 10^{-3} $cm^2V^{-1}s^{-1}$ and a drain current on/off ratio of 10^3. This result constitutes a possible basis for the further developments involving the integration of receptors and proteins in general.

Task 1.4. EGOFET devices also comprising simple fluidic/reservoir systems

A fluidic reservoir structures was developed during the first year, utilizing a gasket tape structure to attain a simple fluidic channel. No change in device performance was noticed, which led to the decision of leaving further development to WP5, now instead using an inkjet manufacturing technique.

Task 1.6. Current-voltage characteristics and performance optimization

This task has primarily focused on modelling the EGOFET device. We have used the following model for the gate-insulator-channel configuration:

1. The first regime is when the gate voltage V_{GS} is lower than the flat-band voltage V_{FB}. Here, the semiconductor layer behaves as a dielectric since electrons cannot be injected from gold source electrode. Accordingly, we have two capacitances in series, the double layer capacitor and the semiconductor capacitor, and we see the latter, which is the smallest one.
2. Below V_{FB}, the regime of accumulation of holes takes place, so that the P3HT film becomes conductive and in this case the double layer that is formed at the electrolyte-semiconductor interface represents the entire capacitor.
3. Based on this simple model, the flat-band voltage (third regime) can be estimated as the onset of the capacitance increase (i.e. the transition between the first and second regime), which would correspond to the onset of hole accumulation.

We have explored metal-electrolyte-semiconductor stacks in which the top electrode material was changed (six different metals were tested) to mimic a sensor reaction. The bottom (drain/source) electrode is always made of gold. We observe a linear correlation between the flat band voltage and the shift of the threshold voltage. The threshold voltage can vary over almost 1 V as the metal electrode material is changed from a low work function to a high work function material. The corresponding change in drain current is around 500, i.e. a sensing gain ($\Delta I_D/\Delta V$) of almost 10^3 can be achieved.

Task 1.7. EGOFET stability and speed using automated measuring setup

The speed limitation of the EGOFET device was assessed in Tasks 1.1 and 1.2. We reached the conclusion that it can take up to 1 s to establish a stable double layer, and thus a corresponding stable sensor measurement.

The stability of the transistors was shown to depend on the exposure to both ambient air and light. Under bad conditions, extensive exposure to air and light, the device performance rapidly deteriorates. Storing devices for several days without degradation of performance was also proved possible as long as samples were stored in the dark.
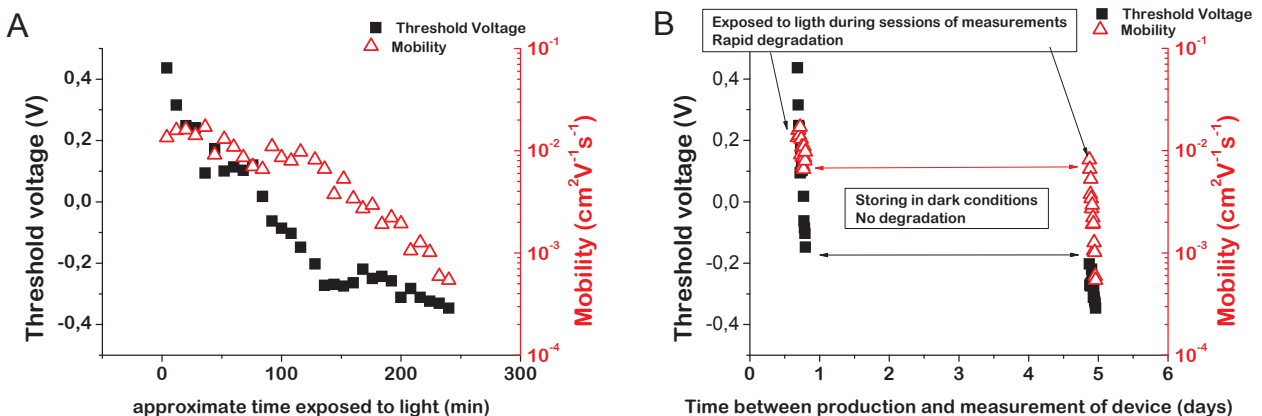

Figure 1.4. Degradation of EGOFET characteristics over time plotted as function of (A) time exposed to ambient light and (B) time from production to characterization.

The degradation process of the components was investigated by extracting characteristic transistor parameters from devices exposed to relevant environments for different time periods. The environment was normal ambient conditions (in ambient air and indoor lighting) and unsealed storage (in ambient air but in darkness).

1. In a first set of measurements, the effect of degradation was explored on many transistors on a single chip.
2. The second set aimed at estimating the possible shelf life time for just one single device.

The threshold voltage and mobility values are given as a function of time when the transistor is exposed to ambient light and air prior to the measurement (Figure 1.4A) and also as function of time from the finalized production and the actual measurement of device (Figure 1.4B). In Figure 1.3B, the time during which the devices were stored under dark conditions is given (during a period of four days). It becomes obvious that very little degradation occurs during this period as compared to the two periods when the devices are exposed to light (measured one and five days after the actual production).

WP2. Bio-recognition elements

Three various strategies for anchoring bio-recognition elements to the semiconductor were developed. C reactive protein (CRP) was chosen as a target molecule for the final application. The simplest immobilization strategy with a high binding of avidin and antibodies to CRP based on physisorption of the bio-recognition molecules and a post-treatment with a blocking polymer to reduce non-specific binding was chosen for the biosensor implementation.

Task 2.1. Selection, modification or cleavage of recognition molecules

C reactive protein (CRP) was selected for the final target application. Antibodies to CRP were digested to F(ab´)$_2$-fragments and purified (VTT).

Task 2.2. Biotin immobilization on the EGOFET surface

In strategy #1, the commercially available poly(3-carboxypropylthiophene) (P3CPT) was activated by *N*-hydroxy-succinimide and 1-ethyl-3-(3-dimethylaminopropyl)carbodiimide (NHS/EDC) chemistry at VTT and both biotin and maleimide groups were bound to the layers to enable a binding of avidin. A cysteine modified chimeric avidin (ChiAvd-Cys) was also bound through maleimide to the surface. Avidin with monolayer surface coverage (430 - 470 ± 30 ngcm^{-2}) could be bound to the P3CPT-COOH activated surface. Additionally, VTT studied the possibility to immobilize ChiAvd-Cys directly by adsorption onto poly(3-hexylthiophene) (P3HT) and blocking with the polymer N-[tris(hydroxy-methyl)methyl]acrylamide - lipoic acid conjugate (pTHMMAA). Also, a new method based on photobiotin was studied by ACREO and evaluated. These two new approaches would be more compatible with the demands for printing of the avidin-antibody layer, by reducing the amount of immobilisation steps. An activation of the surface by NHS/EDC chemistry is too complicated when layers are to be printed on the semiconductor surface.

In order to enable avidin coupling to the semiconductor surface, new functionalized polythiophenes that allow the grafting of biotin was synthesised at UPD (strategy #2). The biotin groups were, however, not available for a specific binding to the surface, but avidin was non-specifically adsorbed on the surface. The most promising result was obtained by a combination with strategy #1, an additional treatment of the co-polythiophene surface with a short hydrophilic blocking molecule.

Biotinylated phospholipids were covalently anchored to P3HT-COOH and avidin were coupled corresponding to 150 ± 20 ng cm^{-2} were coupled to the layers (UNIBA - strategy #3).

Task 2.3. Immobilisation of antibodies

Synthesis of non-ionic hydrophilic polymers to be used as bio-repellent molecules (VTT)

Various lipoate conjugates were synthesised by VTT (N-[tris(hydroxy-methyl)methyl]acrylamide - lipoic acid conjugate (pTHMMAA), Lipa-biotin linkers, Lipa-DEA and Lipa-TRIS) and used at VTT and UPD for reducing a non-specific binding to the surface and at VTT for coupling avidin to the surface.

Procedure (1) – Attachment of antibodies and polymer to the semiconductor surface (VTT)

Due to the need for several activation steps to anchor antibodies on polythiophene via carboxyl groups, a direct physisorption of the antibodies to the semiconductor surface and a blocking with pTHMMAA was evaluated and turned out to give very high specific binding of the target molecule CRP.

Task 2.4. Characterisation of immobilised bio-recognition elements

The surfaces were characterised by contact angle measurements, surface plasmon resonance (SPR) and AFM at VTT in order to determine the hydrophobicity, surface density and topography of the molecular layers. The phospholipid layers were also characterized by fluorescence microscopy by UNIBA.

WP3: Biosensors Implementation

The main objective of WP3 was to develop protocols for the fabrication of biosensors using the three strategies to anchor the bio-recognition elements. For all the biosensors the electrical and analytical performances were evaluated. Implementation of sensors for CRP target analyte detection was also investigated. The compatibility of the fabrication processes with printing techniques was explored as well.

Task 3.1. Fabrication of the Bio-EGOFET with biotin recognition element

Sensors were fabricated using the device geometry developed by LIU in WP1 including interdigitated gold source and drain electrodes with different channel lengths and widths. The organic semiconductor was functionalized with biotin using two procedures developed by UPD and UNIBA. In the first one, sensors included an organic semiconductor bearing biotin groups. Several strategies were followed to increase the hydrophilicity of the polymer in order to decrease non-specific interactions.

The second approach consisted of sensors with biotinylated phospholipid layers covalently attached to P3HT-COOH surface. Either a P3HT-COOH15 copolymer synthesized by UPD or a radio frequency (RF, 13.56 MHz) plasma enhanced chemical vapour deposition (PE-CVD) process was used to attach carboxylic acid functionalities on P3HT. The biotinylated phospholipids were covalently attached to the carboxyl groups through the formation of amidic bonds using the EDC/NHS chemistry.

Both sensors show good electrical performance, with a field-effect mobility of 10^{-3} cm^2/Vs and a current on-off ratio around 10^2, in good agreement with what found in the literature.

Task 3.2. Measurements of Bio-EGOFET responses to model avidin solutions

Sensors following the strategies described above were checked by analysing their response to (strep)avidin solutions and evaluating their analytical figures of merit. The measurements were made in (strept)avidin solutions of concentration ranging from 1 µg/ml (0.016 μM) to 100 µg/ml (1.6 μM). A *decrease* of the drain current was observed after the addition and incubation of streptavidin solutions on sensors fabricated with biotin directly bound to P3HT, while an *increase* of the drain current occurred on sensors with biotin embedded in a phospholipid layer.

The difference in the biosensors response can be explained by assuming that a different mechanism is involved in the transduction mechanism of the bio-recognition process. However, in both cases, the current change (decrease or increase) follows the variation of the streptavidin concentrations. Importantly, the streptavidin detection limit was much lower than the micro-molar range targeted by the project. Control measurements with HSA (human serum albumin) or BSA (bovine serum albumin), which are proteins not recognized by biotin, also demonstrated a good specificity of the sensors.

The main drawback of these sensors is a low reproducibility. This point is still under investigation. Reproducibility could be improved by employing a microfluidic system to perform the electrical measurements.

The compatibility with a printing process is another important issue. While the strategy with biotin directly bound to P3HT can be easily implemented with a printing process, sensors with biotin embedded in a phospholipid layer are difficult to make by printing techniques because of the functionalization step with carboxyl groups and subsequent activation with NHS/EDC chemistry to allow a stable deposition of biotinylated phospholipids layers.

Task 3.3. Fabrication of biosensors with the bio-receptors identified in Task 2.1

Sensors with antibodies to CRP physisorbed on both P3HT-COOH15 and P3HT were selected to evaluate the response to CRP. This immobilization strategy, suggested by VTT, reduces the number of activation steps needed to anchor the antibodies on P3HT. Moreover, this approach is compatible with printing techniques. After deposition of the anti-CRP, the organic semiconductor surface was post-treated with pTHMMAA polymer to reduce non-specific bindings. The sensor response to CRP was evaluated with solutions at concentrations ranging from 50 µg/ml to 0.1 µg/ml in PBS 10 mM. Devices fabricated with the organic semiconductor only treated with pTHMMAA or functionalized with the anti-hIgG (an antibody nonspecific for the CRP protein) and treated with pTHMMAA were used as reference (negative controls). A decrease of the drain current and a shift of the threshold voltage after exposure to CRP solutions at different concentrations were observed for biosensors functionalized with the anti-CRP antibody, while no change of the current occurred when the antibodies to CRP were not present on the P3HT surface, thus confirming the biosensor specificity.

All the CRP calibration curves of the biosensors fabricated using the P3HT polymeric semiconductor are given in the Figure below. Note that a sub µM detection limit was also achieved with this biosensor.
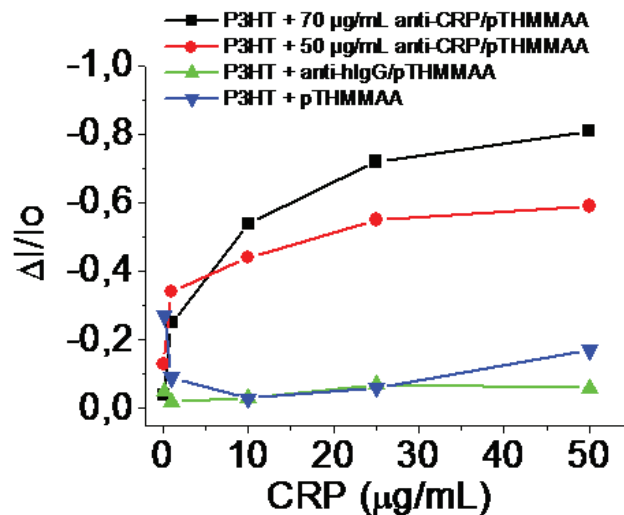
Figure 3.1. Calibration curves for CRP at different concentrations for sensors with P3HT functionalized with 70 µg/mL anti-CRP (black curve), 50 µg/mL anti-CRP (red curve), only pTHMMAA (blue curve) and anti-hIgG (green curve).

The reproducibility of the biosensors with antibodies placed directly on the organic semiconductor surface would require further verification. However, these biosensors were selected for implementation with printing techniques as described in WP5.

All the deliverables and milestones planned for WP3 were fulfilled and completed.

WP4. Modelling and testing

This work package was divided into five tasks. Tasks 4.1 to 4.3 were devoted to testing the final sensors at different level: Test against avidin in differential circuits (task 4.1), test of the device stability (task 4.2) and test of devices for final application (task 4.3). Because of redundancy of these tasks with objectives in other work packages, it was decided to transfer these tasks to the appropriate WP, namely WP3 for task 4.2 and WP5 for tasks 4.3. Note that no work were done with task 2.1.

The remaining tasks dealt with modelling of the bioEGOFET biosensor. This was divided into two tasks, 4.4 (modelling the electrolyte-gated organic transistor) and 4.5 (modelling the transduction mechanism).

Task 4.4. Modelling the electrolyte-gated organic transistor

Most organic field-effect transistors work in the *accumulation* (aka *enrichment*) mode, where one type of the charge carriers (electrons or holes) accumulate at the semiconductor/ insulator interface under the action of the electric field generated by the voltage applied between the gate and the source electrodes, as illustrated in Fig. 4.1 for a p-channel device.
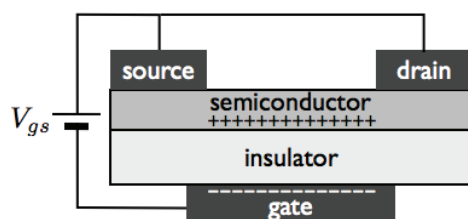

Figure 4.1. Formation of the conducting channel in an organic field-effect transistor.

Because the density of charge in the conducting channel is proportional to the capacitance of the insulator, low-voltage operation requires high capacitances, which can be done by using an electrolyte instead of a dielectric. Upon applying an appropriate voltage to the gate, an electrical double layer forms at both the gate and the semiconductor interfaces (Fig. 4.2). The latter provides the aforementioned conducting channel for field-effect operation.
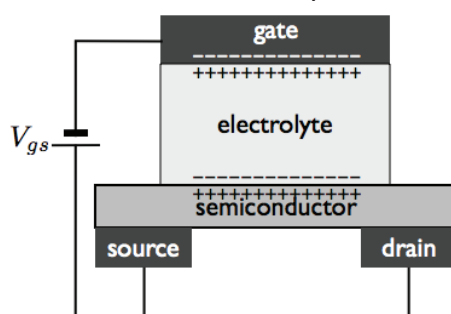
Figure 4.2. The conducting channel in an electrolyte-gated organic transistor. Note that an electrical double-layer forms at both the semiconductor / electrolyte and gate / electrolyte interfaces.

However, there exists another kind of organic transistor using an electrolytic solution: The organic electrochemical transistor (OECT), which takes advantage of the ability of many conjugated polymers to electrochemically dope and undope, thus reversibly passing from a conducting to an insulating state. These devices are characterised by the fact that ions from the electrolyte penetrate into the semiconductor during the operation of the transistor.

The work achieved for task 4.4 mainly consisted of a comprehensive electrochemical characterisation of the semiconductor-electrolyte interface (UPD, LIU). The measurements were mainly done is a three-electrode configuration, including the semiconductor layer as a working electrode, a reference electrode and a platinum counter-electrode. Cyclic voltammetry (CV) and electrochemical impedance analysis (EIS) were the characterisation tools.

All CV data clearly indicated that the device operates a in the stability domain of the aqueous electrolyte, when no electrochemical reaction occurs. This led us to the conclusion that the electrolyte-gated transistors operate in the field-effect regime, and not in the electrochemical regime.

Of particular interest were the EIS data. They were analysed in terms of equivalent circuit. In all cases, the data could be fitted to a circuit that comprised a small resistance in series with one or two capacitances, each of which was in parallel with a large resistance. The small resistance could be interpreted as coming from the contacts to the external electrical circuit. The capacitance was attributed to the electrical double-layer (EDL) that forms at the semiconductor-electrolyte interface and plays the role of the gate dielectric in the electrolyte-gated transistor. The origin of a second capacitance could not be clearly elucidated. It was decided to discard the measurements showing a second parallel capacitance-resistance system, so that the EDL capacitance could be unambiguously determined. In some occasions, the capacitance was fitted to a constant phase element (CPE), which resulted in a better matching between the experimental and calculated data. However, in all cases the parameter n of the CPE was close to one, thus indicating that the element behave much like a capacitance. In any cases, the general trends were identical, whichever the circuit element used (capacitance or CPE).

The EDL capacitance was found to be strongly dependent on the hydrophobicity of the semiconductor surface. Poly(3-hexylthiophene) (P3HT) is a highly hydrophobic material, and the corresponding EDL capacitance was found in the order of 1 $\mu F/cm^2$, ten times lower than that of the bare metal electrode. Substituting the thiophene rings with a carboxyl group considerably increases the hydrophilicity of the polymer. This also results in a substantial increase of the EDL capacitance (by a factor of 100). We interpreted this trend in terms of wettability of the polymer surface. Substitution by biotinylated groups, as done when immobilising the bio-recognition element of the sensor, also leads to an increase of the hydrophilic character of the polymer surface, and the associated increase of the EDL capacitance.

The model of the EGOFET also included a description of the potential distribution in the device. Unlike the convention FET, the EGOFET contains two EDLs, one at the semiconductor-electrolyte interface and the second one at the gate-electrolyte interface. The voltage of the electrolytic solution is controlled by the respective values of the capacitance of these two EDLs. If the capacitance of at the gate is much higher than that at the semiconductor, then the solution potential is close to the gate potential, so that the device operates much like a conventional OFET (that is, the charge in the conducting channel is fully controlled by the bias applied to the gate electrode.) On the other hand, if the capacitance at the semiconductor is much higher than that at the gate, the solution potential comes close to the semiconductor potential and the gate voltage cannot be used to control the charge in the conducting channel. In that case, the current that flows in the transistor is very little affected by the gate voltage. Instead, it can be changed by acting directly on the solution potential. Such an operating mode has already proved its usefulness in biosensors based on organic electrochemical transistors. However, in the frame of this project, the model has not been experimentally checked and all the sensors reported worked in the field-effect regime.

The work done in task 4.5 also included EIS measurements on semiconductor equipped with a phospholipid layer (UNIBA). The equivalent circuit in this case is much more complex. Its various elements could be associated either to the counter-electrode EDL or to the phospholipid layer. Interestingly, no evidence was found for an EDL at the semiconductor surface, which can be explained by the fact that the semiconductor surface is isolated from the electrolyte by the phospholipid layer.

Task 4.5. Modelling the transduction mechanism

We have analysed and compared various conceivable origins for the change of the drain current upon contact of the analyte by the modified semiconductor surface. Namely, the decrease of the the drain current might come from (1) an electrochemical reaction at the semiconductor surface, (2) an electrochemical charging of the interface due to charges localised on the analyte, or (3) a change of the EDL capacitance upon attachment of the analyte on the bio-recognition element. The likelihood of conjectures (1) and (2) is poor because (1) the sensor operates within the stability domain of the aqueous electrolyte, when no electrochemical reaction takes place at the semiconductor electrode surface and (2) proteins are known to bear only very small electric charges.

Conjecture (3) was checked by UPD through a complete EIS analysis of sensors before and after analyte recognition.

The electrode consisted of a P3HT layer spin-coated on an ITO substrate. The electrolyte was 1 mM PBS. The EIS analysis was made at the following steps:

1. Bare electrode
2. Attachment (by physisorption) of the recognition element
3. Passivation of the sensor
4. Recognition of the analyte

In all cases, the recognition element was lipa-biotin. The passivation of the sensor to non-specific binding consisted of attaching a hydrophilic element on the surface. Two elements were tried: Human serum albumin (HSA) and lipa-tris. A much better selectivity is obtained in the latter case. At the measured decrease of the drain current was 49% for streptavidin, 40% for avidin and only 15% for HSA when the sensor is passivated with lipa-tris, thus demonstrating a good selectivity. For sensors passivated with HSA, the respective values of the decrease are 16, 10 and 13%, thus indicating a poor selectivity of the sensor in this case.

EIS measurements were made in a three-electrode configuration. All data could be fairly fitted to an equivalent circuit made of a resistor in series with a constant phase element (CPE) in parallel with a second resistor.

In all our measurements, the exponent factor n was 0.95±0.02, thus indicating that the value of Q can reasonably be interpreted in terms of a "non-ideal" double-layer capacitance. With the best sensor selected (lipa-biotin / lipatris) we detected a decrease of the EDL capacitance by 3% with HSA and 9% with avidin. Although the amplitude of the variation is not quantitatively in line with the corresponding variation of the drain current, we observe a clear decrease of the capacitance when the analyte is a target for the immobilised bio-recognition element.

WP5. All-printed bioEGOFET sensor

Task 5.1. Printable materials

Materials for the realization of the sensors were identified and evaluated. Tentative experimental tests were carried out for some of them. Focus was set on functionality but cost is also considered.

Substrates. Important parameters for substrates are smoothness, resistance to chemicals and heat. Both Teonex and Upilex were selected for further development. Upilex is more resistant to heat, while Teonex is cheaper and non-coloured, which makes processing easier.

Source, drain and gate materials. We have tentatively investigated Ag, PEDOT:PSS and Au. The latter was selected for the final demonstrator because the process was most mature and evaporated Au is used in the other WPs. Printed Ag and PEDOT are cheaper and require lower curing temperature and are therefore interesting candidates for future developments.

Semiconductor. P3HT was selected for final developments since the process was mature and the material commercially available. Other semiconductors tested were biotinylated P3HT, P3HT-COOH15 (both synthetized at UPD) and P3CPT.

Electrolyte. DI water and PBS were selected and used also in other WPs.

Task 5.2. Process developments

Printing processes for a wide range of materials were developed. The level of maturity varies. For some materials printability was evaluated, for others further developments were made.

Printability was confirmed for semiconductors synthesized at UPD as well as for the commercially available P3CPT. A special technique for patterning P3HT was developed, in which an SU8 well is printed around the channel area. When P3HT is printed inside the well, SU8 attracts it, resulting in a thin and smooth P3HT layer in the middle of the well, where the channel is. This is important because thick and non-uniform layers are believed to decrease performance and stability.

Ag, Au and PEDOT electrodes were printed by inkjet. All electrode materials were also overprinted with P3HT and transistor characterizations made using DI water as electrolyte. A modulation was observed in all structures but processes were only optimized further for Au/P3HT. Characterization data for an EGOFET with printed Au electrodes (UTDAu25TE) and printed P3HT is shown in Figure 5.1. The channel length was 10 µm. Unfortunately, we could not reproduce the result because the next Au ink batch did not perform well. Though commercial, the product is seemingly not well controlled. It should be noted that the P3HT was printed without SU8 in this case, so the device can be further improved through the technique developed later.
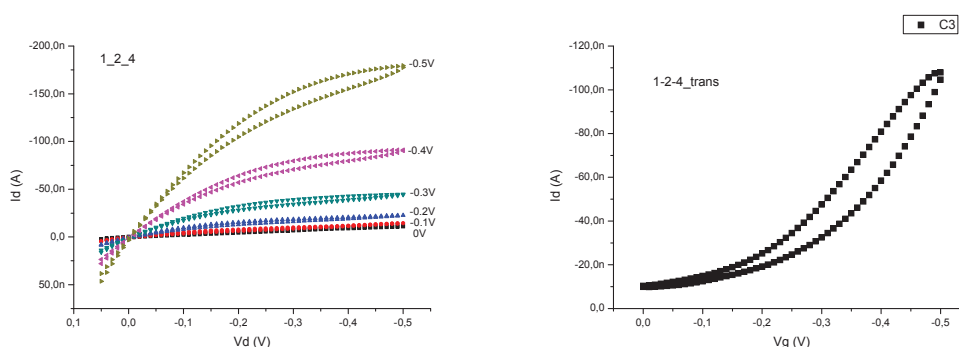

Figure 5.1 Output (left) and transfer (right) curves for an all-printed transistor (UTDAu25TE/P3HT).

Bioreceptor materials and blocking molecules were also printed. Best results were achieved with an automatic dispensing platform. ChiAvd-Cys was mixed with glycerol and PBS and printed (spotted) onto spin-coated P3HT. SPR characterizations showed high binding of the target molecule CRP.

The sensor that was selected for the realization of the final demonstrator is based on physisorbed anti-CRP. Time did not allow for the development of an anti-CRP ink. It was confirmed, though, that anti-CRP loses its activity when mixed with glycerol. One idea is to replace glycerol by sugar.

A printing process for rational production of the fluidics was developed (see Task 5.3). A screen printable adhesive (SP7555, SP4533 3M) is used to pattern the fluidics and also for lamination of the structure. The concept was successfully evaluated.

Several strategies for immobilization of the bio-recognition element were investigated in the project. In WP5 these were evaluated from a printability point-of-view. Important parameters are number of printing steps required, incubation time, compatibility with inks and printing processes. The idea to use functionalized semiconductors is interesting since we then print both semiconductor and receptor in one step. For very long immobilization times we must consider dip coating, which may not allow for reel-to-reel manufacturing (but which may still be a rational process). Moreover, some of the immobilized layers decreases the transistor current very much and are therefore not a good choice. Logically, the receptor performance was also taken into account and the decision was to use physisorbed anti-CRP on P3HT. Time did not allow for the development of an anti-CRP ink, see above.

Task 5.3. Design and realization

Different approaches to fluidics were evaluated. Both fluidics for the final demonstrator and for the developmental work was considered. In the former case, it is important to find a solution suitable for rational fabrication and for analyses carried out within minutes. In the latter case, we need a solution that is easy to assemble for all partners (also those who have not access to printing equipment) and suitable for long-time measurements, say 1 h. We have tested flow channels of different plastic materials combined with either SU8, double-sided adhesive (MACTAC) or a screen printable adhesive (SP7555, MS) to confine the analyte reservoir and flow channel. SU8 is difficult to laminate and is hydrophobic. MACTAC works better, and is easily applied, but during long-time exposures we believe that the liquid penetrates the adhesive layer to some extent. Best results were achieved for the screen printable adhesive SP7555, which is more resistant to liquids than the double-sided tape. It is also transparent and easily laminated to the top layer of the structure. During our experiments we observed slow or interrupted flow in some of the test structures. The project reviewers advised us to evaluate the flow properties in an intermediate structure with a functionalized semiconductor surface. A test was made in order to confirm that flowing liquids into our selected sensor system would not present any problem. Especially, we wanted to confirm that the structure would allow for liquid flow after functionalization of the semiconductor area. A structure using materials that have shown to inhibit flow of DI water (P3HT, teonex, evaporated Au) was made. pTHMMAA was incubated on the P3HT surface. Successful low tests with AZO-colored DI water and PBS were made (Figure 5.2). Also in structures without pTHMMAA flow was confirmed but it was not as fast as in the incubated structure.
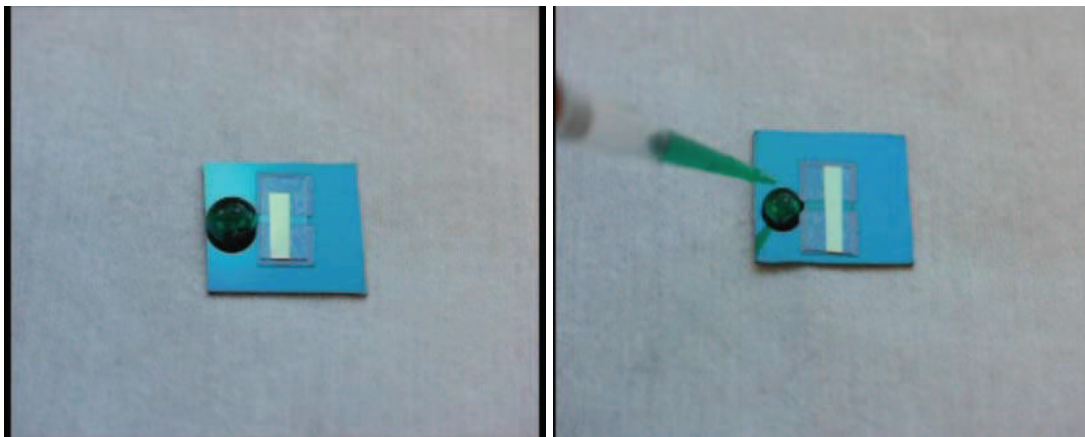
Figure 5.2. Flow test through Teonex/Au and P3HT/SiO2 without (left) and with (right) pTHMMAA incubated on the P3HT. The green AZO colored DI drop has not yet flown through the structure without pTHMMAA after > 1 minute (left). The green AZO colored DI drop immediately flow through the structure with pTHMMAA (right).

Poor stability was observed in the printed EGOFETs. We believe that this among other is due to degradation of the P3HT semiconductor, which results in a decrease in the drain current. Since bio-sensing measurements are based on changes in the drain current associated with the bio-recognition process, the devices must be enough stable to allow for sufficient resolution in the sensor measurements. Also, the on/off is much lower in printed than in spin-coated devices. The printed semiconductor is comparatively thicker (100 nm against 30 nm) and not as smooth as spin coated films. Different materials, treatments and process parameters were evaluated in order to improve the stability and performance of the EGOFET devices. The most important achievement was a new technique for printing P3HT, which results in smooth and thin layers. An SU8 well was first printed to confine the channel area. When P3HT is printed inside the well, it is attracted by SU8 and only a thin and smooth layer of the semiconductor is left over the channel area. Best stability was achieved in sensors printed on Upilex, that is, sensors printed on SiO_2 were less stable. Pre-treatments with ozone before printing P3HT and post-treatment with hot plate or pTHMMAA did not improve stability.

A few of the transistors evaluated in the investigation described above were functionalized with anti-CRP and bio-sensing measurements were made with CRP. The printed P3HT was functionalized with anti-CRP antibody at 70 µg/mL and post-treated with a pTHMMAA polymer. Devices fabricated with P3HT treated only with pTHMMAA were used as reference. The most promising result was obtained for a sensor fabricated on Teonex (evaporated Au electrodes, printed P3HT). Figure 5.3 shows the output curves for sensors with the antibody physically adsorbed on P3HT and for EGOFETs only treated with pTHMMAA. In the right hand side of each panel, the relevant responses to CRP are evaluated with protein solutions at concentrations from 0.1 µg/ml to 50 µg/ml in PBS 10 mM.
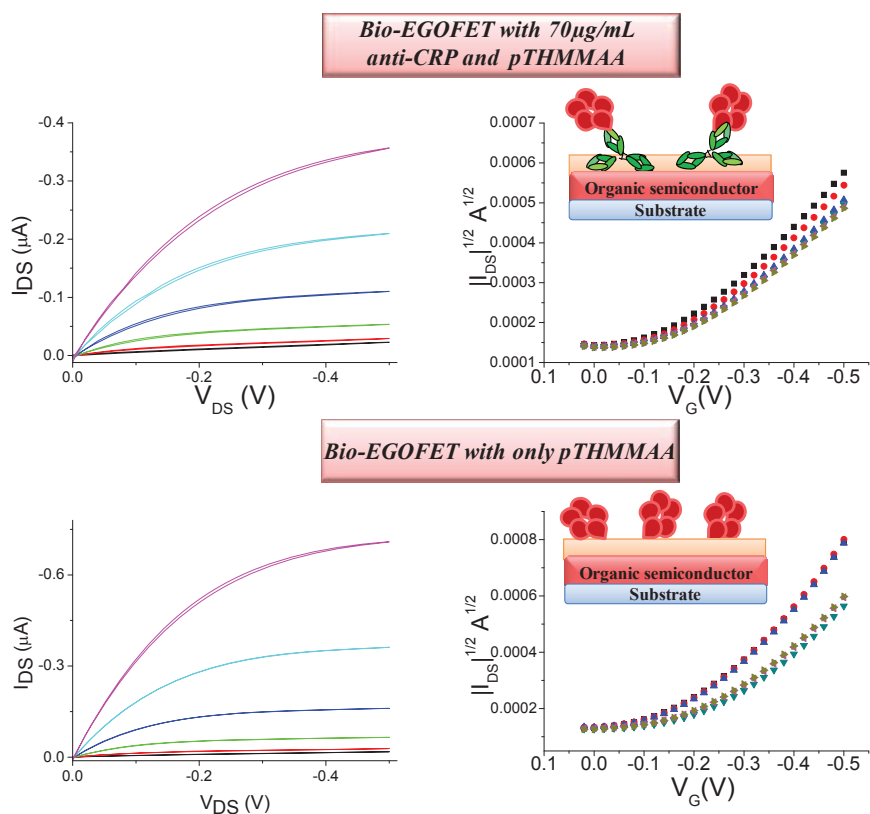
Figure 5.3. Left: Output curves of a biosensor fabricated on Teonex substrate with printed P3HT. Right: Response to CRP solutions at different concentrations in PBS 10 mM (black), 0.1 µg/ml (red), 1 µg/ml (blue), 10 µg/ml (green), 25 µg/ml (pink) and 50 µg/ml (brown).

Rational fabrication methods for all the process steps necessary for the fabrication of BioEGOFET sensors were proven. Also, we have observed a sensor response to CRP in a functionalized sensor with a printed semiconductor (P3HT). The response (ΔId) was proportional to the concentration of CRP in the analyte. We have, however not succeeded in fabricating an all-printed functional sensor, and accordingly circuitry was not developed. We believe that the realisation of an all-printed structure is only a matter of process development and optimization, that is, we see no fundamental problems. The main reasons for not achieving the goal are (1) the selection of immobilization approach and thereby also the selection of semiconductor were delayed and (2) though commercial, the selected gold ink for electrodes unfortunately was not a stable product giving reproducible results. Remaining tasks are to optimize the P3HT printing process in order to achieve more stable devices with better on/off and also to modify the anti-CRP ink so that the activity does not decrease, that is, glycerol must be replaced with another substance that prevents the ink from drying during incubation.

1.4 Potential impact

Biosensors market

The global market for biosensors is forecast to reach US$12 billion by the year 2015. Growth in population and increase in number of people getting affected with various health issues such as diabetes and obesity is driving the need for periodic medical care, in turn propelling market growth of biosensors in medical diagnostics. Expanding medical applications, increase in R&D activities, and emergence of new technologies such as microfluidics, and non-invasive biosensing alternatives are the other market propellers. In addition, key factors driving growth for glucose biosensors include growing diabetic population, user-friendly designs, and increased point-of-care applications. Although glucose monitoring dominates the market because it is used in both point-of-care (POC) and home diagnostics applications, the demand for POC devices is increasing, and there is a need for a new technology to be developed to meet market needs in term of wider range of disease marker detection, multiple tests, better specificity and sensitivity and critically, the possibility to perform quantitation at low cost. In fact, the potential for the development of nanotechnology-enabled POC diagnostic devices is high. Their increasing availability and use could have significant patient benefits and cost-saving implications for health services. Some of the most important challenges for nanomedicine currently include:

- Finding ways to translate advances in both technology and the understanding of the pathology of disease into preventive medicine;
- Detecting disease at the earliest, most treatable stage;
- Decreasing costs to healthcare services and increasing productivity at the same time;
- Ensuring the cost of medical technologies is aligned with better patient prognosis;
- Finding ways of improving on existing treatments and technologies.

Potential impact of the project

The sensor technology proposed in the BioEGOFET project can readily be adapted and indeed, would be ideal for these market requirements. The main goal of this research program is to produce a biosensor system that is more robust, reliable and user-friendly than current approaches and which will have a very wide range of applications due to the nature of the detection scheme, making new applications much quicker and cheaper to develop. It should be emphasized that the flexibility of the device proposed renders it suitable for isolated, individual, hand-held, portable use. On the other hand electronic output allows for easy quantitative analysis with possibility of interfacing with a computer. The implementation of biosensors on flexible and paper substrates besides to allow for low-cost fabrication could also be in principle compatible with recycling processes

It is clear that a hypothetical launch of a BioEGOFET product on the biosensor market would have a huge impact on the diagnostic POC field.

In addition, many other important economic areas, including medicine, biodefense, forensic science, environmental monitoring and food industry, could benefit by the "BioEGOFET" technology.

2. Use and dissemination of foreground

Section A (public)

Part A1

TEMPLATE A1: LIST OF SCIENTIFIC (PEER REVIEWED) PUBLICATIONS, STARTING WITH THE MOST IMPORTANT ONES

NO.	Title	Main author	Title of the periodical or the series	Number, date or frequency	Publisher	Place of publication	Year of publication	Relevant pages	Permanent identifiers[1] (if available)	Is/Will open access[2] provided to this publication?
1	A water-gate organic field-effect transistor	L. Kergoat	Advanced Materials	Vol 22, Issue 23	Wiley		2010	2565-2569	DOI: 10.1002/adma.20 0904163	No
2	Advances in organic transistor-based biosensors: from organic electrochemical transistors to electrolyte-gated organic field-effect transistors	L. Kergoat	Analytical and Bioanalytical Chemistry	Vol 402, N°5	Springer		2011	1737-1856	DOI 10.1007/s00216-011-5363-y	No
3	DNA detection with a water-gate organic field-effect transistor	L. Kergoat	Organic Electronics	Vol 13, Issue 1	Elsevier	Online	2012	1-6	http://dx.doi.org/10.101 6/j.orgel.2011.09.025	No

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[1] A permanent identifier should be a persistent link to the published version full text if open access or abstract if article is pay per view) or to the final manuscript accepted for publication (link to article in repository).

[2] Open Access is defined as free of charge access for anyone via Internet. Please answer "yes" if the open access to the publication is already established and also if the embargo period for open access is not yet over but you intend to establish open access afterwards.

#	Title	Author	Journal	Vol/Issue	Publisher		Year	Pages	DOI/URL	
4	Synthesis and characterization of all-conjugated copolymers of 3-hexyl-thiophene and EDOT by grignard metathesis polymerization	L.Miozzo	Journal of Polymer Science Part A: Polymer Chemistry	Vol 50, Issue 3	Wiley	Online	2011	534–541	DOI: 10.1002/pola.25062	No
5	Use of poly(3-hexylthiophene)/poly(methyl methacrylate) (P3HT/PMMA) blends to improve the performance of water-gated organic field-effect transistors	L. Kergoat	Organic Electronics	Vol 12, Issue 7	Elsevier		2011	1253-1257	http://dx.doi.org/10.1016/j.orgel.2011.04.006	No
6	Amphiphilic conjugated block copolymers for efficient bulk heterojunction solar cells	C. Suspène	Journal of. Materials. Chemistry	Vol 22, Issue 10	RSC		2012	4511–4518	DOI:10.1039/C2JM14960E	No
7	Phospholipid film in electrolyte-gated organic field-effect transistors.	S Cotrone	Organic Electronics	Vol 13, April 2012	Elsevier	Online	2012	638-644	http://www.sciencedirect.com/science/article/pii/S1566119912000201	No
8	Cysteine-tagged Chimeric Avidin forms High Binding Capacity Layers Directly on Gold	I. Vikholm-Lundin	Sensors & Actuators B		Elsevier		In press			
9	Structural and functional characteristics of chimeric avidins physically adsorbed onto poly-3-hexylthiophene thin films	W. M. Albers					To be submitted			

Part A2

TEMPLATE A2: LIST OF DISSEMINATION ACTIVITIES

NO.	Type of activities[3]	Main leader	Title	Date	Place	Type of audience[4]	Size of audience	Countries addressed
1	Conference	UPD	E-MRS Spring Meeting	17 May 2012	Strasbourg, FR	Scientific Community	More than 500	International
2	Conference	VTT	E-MRS Spring Meeting	17 May 2012	Strasbourg, FR	Scientific Community	More than 500	International
3	Conference	UNIBA	EMRS Spring Meeting	17 May 2012	Strasbourg, FR	Scientific Community, Industry	More than 500	International
4	Conference	ACREO	EMRS Spring Meeting	17 May 2012	Strasbourg, FR	Scientific Community, Industry	More than 500	International
5	Conference	UNIBA	Biosensors 2012	15-18 May 2012	Cancun, MX	Scientific Community, Industry	More than 500	International
6	Conference	UNIBA	MRS 2012	9-13 April 2012	San Francisco, CA	Scientific Community, Industry	More than 500	International

[3] A drop down list allows choosing the dissemination activity: publications, conferences, workshops, web, press releases, flyers, articles published in the popular press, videos, media briefings, presentations, exhibitions, thesis, interviews, films, TV clips, posters, Other.

[4] A drop down list allows choosing the type of public: Scientific Community (higher education, Research), Industry, Civil Society, Policy makers, Medias ('multiple choices' is possible.

#	Type	Org	Title	Date	Location	Audience	Participants	Scope
7	Presentation	VTT	Functionalization of Surfaces for Molecular Recognition	15 Feb 2012	Tampere, FI	Higher education	30	National
8	Company day	VTT	Sensors & Coatings	12 Jan 2012	Tampere, FI	Industry	25	International
9	Conference	UPD	14th International Union of Pure and Applied Chemistry Conference on Polymers and Organic Chemistry (IUPAC POC 2012)	6-9 January 2012	Doha, Quatar	Scientific Community	More than 500	International
10	Conference	UPD	14th International Union of Pure and Applied Chemistry Conference on Polymers and Organic Chemistry (IUPAC POC 2012)	6-9 January 2012	Doha, Quatar	Scientific Community	More than 500	International
11	Workshop	ACREO	2nd workshop of the French Organic Electronics Network (CNRS-GDR Electronique Organique)	November 2011	Grenoble, FR	Scientific community	100	National
12	Exhibition	UNIBA	Researchers' Night 2011	23 September 2011	Bari, IT	Civil Society	More than 1000	National
13	Conference	UNIBA	XXIV National Conference of the Italian Chemical Society	11-16 September 2011	Lecce, IT	Scientific Community, Industry	More than 400 participants	National
14	Presentation	UPD	4th International Symposium on Flexible Organic Electronics (ISFOE11)	July 2011	Thessaloniki, GR	Scientific Community, Industry	200	International

#	Type	Org	Title	Date	Location	Audience	Number	Scope
15	Conference	UNIBA	ICOE 2011	22-24 June 2011	Rome, IT	Scientific Community, Industry	More than 200	International
16	Conference	UNIBA	GS2011	15-17 June 2011	Teramo, IT	Scientific Community, Industry	More than 200	National
17	Seminar	ACREO		5 April 2011	Stockhlom, SE	Research, municipal, industry		National
18	Web, press release	UNIBA	Apulia Innovation website	January 2011	Bari, IT	Civil Society, Scientific Community		National
19	Web, press release	UNIBA	Official website of the University of Bari "Aldo Moro"	23 December 2010	Bari, IT	Civil Society, Scientific Community		National
20	Article	UPD	Optics Valley's newsletter N°50	October 2010	Paris, FR	Scientific Community, Industry		National
21	Conference	UNIBA	GS2010	26-28 October 2010	Firenze, IT	Scientific Community, Industry	More than 200	National
22	Conference	UNIBA	XXII National Conference of the Analytical Chemistry Division of the Italian Chemical Society	12-16 September 2010	Como, IT	Scientific Community, Industry	More than 200	National
23	Conference	UNIBA	SPIE 2010	3-5 August 2010	San Diego, CA	Scientific Community, Industry	More than 500	International

24	Conference	UNIBA	3rd International Symposium on Flexible Organic Electronics	6-9 July 2010	Halkidiki, GR	Scientific Community, Industry	More than 300	International
25	Conference	UNIBA	International Symposium on Industrial Electronics, ISIE	4-7 July, 2010	Bari, IT	Scientific Community, Industry	More than 300	International
26	Conference	UNIBA	International Conference on Organic Electronics	23-25 June 2010	Paris, FR	Scientific Community, Industry	More than 200	International
27	Conference	UPD	International Conference on Organic Electronics	23-25 June 2010	Paris, FR	Scientific Community	More than 200	International
28	Conference	UNIBA	Fπ9 - International Symposium on Functional π-Electron Systems	23-28 May 2010	Atlanta, GA	Scientific Community, Industry	More than 500	International
29	Conference	UNIBA	Materials Research Society (MRS) Spring Meeting	5-9 April 2010	San Francisco, CA	Scientific Community, Industry	More than 500	International
30	Article	UPD	Plastic Electronics	March 2010	Paris, France	Civil Society, Scientific Community		International
31	PhD Thesis	UNIBA Author: Cotrone S.	Multilayer structures of biomolecules immobilized on organic semiconductors for advanced analytical devices	Dottorato di Ricerca in Scienze Chimiche XXIV CICLO	Bari, IT	Scientific Community		National

32	Bachelor Thesis	UNIBA Author: Di Benedetto B.	Integrazione di strati lipidici in dispositive OTFT per lo sviluppo di biosensori elettronici	Bachelor degree in Biotechnology	Bari, IT	Scientific Community		National
33	Bachelor Thesis	UNIBA Author: Del Rosso M.	Implementazione di strati lipidici in dispositivi OFET per future applicazioni in sensori: caratterizzazione elettrica ed elettrochimica	Bachelor degree in Materials Science	Bari, IT	Scientific Community		National
34	Workshop	ACREO	MSW Micronano System Workshop 2010		Stockhlom, SE		
35	Conference	ACREO	BIOTECH FORUM Biotech & Medtech Partnering		Copenhagen, DK		