Flexible and Printed Electronics

LETTER

Large-area printed organic electronic ion pumps

Dennis Cherian 1, Astrid Armgarth 1, Valerio Beni 2, Ulrika Linderhed 1, 2, Klas Tybrandt 1, David Nilsson 2, Daniel T Simon 1 and Magnus Berggren 1

1 Laboratory of Organic Electronics, Department of Science and Technology, Linköping University, Norrköping, Sweden
2 Department of Printed Electronics, RISE Acreo, Research Institute of Sweden, Norrköping, Sweden
E-mail: daniel.simon@liu.se

Keywords: bioelectronics, drug delivery, organic electronics, electrophoresis, printed electronics

Abstract

Biological systems use a large variety of ions and molecules of different sizes for signaling. Precise electronic regulation of biological systems therefore requires an interface which translates the electronic signals into chemically specific biological signals. One technology for this purpose that has been developed during the last decade is the organic electronic ion pump (OEIP). To date, OEIPs have been fabricated by micropatterning and labor-intensive manual techniques, hindering the potential application areas of this promising technology. Here we show, for the first time, fully screen-printed OEIPs. We demonstrate a large-area printed design with manufacturing yield >90%. Screen-printed cation- and anion-exchange membranes are both demonstrated with promising ion selectivity and performance, with transport verified for both small ions (Na +, K +, Cl −) and biologically-relevant molecules (the cationic neurotransmitter acetylcholine, and the anionic anti-inflammatory salicylic acid). These advances open the ‘iontronics’ toolbox to the world of printed electronics, paving the way for a broader arena for applications.

1. Introduction

Unlike electronic systems, biological systems utilize ions and molecules for signaling, transmission of information, and regulation of health status. Thus, emerging and future technologies for interfacing biology—such as therapeutic implants, functional prostheses, or ‘smart’ wound dressings—can utilize ionic and molecular signaling to achieve even greater integration and efficacy. Organic electronics, and in particular, so-called ‘iontronics’, have been proposed as an ideal platform for translating electronic signals into ionic signals for such bioelectronic technologies [1–4]. In the context of this article, iontronic devices are organic electronic components and systems that utilizes the coupling of the ionic and electronic transport in conducting polymers and hydrated polyelectrolytes. The prototypical iontronic component is the organic electronic ion pump (OEIP) (figure 1), which uses conducting polymer electrodes to drive electrophoresis (i.e. drug delivery) of charged substances for delivery at high spatiotemporal resolution and without liquid flow [5, 6]. OEIPs have been demonstrated in a wide array of biological applications including cell stimulation in vitro [5, 7], acute [6] and therapeutic applications [8–10] in vivo, and even for regulating plant physiology [11]. Additionally, the OEIP, which represents an iontronic resistor, has been expanded to iontronic diodes [12] (AEM-CEM junctions) and transistors [13, 14] (e.g. AEM-CES-AEM junctions), promising more precisely control of delivery and eventually more functionally complex bioelectronic systems and therapies.

To date, all of the reported OEIP and related iontronic technologies have been microfabricated using spin coating [12, 15–17] and photolithographic patterning technique [1, 5–7, 9, 12, 13, 18–20]. These high-precision technologies enable architectures with highly-resolved features and micron-scale delivery points. While such precision is necessary for eventual chronic applications in the brain or spinal cord where production cost is less important and performance is paramount, there remain many more application areas where higher-throughput, lower-cost, and even...
disposability are favored attributes. For example, acute or medium-term wound-care could benefit from precision delivery of anti-inflammatory substances, and peripheral nerve treatments could benefit from larger-area short-term implantable drug delivery 'patches'.

To address this different perspective of production and optimization, we turn to fabrication via printing techniques.

One of the hallmarks and primary justifications for OEIPs and similar iontronic technologies is the selective transport of either cations or anions, resulting in the equivalence between operating current and delivery rate. Selective ion delivery is accomplished by use of either anion- or cation-exchange membranes (AEM, CEM) as the electrophoretic transport channel (figures 1(a) and (b)). For example, if the ions intended for transport are anions, an AEM is used, and the source electrode is negatively biased relative to the target electrode (figure 1(a)). Ionic conductivity through the AEM (or CEM) is affected by the density of fixed charges and effective pore size between adjacent polymer chains [4, 17]. The high density of fixed charges is important to achieve Donnan exclusion, i.e. selectivity for only anions (or cations) [21]. Materials for developing OEIPs and other iontronics are thus selected based on these parameters.

Over a decade of research iontronics has mapped out and explored a range of devices, circuits, systems, and fabrication strategies [4, 9, 10]. However, most of these demonstrations utilized photolithographic processes, and those that did not [8, 22] involved significant manual fabrication of some kind. To date, printing technologies have only been used to add additional electrode or AEM material to devices [22] or where inkjet printing was used to deposit the electrolyte [5, 19]. As an first step towards reaching the goal of fully printed ionic circuits and devices, here we demonstrate all-printed polymer-based flexible OEIPs as a large-area manufacturing alternative to the typical wafer size lab-scale microfabricated devices. The investigations explores sequential layer printing and additional encapsulation steps, characterization of ion selectivity and device performance, and demonstration of delivery of biologically relevant ions: the

---

**Figure 1.** Illustration of the working principle of organic electronic ion pumps (OEIPs). A potential applied between electrodes creates an electrochemical circuit where either cations or anions in a source electrolyte are selectively transported through an ion exchange membrane to a target electrolyte. Investigated OEIPs had (a) anion exchange membranes (AEM) of pDADMAC\(^+\) or (b) cation exchange membranes (CEM) comprised of PSS\(^-\). (c) Schematic of fabricated screen-printed OEIPs (top view) displaying the source (S) and target (T) electrolytes and a zoomed area and specified dimensions of the ion exchange membrane. (d) Screen designs were made for separate layers to be printed. The screen printing process flow for the devices are shown from steps 1 to 9 starting from the pre-treated PET substrates, silver and carbon contacts, ion exchange membrane (pDADMAC\(^+\) or PSS\(^-\)), PEDOT: PSS electrodes, and dielectric. Electrolytes were applied for further testing or enclosed with adhesive layers and a PET lid.
neurotransmitter acetylcholine \((\text{ACH}^+)\) and the anti-inflammatory compound salicylic acid \((\text{SA}^-)\).

2. Methods

2.1. Device manufacturing

Screen-printing was performed using both semi- and fully-automated (DEK Horizon 03iX and EKRA Type E2) screen-printing equipment (supplementary info, table S1 is available online at stacks.iop.org/FPE/4/022001/jmmedia). Polyelectrolyte terephthalate foils (PET; 150 \(\mu\)m thick, Polifoil Bias, Policom Screen) were used as substrates and pre-conditioned at 130 °C for 1 h to avoid shrinking or other deformations during the printing. The screen-printing steps are illustrated in figure 1(d) and used to obtain A3-sized sheets each containing 54 OEIPs. Screen-printable silver (Ag 5000, DuPont) and carbon inks (7102, DuPont) were patterned into electrical conductors and contact pads to facilitate connection, e.g. using zero insertion force connectors. The carbon is introduced between the silver and PEDOT:PSS to act as a buffer layer reducing the possibility of electrochemically-released Ag ions penetrating into the PEDOT: PSS system [23]. The ion exchange membranes layer was printed using either a polyanionic or polycationic ink. Anion exchange membranes (AEMs) were printed with a composition based on polydiallyldimethylammonium chloride-based (pDADMAC) supplied by RISE Acreo. Cation exchange membranes (CEM) were printed using electronically and ionically conducting material poly(3,4-ethylenedioxythiophene) doped with poly(styrenesulfonate) (PEDOT:PSS; Clevios SV3, Heraeus). After screen printing PEDOT:PSS channels—which are both electronically and ionically conducting—the channels were overoxidized [6, 24] using freshly prepared sodium hypochlorite solution \((1 \text{v/v} \% \text{aq.})\) for 30 s to disable the electronic conducting pathways in the PEDOT phase while leaving the PSS phase (which is a CEM) intact. The over oxidation step is performed precisely by manually dropping sodium hypochlorite solution with a 100 \(\mu\)l multichannel micropipette on the screen printed PEDOT: PSS. Prior to further printing the sheets were rinsed with DI water and dried at 80 °C for 5 min. Following the membrane deposition two sequential layers of PEDOT:PSS were printed on top as electrodes. The silver, carbon, PEDOT:PSS, and pDADMAC patterned layers were each dried at 120 °C for 5 min. The printing was followed by an ion exchange step, prior to depositing the electrical insulator and encapsulant, to remove excess of unspecified ions present in the various inks: sheets were immersed in 1 M KCl(aq) for 5–10 min followed by washing with DI water and drying at 80 °C for 5 min. Two dielectric layers (3018 A, Dupont), which functioned both as an electrical insulator and source and target reservoir outlines (hydrophobic ‘enclosures’ to confine the source and target electrolytes, see figure 1(c)), were printed and cured by UV light exposure. Encapsulation of the source electrolyte in its reservoir was demonstrated by printing multiple layers of a UV-cured adhesive (3 M, UV-Curing Adhesive 7555) to create a deeper well, then gelling the source electrolyte in the well before covering it with a PET lid. Gelling was achieved by adding poly(ethylene glycol) diacrylate (PEGDA; 100 mg ml\(^{-1}\) electrolyte, M\(_{\text{w}}\) 700, Sigma-Aldrich) and a photoinitiator (lithium phenyl-2,4,6-trimethylbenzoyl-phosphinate, 1 mg ml\(^{-1}\) electrolyte, TCI Chemicals) to the source 100 mM KCl(aq) or HCl(aq) electrolyte, and UV curing the blend in a belt dryer at a speed of 1 m min\(^{-1}\) (AKTIPRINT UV table Dryer T28–1, lamp power 120 W cm\(^{-2}\), spectrum 200–400 nm). Electrical characterizations were carried out on devices without the adhesive encapsulation and gelled source electrolyte, to allow facile switching between different source and target electrolytes. Additional Ag/AgCl paste was manually painted onto a small portion of the exposed PEDOT:PSS source electrode. This was done to reduce the voltage drop during ion pump operation. The redox active material, Ag/AgCl, is in contact with the electrode connectors (carbon/Ag) through the PEDOT:PSS electrode and the electric current that reaches the Ag/AgCl interface through the PEDOT:PSS is converted to an ionic current by the redox reaction at the Ag/AgCl. This ionic current then goes into the ion exchange channel [25]. Manufactured devices were cut out either manually or with an automated laser engraver (Speedy 300 flex, Trotec). A profilometer (Dektak3ST, Veeco) was used to analyze the surface roughness and layer profile of the printed patterns. A Sagitta Optical reflection microscope was used in between screen-printing of different layers for the detection of defects and continuous quality and alignment checks.

2.2. Electrical characterization

All devices were immersed in 10 mM KCl(aq) for at least 12 h prior to testing. Electrical characterizations were carried out using a Keithley 2602 A and/or 2612 SourceMeter, controlled and monitored by a custom-made LabVIEW program. The source electrode was either positively or negatively biased depending upon the nature of ions to be transported (cations or anions). All experiments were performed after operating the devices at 1 V until reaching a stable current (\(\sim\)15–30 min). Source electrolytes (i.e. containing the ion intended for transport) and target electrolytes were varied as specified for each delivery experiment while applying a constant voltage of 1.3 V. In brief, small ion delivery was characterized in three ways: (i) varying the source electrolyte with constant source concentration \((100 \mu l;\text{ 10 mM; KCl, NaCl, LiCl, LiClO}_4, \text{KClO}_4 \text{or KOAc; aq})\) and a constant target electrolyte \((100 \mu l;\text{ 10 mM; KCl(aq)})\); (ii) the reverse of \(i\): constant source...
electrolyte and varying target electrolytes (constant concentration); and (iii) varying source electrolyte concentrations (10, 100, or 1000 mM KCl(aq)) and constant target electrolyte (10 mM KCl(aq)). Resistance was calculated by dividing the voltage by the current averaged over 200 s after reaching a stable value (i.e. the 200 s after ~15–30 min). Resistance was converted to conductivity and normalized to the conductivity of the 10 mM KCl(aq) run. Each data point represents the average results from at least 4 different devices.

2.3. Measurement of ACh⁺ and SA⁻ delivery

Larger biologically relevant ions were transported using cationic OEIPs (PSS-based) to deliver acetylcholine (10 mM AChCl(aq), Sigma) and anionic OEIPs (pDADMAC-based) to deliver salicylic acid (10 mM SA(aq), Sigma; pH adjusted to ~7 by adding KOH, Sigma). A constant positive or negative bias of 1.3 V was applied to the source electrode, respectively, and aliquots of the target electrolyte were sampled after set amounts of charge had run through the circuit. SA⁻ was delivered to 10 mM KCl(aq) target solution from which 5 μl aliquots were collected at specific intervals and replaced with 5 μl of fresh 10 mM KCl(aq). The amount of SA was quantified from the collected 5 μl sample aliquots using an ELISA kit (Neogen) following the supplier’s protocol and measured by a Synergy H1 microplate reader (BioTek). All the measurements were done for 3 different devices and performed in duplicate. ACh⁺ was delivered to 10 mM KCl(aq) target solution, from which 100 μl samples were collected after pumping specific charge/amount of ACh⁺ by rinsing the reservoir with 10 mM KCl. ACh in the 100 μl sample aliquot was quantified using an enzymatic choline biosensor (Model 7001, Pinnacle). Measurements were performed on four different devices sampled two times each. Concentration of the ACh⁺ transported was alternatively quantified with Synergy H1 Reader (BioTek) using the Amplex Red assay kit (Molecular Probes). 10 mM AChCl (aq) and 10 mM KCl (aq) was kept at the source and target respectively. The devices were operated for 2000–3000 s and 200 μl samples were collected after transporting required amount of ACh⁺. Measurements were done on 4 different devices and performed in duplicate.

3. Results and discussion

OEIPs were fabricated on flexible PET substrates using only sequential screen-printing of contact pads, ion selective membranes, organic electrodes, and dielectric materials (figure 1(c)). The ion exchange membrane layer in specific devices was selective for the delivery of either cations or anions. The electrodes need to be capable of converting electronic current to ionic current and the device function thereby relies strongly on the electrodes’ redox capacity, the measure of the materials’ available sites for oxidation at anodic potentials or reduction at cathodic potentials. Our design incorporated the most commonly used ion pump electrode material, and one of the most studied organic electronic materials: the electrically conducting polymer poly(3,4-ethylene-dioxythiophene) doped with the polyanion poly(styrene sulfonate) (PEDOT:PSS) [26, 27]. Electrode redox capacity, and thus device operational lifetime, was increased by printing multiple layers of PEDOT:PSS to increase its thickness (1 μm) and overall volume. Unlike previous OEIP demonstrations, the screen-printing process allowed us to easily add multiple layers of UV-curable dielectric to build up reservoirs (marked S in figure 1(c)) for the source and target electrolytes. The high-throughput processes allowed manufacture of 54 OEIPs per A3-sized PET sheet (figure 2). The thicker dielectric-based reservoirs enabled encapsulation and increased the robustness of the overall design. In addition, the source electrolyte containing the ionic species to be delivered, typically 100 mM KCl or HCl (aq), was gelled (using PEGDA and photoiniator) and enclosed under a PET, further enhancing robustness. Devices were patterned on the A3-sized PET sheet with the reservoirs spaced 1.8 mm apart, corresponding to the spacing between the tips of a standard multi-channel pipette.

The most important single component of OEIPs is the ion exchange membrane (IEM), of which the chemical, material and geometrical properties govern the efficiency and selectivity of ion transport. Here we demonstrate two different screen-printable IEMs: pDADMAC as AEM (figure 1(a)) and overoxidized PEDOT:PSS as CEM (figure 1(b)). Overoxidation of PEDOT:PSS chemically disables electronic conductivity in the PEDOT phase while keeping the polyanion PSS phase intact [5, 24]. This form of CEM has previously been demonstrated for iontronic devices and circuits fabricated by lab scale photolithographic techniques, but never in screen-printed iontronics. pDADMAC has been demonstrated in microfluidics-based iontronic circuits [28], but has never before been studied in an OEIPs or associated devices. Here we demonstrate screen-printed pDADMAC AEMs for the first time in OEIPs, i.e. solid/hydrogel-state iontronics.

After integrating the IEMs into the screen-printed design, the selectivity for cations or anions of each membrane was evaluated. For a highly selective AEM, the majority of the current through the membrane will be due to anion transport, while for a CEM the current will primarily be due to the transport of cations [29]. As the conductivity is dependent on the ions that are transported (i.e. ions in the source electrolyte), for an efficient ion exchange membrane this conductivity should be in direct relation with the ion diffusion coefficient [16]. In accordance with Graham’s law, and assuming constant temperature of all experiments, the ionic size is inversely related to the diffusion...
Figure 2. Screen-printed OEIPs on an A3-sized PET substrate and a zoomed-in image shows an individual ion pump in action delivering $\text{H}^+$ to the target electrolyte containing a pH indicator (delivery indicated by red dot).

Figure 3. Diffusion coefficient versus normalized conductivity for different source and target electrolytes (a) device response for pDADMAC$^+$ based printed OEIPs with varied source anions ($\text{Cl}^-$, $\text{OAc}^-$ and $\text{ClO}_4^-$) and fixed counter ion ($\text{K}^+$) and target electrolyte ($\text{KCl}$). (b) pDADMAC$^+$ OEIP response with fixed source electrolyte ($\text{KCl}$) and counterion ($\text{Cl}^-$) but varied target cations. (c) Device response for PSS$^-$ based printed OEIPs by varying source cations ($\text{K}^+$, $\text{Na}^+$ and $\text{Li}^+$) and maintaining counter ion ($\text{Cl}^-$) and target electrolyte ($\text{KCl}$). (d) PSS$^-$ OEIP response for fixed source electrolyte and counterion ($\text{K}^+$) and varying target anions.
coefficient; larger ions have lower diffusion coefficient. Thus, larger ions travel more slowly than smaller ions and hence decrease the conductivity. Ions in the target solution should ideally be excluded by the IEM and therefore not influence the conductivity. To this aim, sets of experiments were carried out to measure the conductivity for various source and target electrolytes comprising ions of varying sizes/diffusion coefficients [30]. The source anions used for the devices with pDADMAC AEMs were Cl\(^-\), ClO\(_4^-\), and OAc\(^-\) (acetate), with K\(^+\) as counter ion. Similarly, devices with PSS CEMs were tested with different source cations, including Li\(^+\), Na\(^+\), and K\(^+\), with Cl\(^-\) as counter ion. In all cases the target electrolyte was KCl. For pDADMAC AEMs, the resulting conductivity decreased with decreased diffusion coefficients of the anions in the order Cl\(^-\) > ClO\(_4^-\) > OAc\(^-\) (figure 3(a)). Results obtained with PSS CEMs showed that the conductivity of the cations transported scaled with the diffusion coefficients and decreased in the order of K\(^+\) > Na\(^+\) > Li\(^+\) (figure 3(c)). For comparison, the experiments were inverted: source electrolyte was consistently KCl and the target cations and anions were varied. pDADMAC AEMs showed no significant dependence on the diffusion coefficient of the target cation (figure 3(b)). Similar observations for PSS CEMs showed only minimal dependance on the properties of the target anions (figure 3(d)). The combined results indicate that anions were indeed preferentially transported with the printed pDADMAC devices, whereas cations were preferentially transported through PSS-based devices. The screen-printed OEIPs’ IEMs were further investigated by varying the source concentration of KCl. As charge transport depends on the number of fixed charges in the IEM, for a selective membrane the transport should be independent of the source or target electrolyte concentration [21]. Devices were operated at 1 V until a predetermined amount of charge (time-integrated current) was transported. The charge delivered was measured and compared for different ion pumps. Both types of OEIPs showed no dependency on the source concentration (supplementary figure S1), further indicating functioning IEMs in the screen-printed devices.

To investigate the possibility of delivering larger and biologically relevant ions we chose acetylcholine (ACh\(^+\)) as a model cation and salicylic acid (SA\(^-\)) as a model anion. Acetylcholine is a neurotransmitter which is responsible for proper transmission between motor neurons and muscle cells via brain synapses [25, 31]. By precisely delivering ACh to the location could help in management of diseases like Alzheimers which is linked to degeneration of Acetylcholine producing cells [32]. Salicylic acid is the basic component of acetylsalicylic acid (aspirin), one of the primary non-steroidal anti-inflammatory drugs (NSAIDs) in use [33]. It is also keratolytic, causing epidermal shedding, and is commonly used for the treatment of acne, dandruff, seborrhea, psoriasis, and other skin disorders [34, 35]. Corresponding PSS and pDADMAC devices were operated at 1.3 V to deliver ACh\(^+\) and SA\(^-\) to separate target electrolytes containing 100 \(\mu\)l KCl. Samples were collected from the target at various time points, corresponding to set amounts of charge transported through the IEM, and chemically quantified using assay kits and choline biosensors as explained in the previous section. Quantified amounts were converted into units of charge, as \(q = nF\), where \(n\) is number of moles and \(F\) is Faraday’s constant. Detected quantities were compared to the total charge, yielding the OEIP efficiency: the ratio of intended ions transported (i.e. ACh\(^+\) or SA\(^-\)) in units of charge to the measured charge (time-integrated current) at the corresponding time point. The direct characterization of ACh\(^+\) in the collected samples confirmed that the PSS pumps could indeed be used to deliver neurotransmitters. The amount of ACh\(^+\) increased linearly with time/charge, allowing controlled delivery of ions (figure 4(a)). The average efficiency of acetylcholine delivery using the PSS OEIPs was approximately 45%. This efficiency is lower than that reported for previous PEDOT:PSS-based micro-patterned OEIPs [7]. This is likely due to the different fabrication methods and material formulations used herein; no ion transport

- **Figure 4.** Direct characterization of (a) acetylcholine (ACh\(^+\)) transport using screen-printed PSS\(^-\)-based OEIPs, and (b) salicylic acid (SA\(^-\)) transport using screen-printed pDADMAC\(^-\)-based OEIPs.
studies have been reported for screen-printed and overoxidized films of this type. In addition, the choline biosensor used may have experienced drift or time-related effects (there was some indication of decreasing efficiency when operating the devices for longer times). To verify ACh⁺ transport, we used a secondary method of quantification (immunofluorescence assay) which gave similar results (figure S2). pDADMAC based OEIPs could in a similar fashion deliver SA⁻ (figure 3(b)). The rate of SA⁻ delivery appeared to increase for longer delivery times. However, the average efficiency of salicylic acid delivery only appeared to be around 20%. Unfortunately, there are no reported studies on SA⁻ (or other such aromatic substance) delivery using pDADMAC for further comparison.

The relatively low efficiencies of both PSS and pDADMAC OEIPs indicates that while controlled ACh⁺ and SA⁻ delivery was possible, much of the ionic current was due to transport of other species. Presence of additional ions can for example depend on the choice of materials or inks for device fabrication. To minimize the potential contribution from unspecified ions present in the printed inks, we performed an additional ion exchange step with concentrated KCl (1 M) before encapsulating the devices with the dielectric material. Each ionic membrane was also preloaded with either ACh⁺ and SA⁻ prior to the actual experiment by pumping for ~30 min or until the current stabilized; i.e. until ions present in the channel from the fabrication stages are fully exchanges with the ionic species intended for delivery. Another possible explanation for lowered efficiencies is water splitting, which is known to occur for ion pumps of certain geometries, membrane types, or at high operating voltages. The electric field at the electrolyte-IEM junction could enhance water dissociation, forming ions [36]. Hence these devices should only be operated under low voltages [12]. The diffusion coefficient of H⁺ and OH⁻ (9.3 × 10⁻⁵ and 5.3 × 10⁻⁵ cm² s⁻¹, respectively) are higher compared to other ions. Ions created from water splitting are therefore more easily transported than larger ions such as ACh⁺ and SA⁻ and could account for the decreased efficiency. To investigate if water splitting might be contributing to the low efficiencies, H⁺ and K⁺ were transported at different voltages ranging from 0.7 to 3 V through a PSS−IEM to a target containing a pH indicator. The change in the color at the tip of the IEM indicates transport of H⁺ to the target (figure 5(a)). As expected the pH change was dominant while intentionally transporting H⁺. However, a pH change was also observed while transporting K⁺, albeit to a lesser extent compared to H⁺, as shown by the extracted intensity profiles (figure 5(b)). Similar experiments were carried out with the source-target voltage <1 V and no change in pH was observed. This indicates that under higher voltage there could be unknown species within the printed membranes, such as H⁺, OH⁻, or other.

Figure 5. Ion pump in action. Optical micrographs of transporting (a) H⁺ and (b) K⁺ using PSS−OEIPs into target electrolytes containing pH indicator operated at 3 V and (c) intensity profiles.
substances resulting from exposure to large changes in pH [19]. Under careful investigation with pH indicators, the safe operating region for our ion pumps was found to be less than 1.2 V.

4. Conclusion

We demonstrate fully-encapsulated flexible OEIPs fabricated via a screen-printing process conducive to large-scale manufacturing. This design demonstrates a system with enclosure and gelling to mitigate failure by leakage, as well as a geometry allowed fast filling of electrolytes for rapid testing. The development and fabrication process is easily tunable as demonstrated by screen-printing of two different ion exchange membrane formulations and integration into the same design. These screen-printed OEIPs can thus be used to deliver either cations and anions. The cation-exchange membrane was based on the familiar material PSS, whereas the anion-exchange membrane was based on pDADMAC in one of its complementary materials could thus be used to print a variety of more complex iontronic devices and circuits [4].

Screen-printed OEIPs were demonstrated to transport small ions (Na⁺, K⁺, Cl⁻, H⁺, etc.) as well as ‘larger’, more biologically relevant ions: the cationic neurotransmitter acetylcholine and the anionic anti-inflammatory salicylic acid. Even this small repertoire of substances highlights potential applications in neuroscience, inflammatory disease and epidermal treatment of chronic wounds. Combined with the concept of low-cost and high-throughput printing techniques, salicylic acid transport in particular points toward applications in smart skin-care patches.

In addition to demonstrating potential future uses, we also addressed the limitations of these screen-printed OEIPs in terms of operating voltage and design. With additional optimization and modification on the devices and structures, we believe these and future printed OEIPs can be expanded for complex and faster-delivery ion pumps for use in a variety of biologically relevant applications.

Acknowledgments

Major funding for this work was provided by the Swedish Foundation for Strategic Research. Additional funding was provided by Vinnova and the Knut and Alice Wallenberg Foundation and the Swedish Government Strategic Research Area in Materials Science on Advanced Functional Materials at Linköping University (Faculty Grant SFO-Mat-LiU No. 2009-00971). Additional assistance with mask design and printing was provided by Marie Nilsson at RISE Acreo, Norrköping, Sweden.

ORCID iDs

Klas Tybrandt https://orcid.org/0000-0002-9845-446X
Daniel T Simon https://orcid.org/0000-0002-2799-3490

References

[1] Simon D T, Gabrielsson E O, Tybrandt K and Berggren M 2016 Organic bioelectronics: bridging the signaling gap between biology and technology Chem. Rev. 116 13009–41
[2] Owens R M and Malliaras G G 2010 Organic electronics at the interface with biology MRS Bull. 35 449–56
[3] Tarabell G, Mohammedi F M, Coppede N, Barbero F, Iannotta S, Santato C and Cicoiro F 2013 New opportunities for organic electronics and bioelectronics: ions in action Chem. Sci. 4 1395–409
[4] Arbring Sjöström T, Berggren M, Gabrielsson E O, Janson P, Poxxon D J, Seitanidou M and Simon D T 2018 A decade of iontronic delivery devices Adv. Mater. Technol. 3 1700360
[5] Isaksson J, Kiöll P, Nilsson D, Robinson N, Berggren M and Richter-Dahlfors A 2007 Electronic control of Ca²⁺ signaling in neuronal cells using an organic electronic ion pump Nat. Mater. 6 673–9
[6] Simon D T, Kurup S, Larsson C K, Horii R, Tybrandt K, Goiny M, Jager E W H, Berggren M, Carlon B and Richter-Dahlfors A 2009 Organic electronics for precise delivery of neurotransmitters to modulate mammalian sensory function Nat. Mater. 8 742–6
[7] Tybrandt K, Larsson K C, Kurup S, Simon D T, Kiöll P, Isaksson J, Sandberg M, Jager E W H, Richter-Dahlfors A and Berggren M 2009 Translating electronic currents to precise acetylcholine-induced neuronal signaling using an organic electrophotogetic delivery device Adv. Mater. 21 4442–6
[8] Jonsson A, Song Z, Nilsson D, Meyerson B A, Simon D T, Linderoth B and Berggren M 2015 Therapy using implanted organic bioelectronics Sci. Adv. 1 e1500039
[9] Uguz I, Proctor C M, Curto V F, Pappa A M, Donahue M J, Perro M, Owens R M, Khodagholy D, Inal S and Malliaras G G 2017 A microfluidic ion pump for in vivo drug delivery Adv. Mater. 29 1701217
[10] Proctor C M, Slezia A, Kasya A, Gheestam A, del Agua I, Pappa A M, Bernard C, Williamson A J and Malliaras G G 2018 Electrophotogetic drug delivery for seizure control Sci. Adv. 4 eaaas1291
[11] Poxxon D J et al 2017 Regulating plant physiology with organic electronics Proc. Natl Acad. Sci. 114 4597–602
[12] Gabrielsson E O, Tybrandt K and Berggren M 2012 Ion diode logics for pH control Lab Chip 12 2507–13
[13] Tybrandt K, Larsson K C, Richter-Dahlfors A and Berggren M 2010 Ion bipolar junction transistors Proc. Natl Acad. Sci. 107 9929–32
[14] Tybrandt K, Gabrielsson E O and Berggren M 2011 Toward complementary ionic circuits: the npn ion bipolar junction transistor J. Am. Chem. Soc. 133 10141–5
[15] Gabrielsson E O, Janson P, Tybrandt K, Simon D T and Berggren M 2014 A four-diode full-wave ionic current rectifier based on bipolar membranes: overcoming the limit of electrode capacity Adv. Mater. 26 5143–7
[16] Arbring Sjöström T, Jonsson A, Gabrielsson E O, Karge L, Tybrandt K, Berggren M, Simon D T and Strö S 2017 Cross-linked polyelectrolyte for improved selectivity and processability of iontronic systems ACS Appl. Mater. Interfaces 9 30247–52
[17] Tybrandt K 2017 Exploring the potential of ionic bipolar diodes for chemical neural interfaces Soft Matter 13 8171–7
[18] Daniel T, Jager E W H, Tybrandt K and Larsson K C 2009 An organic electron ion pump to regulate intracellular signalling at high spatiotemporal resolution Int. Conf. Solid-State Sensors, Actuators and Microsystems 2009 1790–3
[19] Gabrielsson E O and Berggren M 2013 Polyphosphonium-based bipolar membranes for rectification of ionic currents Biomicrofluidics 7 64117
[20] Gabrielsson E O, Tybrandt K and Berggren M 2014 Polyphosphonium-based ion bipolar junction transistors Biomicrofluidics 8 64116
[21] Kontturi K, Murtomäki L and Manzanares J A 2008 Ionic Transport Processes (Oxford: Oxford University Press)
[22] Simon D T, Larsson K C, Nilsson D, Burström G, Galter D, Berggren M and Richter-Dahlfors A 2015 An organic electronic biomimetic neuron enables auto-regulated neuromodulation Bioens. Bioelectron. 71 359–64
[23] Berto M, Diacci C, Theuer L, Di Lauro M, Simon D T, Berggren M, Biscarini F, Beni V and Bortolotti C A 2018 Label free urea biosensor based on organic electrochemical transistors Flex. Print. Electron. 3 24001
[24] Krische B and Ziegler S 2018 Overoxidation in conducting polymers Synth. Met. 28 257–62
[25] Lodish H, Berk H and Zipursky S 2000 Neurotransmitters, synapses, and impulse transmission Molecular Cell Biology (San Francisco, CA: Freeman)
[26] Heywang G and Jonas F 1992 Poly(alkylenedioxythiophene) s—new, very stable conducting polymers Adv. Mater. 4 116–8
[27] Groenendaal L, Bert, Jonas F, Freitag D, Pielartzik H and Reynolds J R 2000 Poly(3,4-ethylenedioxythiophene) and its derivatives: past, present, and future Adv. Mater. 12 481–94
[28] Han J-H, Kim K B, Kim H C and Chung T D 2009 Ionic circuits based on polyelectrolyte diodes on a microchip Angew. Chem. 48 3830–3
[29] Miyoshi H 1999 Donnan dialysis with ion-exchange membranes: III. Diffusion coefficients using ions of different valence Sep. Sci. Technol. 34 231–41
[30] Vaněšek P 1996 Ionic conductivity and diffusion at infinite dilution CRC Handbook of Chemistry and Physics (Boca Raton, FL: CRC Press) pp 77–9
[31] Koelle G B 1972 Acetylcholine—current status in physiology, pharmacology and medicine N. Engl. J. Med. 286 1086–90
[32] Francis P T, Palmer A M, Snape M and Wilcock G K 1999 The cholinergic hypothesis of Alzheimer’s disease: a review of progress J. Neurol. Neurosurg. Psychiatry 66 137–47
[33] Amann R and Peskar B A 2002 Anti-inflammatory effects of aspirin and sodium salicylate Eur. J. Pharmacol. 447 1–9
[34] Kootiratkar K, Kampirapap K and Chunhaswwa C 2015 Epidermal permeability barrier in the treatment of keratosis pilaris Dermatol. Res. Pract. 2015 205012
[35] Arif T 2015 Salicylic acid as a peeling agent: a comprehensive review Clin. Cosmet. Investig. Dermatol. 8 455–61
[36] Strathmann H, Krol J, Rapp H-J and Eigenberger G 1997 Limiting current density and water dissociation in bipolar membranes J. Memb. Sci. 125 123–42