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Conducting polymer coated neural recording electrodes

Abstract

Neural recording electrodes suffer from poor signal to noise ratio, charge density, biostability and biocompatibility. This paper investigates the ability of conducting polymer coated electrodes to record acute neural response in a systematic manner, allowing in depth comparison of electrochemical and electrophysiological response. Approach. Polypyrrole (Ppy) and poly-3,4-ethylenedioxythiophene (PEDOT) doped with sulphate (SO₄) or para-toluene sulfonate (pTS) were used to coat iridium neural recording electrodes. Detailed electrochemical and electrophysiological investigations were undertaken to compare the effect of these materials on acute in vivo recording. Main results. A range of charge density and impedance responses were seen with each respectively doped conducting polymer. All coatings produced greater charge density than uncoated electrodes, while PEDOT-pTS, PEDOT-SO₄ and Ppy-SO₄ possessed lower impedance values at 1 kHz than uncoated electrodes. Charge density increased with PEDOT-pTS thickness and impedance at 1 kHz was reduced with deposition times up to 45 s. Stable electrochemical response after acute implantation inferred biostability of PEDOT-pTS coated electrodes while other electrode materials had variable impedance and/or charge density after implantation indicative of a protein fouling layer forming on the electrode surface. Recording of neural response to white noise bursts after implantation of conducting polymer-coated electrodes into a rat model inferior colliculus showed a general decrease in background noise and increase in signal to noise ratio and spike count with reduced impedance at 1 kHz, regardless of the specific electrode coating, compared to uncoated electrodes. A 45 s PEDOT-pTS deposition time yielded the highest signal to noise ratio and spike count. Significance. A method for comparing recording electrode materials has been demonstrated with doped conducting polymers. PEDOT-pTS showed remarkable low fouling during acute implantation, inferring good biostability. Electrode impedance at 1 kHz was correlated with background noise and inversely correlated with signal to noise ratio and spike count, regardless of coating. These results collectively confirm a potential for improvement of neural electrode systems by coating with conducting polymers. 2013 IOP Publishing Ltd.

Keywords

polymer, conducting, coated, neural, recording, electrodes

Disciplines

Engineering | Physical Sciences and Mathematics

Publication Details

Harris, A. R., Morgan, S., Chen, J., Kapsa, R. M., Wallace, G. G. & Paolini, A. (2013). Conducting polymer coated neural recording electrodes. *Journal of Neural Engineering*, 10 (1), 1-16.

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Conducting polymer coated neural recording electrodes

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Abstract

Polypyrrole (Ppy) and 3,4-ethylenedioxythiophene (PEDOT) doped with sulphate (SO₄) or *para*-toluene sulfonate (pTS) were used to coat iridium neural recording electrodes. Detailed electrochemical and electrophysiological investigations were undertaken to compare the effect of these materials on acute *in vivo* recording. A range of charge density and impedance responses were seen with each respectively doped conducting polymer. All coatings produced greater charge density than uncoated electrodes, while PEDOT-pTS, PEDOT-SO₄ and Ppy-SO₄ possessed lower impedance values at 1 kHz than uncoated electrodes. Charge density increased with PEDOT-pTS thickness and impedance at 1 kHz was reduced with deposition times up to 45 s. Stable electrochemical response after acute implantation denoted biostability of PEDOT-pTS coated electrodes while other electrode materials had variable impedance and/or charge density after implantation indicative of a protein fouling layer forming on the electrode surface. Recording of neural response to white noise bursts after implantation of conducting polymer-coated electrodes into a rat model inferior colliculus showed a general decrease in background noise and increase in signal to noise ratio and spike count with reduced impedance at 1 kHz, regardless of the specific electrode coating, compared to uncoated electrodes. A 45 s PEDOT-pTS deposition time yielded the highest signal to noise ratio and spike count. These results collectively confirm a potential for improvement of neural electrode systems by coating with conducting polymers.

Introduction

Electrophysiology is a technique that measures the electrical properties of cells and cellular networks as a way of understanding cell function and pathways. It has been applied to single and multi-cellular studies in cell or tissue culture and in live animals. The technique can be used to measure the current and/or potential in or around the cells of interest or to stimulate cells by the injection of current. It is a popular technique in the examination and control of neural function, and can be applied in acute or chronic behavioural experiments.

The placement of an electrode into the brain of an animal for local neural stimulation or recording experiments requires that the electrode dimensions conform to dimensions (1-100 μm) compatible with the target neurons or neural structures. One electrode used extensively is the glass pipette. The pipette is drawn to a small diameter and then filled with an electrolyte solution and typically a Ag/AgCl wire to maintain a stable electrode potential. The pipette tip is then placed next to or inside the target neuron for stimulation or recording. This type of electrode is very cheap and easy to make, and the controlled, stable electrode/neural interface potential facilitates reliable measurement of neural potential. However, these pipettes are fragile, become blocked by tissue and have a low bandwidth, which limits their use for high frequency measurements and current injection pulses [1]. Therefore, glass pipettes are typically only used for single cell stimulation and recording experiments.

Metal microelectrodes composed of gold, steel, platinum, iridium and tungsten amongst others, are also routinely used [2] and have been shown to be more robust and versatile in terms of shape than glass pipettes; typically in the form of a disc surrounded by an insulating material (eg. silicon oxide). By combining standard electronic manufacturing protocols to these electrodes, 2D and 3D arrays of microelectrodes have now become commercially available [3]. The bandwidth of these electrodes is much wider than glass pipettes, allowing more accurate measurement of action potentials and more rapid current injection pulses. However, the electrode/neural interface potential is not controlled, which results in potential drift necessitating a high pass filter, thus preventing their use in DC potential measurements [1]. The charge capacity of the electrodes is related to their size, and miniaturisation for single cell experiments limits the charge able to be applied before damaging either the electrode or the surrounding tissue [4]. Furthermore, fouling of the electrode surface can occur when implanted into tissue, altering the electrode characteristics including impedance and charge capacity. Currently available electrodes have variable performance levels with a less than desirable signal to noise ratio and poor chronic implantation stability [5, 6].

More recently, a range of new electrode materials and coatings have been proposed: Iridium oxide is a material able to undergo Faradaic electrochemical reactions, resulting in a much larger charge capacity than pure planar or roughened metal electrodes [7] but its long term stability issues, particularly at high current injection levels, limit its potential for chronic implantation [8-11].

Capacitor type electrodes such as tantalum oxide allow charge to be injected without a Faradaic component, resulting in less electrode damage and more stable chronic performance [12]. These types of electrodes are more difficult to manufacture and typically have a lower charge capacity than other electrode materials. Carbon-based materials, including carbon fibres [13], carbon nanotubes [14, 15] and graphene [16] are comparatively economical and present a very high surface area, resulting in a large charge capacity, however long term *in vivo* stability of these materials has not yet been tested and the safety of carbon nanotubes is still being debated [17]. Doped diamond has also recently been reported as a very low fouling, good biocompatibility and large charge capacity material [18]. However, these electrodes are difficult and relatively expensive to manufacture, and are not commercially available.

Another class of electrode material to have been proposed are the organic conducting polymers (OCP). Typically polypyrrole (Ppy) or poly-3,4-ethylenedioxythiophene (PEDOT), doped with various anionic dopants such as poly(styrene sulfonate) (PSS) or perchlorate (ClO_4^-) have been deposited onto metal electrodes [19-21]. These materials allow both Faradaic and non-Faradaic charge transfer to occur at the electrode/neural interface and have a rough surface, resulting in a large electroactive area and charge capacity. The impedance of these materials at 1 kHz (typically around 10-100 kOhm) is also significantly less than uncoated electrodes (typically around 200-500 kOhm) [19, 22], yielding electrodes with superior signal to noise ratio compared to bare metal electrodes. More recent work on conducting polymer coatings for neural implants has incorporated short peptides and nerve growth factors to improve the chronic stability of the electrode and encourage growth of neurons towards the electrode, further improving the signal to noise ratio [23-25]. Growth of the conducting polymer on the electrode *in vivo* has even allowed the encapsulation of neurons for extremely large surface area electrodes [26].

Despite a significant body of work on conducting polymer-coated neural electrodes, a systematic comparison of differently doped conducting polymer coatings - incorporating polymer type (ie physico-chemical properties), thickness or dopant ion - has not been performed. This precludes rational comparison of polymer coating performance, thereby limiting the development of new electrode coatings to further improve the signal to noise ratio, charge capacity, biostability and biocompatibility. As a pertinent example of this shortfall, despite its significantly higher conductivity than previously used dopant ions [27] and potentially very good biocompatibility [28] in OCP systems such as Ppy and PEDOT, *para*-toluene sulfonate (pTS) has yet to be applied as a dopant in OCP-electrode systems *in vivo*. As a counterpoint to gain a more in depth understanding of the chemical functionality that leads to improved OCP-coated electrode performance *in vivo*, the small sulphate (SO_4^{2-}) anion with similar structure to pTS and PSS was chosen for this initial study. Comparison of Ppy and PEDOT doped with pTS and SO_4^{2-} dopants in a controlled manner thus stands to deliver

valuable insight into some of the important influential molecular factors at the electrode/neural interface.

Comparative studies of each respective conducting polymer, doped with pTS and SO_4^{2-} , were performed using 32-electrode multi-electrode array (MEA) probes. The probe contained independent electrodes coated with Ppy-pTS, Ppy- SO_4 , PEDOT-pTS and PEDOT- SO_4 respectively in addition to uncoated electrodes, and which were tested in one animal per probe to eradicate variations between animals, materials or techniques. A total of 4 animals with 128 electrode sites were used to compare the electrode materials. The electrodes were placed into the inferior colliculus of a rat animal model to allow precise and high repetition acoustic stimulation.

Methods

Materials and Electrode Coating

Pyrrole, 3,4-ethylenedioxythiophene (EDOT), sodium sulphate (Na_2SO_4) and sodium *para*-toluene sulfonate (Na_2pTS) (Sigma-Aldrich) and 99.0 % di-sodium phosphate (Fluka) were used as received. Polymer coatings were deposited on 4 shank, 32 electrode (8 electrodes per shank), $413 \mu\text{m}^2$ nominal geometric area iridium electrodes with $200 \mu\text{m}$ pitch (Neuronexus Technologies – A4x8-5mm-200-200-413). Optical microscopy was undertaken before and after polymer deposition to ensure uniform electrode coverage (figure 1), however no attempt was made to measure the electroactive area of each electrode site. While a more accurate correlation of electrode coating and performance could be obtained by correcting for real geometric or electroactive area, the trends seen between polymer and dopant types and with different deposition times when averaged across multiple electrodes and animals should minimise the error between the nominal and the real geometric or electroactive area used in these initial studies.

Conducting polymer coatings with different dopants were electrochemically deposited onto individual microelectrodes via a potentiostat/galvanostat system (Model 283, Princeton Applied Research) from mixed solutions containing 10 mM monomer (pyrrole (10 s deposition time) or EDOT (15 s deposition time)) and 0.1 M Na_2SO_4 or Na_2pTS in water. Potentiostatic growth was performed in a three-electrode configuration using one microelectrode as the working electrode, Ag/AgCl as reference electrode and Pt mesh as counter electrode. 4 electrode sites were coated with each doped conducting polymer in a staggered array, leaving 16 uncoated iridium electrodes as controls (figure 1). The best polymer from the first group of *in vivo* testing in 2 animals with 2 different probes was subsequently deposited at different thicknesses controlled by deposition time (15s, 30s, 45s and 60s) on 4 electrode sites each, in a staggered arrangement, again leaving 16 sites uncoated and tested in 2 animals with 2 different probes. Electrodes were imaged using a BX61 optical microscope (Olympus). Electrochemical analysis was undertaken in 0.3 M phosphate buffer. Test solutions were not

degassed to better represent conditions *in vivo*. A CHI660B potentiostat with CHI684 multiplexer (CH Instruments) were used to perform cyclic voltammetry and electrochemical impedance spectroscopy (EIS) at each of the individually addressable working electrode sites. A 3 electrode configuration was used with a Ag/AgCl (3 M KCl) reference and Pt mesh counter electrode. Cyclic voltammetry was performed over a range of 0.8 to -0.8 V vs Ag/AgCl at a scan rate of 100 mV s⁻¹. EIS was undertaken at 0 V with a 10 mV amplitude over a frequency range of 10-100,000 Hz as the power spectrum of multi-unit neural recordings extends from below 300 Hz to over 3 kHz [29], it is therefore useful to measure the impedance over this entire range. The high frequency region measured in EIS also provides details on the electrodes electrochemical kinetics. As different materials will display different kinetics, measurement of high frequency impedance allows investigation of the material performance.

In vivo Testing

Experimental procedures were performed in a sound attenuating Faraday room on an anti-vibration table (Technical Manufacturing Corporation). Hooded Wistar rats weighing over 200 g were anaesthetised with urethane (20% v/v in distilled water, 1.3 g/kg i.p., Sigma-Aldrich). The animal was placed in a stereotaxic frame (David Kopf Instruments) fitted with a hollow ear bar in the left ear. Animal temperature was monitored continuously via a rectal probe and maintained at 37.5 °C using an ATC1000 DC temperature controller (World Precision Instruments). A craniectomy was performed to access the right inferior colliculus (IC). A Ag/AgCl wire reference electrode wrapped in saline saturated cotton wool was placed into the dorsal region of the animals neck. The multichannel polymer electrode was then inserted at a 19° rostro-caudal angle with reference to Lambda using stereotaxic coordinates and a rat brain atlas [30] approximately 2 mm into the brain, towards the IC. White noise bursts were generated by a RX6 multifunction processor and PA5 programmable attenuator (Tucker-Davis Technologies) controlled by custom software developed in OpenEx. Sound was delivered through the left ear bar using an EC1 electrostatic speaker driven using an ED1 electrostatic speaker driver (Tucker-Davis Technologies). Prior to use, the speaker was calibrated by attachment of the sound generation system to one end of the ear bar with a one-eighth-inch 4138-A-015 microphone and amplifier unit and 2829 4-Channel Microphone Power Supply (Brüel and Kjær) coupled to the other end using a 3 mm long rigid plastic tube to mimic the rat's ear canal. The electrode was then advanced into the IC using a motorized microdrive (Sutter Instruments), whilst monitoring the neural response via a PZ2 high impedance amplifier and RZ2 bioamp processor (Tucker-Davis Technologies) with band-pass filtering (300–5000 Hz), until roughly the bottom 3 electrodes on each shank displayed acoustically driven activity.

An acoustic stimulation protocol of 300 repetitions of 50 ms white noise bursts (rise-fall time 10 ms, Gaussian distributed noise, 1–44 kHz) at a 1 s repetition rate were then delivered through the speaker

at 4 different amplitudes (40-70 dB in 10 dB steps), while recording the multiunit activity at each electrode (acquired at a sampling rate of 24.4 kHz). On completion of the acoustic stimulation protocol, the probe was advanced 200 μm into the IC so that each electrode was in approximately the same position as the more distal electrode from the first measurement. The acoustic stimulation was then repeated and the probe advanced in 200 μm steps until all of the electrodes had recorded acoustically evoked activity. The probe was then retracted in 200 μm steps using the same acoustic stimulation protocol to determine the reproducibility of the measurements and potential damage caused from the probe insertion. After *in vivo* recording, the electrodes were carefully retracted from the animal and gently rinsed with deionised water before testing by voltammetry and EIS. They were then placed in an enzymatic cleaning solution (Ultrazyme, Advanced Medical Optics) for 24 hours, rinsed in deionised water and voltammetry and EIS repeated. All experimental procedures were approved by the La Trobe University Animal Ethics Committee (09-28P).

Data Analysis

Acoustically evoked responses were imported into Matlab for offline analysis. For each electrode site, the average of the raw signal RMS measured during acoustic stimulation (RMS_{stim}) for the complete 50 ms stimulation period was averaged from the 300 repetitions at one electrode depth. The average of the raw signal RMS outside the acoustic stimulation period (RMS_{bkgd}) was only performed over the last 300 ms of the 950 ms time bracket averaged from the 300 repetitions at one electrode depth, to eliminate artifacts from neuron refractory periods. The signal to noise ratio (SNR) was calculated from $(\text{RMS}_{\text{stim}} / \text{RMS}_{\text{bkgd}})$ with the SNR classification taken from [31] (where low SNR < 3.5, medium SNR 3.5-4.0 and good SNR > 4.0). A spike was measured where the recorded potential was $>4.2 \times \text{S.D.}$ of the RMS from the previous 1 s with an exponential weighting of signal. The “during” and “outside” acoustic stimulation spike count was then performed over the same time periods as above. The maximum peak-to-peak amplitude outside the acoustic stimulation period was determined from the whole 950 ms period, where the high and low point was within 5 sample points (~ 2 ms). Each electrode site was considered “in” the IC when the SNR at one electrode depth with a 70 dB acoustic stimulation was greater than 25 % of the maximum recorded SNR over the whole experiment at 70 dB of acoustic stimulation. Data for each electrode site was averaged across all electrode depths “in” the IC to reduce error due to variations in the number of recordable neurons in the vicinity of the electrodes.

Results

Electrochemistry of differently doped conducting polymers on neural electrodes

Each of the doped conducting polymers was deposited onto iridium electrodes with a concomitant increase in current over time, indicating a growth of the electroactive area. Removal of the electrode from the deposition solutions showed the formation of uniform dark blue films of PEDOT-pTS, Ppy-pTS and Ppy-SO₄ that had spread over the electrode surface and extended over the silicon substrate (approximately 98%, 56% and 80% increase in geometric area respectively) while a more transparent blue PEDOT-SO₄ was confined to the electrode surface and the uncoated iridium electrodes remained bright silver (figure 1).

Cyclic voltammetry of each electrode was consistent with previously reported results (figure 2a) [19, 22, 27, 32]. Uncoated iridium electrodes possessed a small reductive current at approximately -0.7 V vs Ag/AgCl and occasionally the current crossed-over itself on the oxidative scan around the same potential region, most likely associated with oxygen reduction in the non-degassed solution. Otherwise, no features were visible over the potential window of 0.8 to -0.8 V and the electrode capacitance was low. PEDOT-SO₄ had a small sigmoidal shaped reductive process around -0.6 V with a slightly larger electrode capacitance. Ppy-SO₄ had a sharp reduction peak around -0.55 V and a broader oxidation peak at -0.33 V, the background capacitance again increasing slightly. Ppy-pTS had a reduction peak around -0.65 V with an oxidation peak at -0.25 V and occasionally shoulders on the reduction scan at -0.5 V and oxidation scan at -0.05 V. The electrode capacitance of the Ppy-pTS was significantly greater than the other electrode coatings; however the oxidation potential limit appeared to be reduced by up to 300 mV. The voltammetry of PEDOT-pTS was almost featureless, having only a small sigmoidal process visible around -0.3 V, otherwise the electrode coating had the largest capacitance within the useful potential window (before water oxidation and reduction).

EIS of each electrode was also performed, with bare iridium electrodes displaying a linear impedance (Z) versus frequency (f) Bode plot (figure 2b), relatively constant -70 to -80° phase angle versus frequency (figure 2c) and linear real (Z') versus imaginary (Z'') impedance Nyquist plot (figure 2d), all consistent with a metal electrode in contact with an aqueous solution, where the electrode-solution electrical path acts as a resistor and capacitor in series [33].

All of the conducting polymer coated electrodes had more complex impedance responses than the bare iridium electrodes, but were consistent with previously reported results [6, 19, 22, 32, 34]. Generally, the impedance decreased with increasing frequencies with a constant impedance plateau at intermediate frequencies (figure 2b). PEDOT-SO₄ approached a stable impedance over the frequency range of 1-10 kHz, Ppy-SO₄ had a reasonably flat Z at 100-10,000 Hz, Ppy-pTS only formed a plateau at frequencies below 1 kHz while PEDOT-pTS had the widest and most stable impedance from below 100 Hz to over 100 kHz. The impedance value at 1 kHz is typically used to compare electrodes for

neural recording [35] and decreased in the order of Ppy-pTS > uncoated > Ppy-SO₄ > PEDOT-SO₄ > PEDOT-pTS (table 1).

The phase angle versus frequency curves for the coated electrodes (figure 2c) were also -70 to -80° at 100 kHz, but approached 0° over the frequency range where the impedance versus frequency curves formed a plateau, before increasing again at lower frequencies as the Z values increased. The decrease in phase angle magnitude is associated with a more resistive behaviour of the electrode [33]. The phase angle measured at the peak became less negative (more resistive) in the order of uncoated < PEDOT-SO₄ < Ppy-SO₄ < Ppy-pTS < PEDOT-pTS. The order of phase angle at 1 kHz was slightly different with uncoated < Ppy-pTS < PEDOT-SO₄ < Ppy-SO₄ < PEDOT-pTS.

The Z' versus Z'' Nyquist plots highlight different behaviour between polymer coatings (figure 2d). Both doped PEDOT coatings had a semi-circular response at high frequencies (indicating a parallel capacitor and resistor equivalent circuit, and slow chemical kinetics at the electrode surface) which became a nearly vertical, pure capacitive response at lower frequencies. Ppy-SO₄ also had a semi-circular plot at high frequencies, but with a 45° Warburg (diffusion controlled) profile at low frequencies. In contrast, Ppy-pTS only displayed part of a semi-circular response over the entire frequency range tested.

A plot of the charge density, measured by integration of the oxidation sweep of the cyclic voltammogram (plots using the reduction sweep were almost identical) and using the nominal geometric area of 413 μm^2 , versus the impedance at 1 kHz revealed an interesting clustering between each electrode coating (figure 3 and table 1). The uncoated electrodes had a range of impedance values centred on 192 kOhm with a very low charge density around 3.1 mC cm^{-2} . PEDOT-SO₄ impedances were centred on 55 kOhm with a charge density of 5 mC cm^{-2} . PEDOT-pTS had the lowest impedance around 35 kOhm, with the charge density clustered in two regions of 25 and 80 mC cm^{-2} . Both Ppy coatings were more variable, Ppy-SO₄ had the lower impedance range of 100-200 kOhm with the charge density spread around 45 mC cm^{-2} , while Ppy-pTS had the highest variability with an impedance from 200-330 kOhm and charge density spread from 31-106 mC cm^{-2} . The charge density measured for each electrode material followed the same trend as the magnitude of the phase angle peak listed above (table 1). There was a significant effect of polymer coating on impedance, $F(4,58) = 76.44$, $p < 0.001$, where significant TUKEY pairwise differences on impedance was found between all coating and uncoated pairs but not between PEDOT-pTS and PEDOT-SO₄. There was also a significant effect of polymer coating on charge density, $F(4,58) = 39.719$, $p < 0.001$, where significant TUKEY pairwise differences on charge density was found between all coating and uncoated pairs except between PEDOT-pTS and Ppy-SO₄, Ppy-pTS and PEDOT-pTS, and PEDOT-SO₄ and uncoated electrodes.

After implantation in the neural tissue of rat models for several hours while performing acute auditory testing, the electrodes were removed, and retested by cyclic voltammetry and EIS to determine the stability of each coating. They were then placed in an enzymatic protein cleaning solution for 24 hours and again retested by cyclic voltammetry and EIS to gain an understanding of the amount and potential effect(s) of protein adsorption on the electrode surfaces. Prior to enzymatic cleaning, the voltammetry of uncoated electrodes after implantation appeared to have a slightly larger background capacitance with a larger gradient and smaller potential window than prior to implantation (figure 4a). The Faradaic currents associated with Ppy-SO₄ (figure 4b) and PEDOT-SO₄ (figure 4d) were both reduced in magnitude and the peak splitting (potential between reduction and oxidation peak currents) was increased while the potential window of Ppy-SO₄ was slightly reduced in the oxidation region. The post-implanted current associated with Ppy-pTS was significantly reduced; however the PEDOT-pTS voltammetry was almost identical after implantation to that observed before implantation. The Nyquist plots for each electrode material (figure 5) had a larger diameter semi-circle after implantation; the uncoated electrodes also possessing a semi-circle. The impedance at 1 kHz on uncoated electrodes and PEDOT-pTS coated electrodes was very stable, but increased for the other coatings (figure 6). There was a significant difference from before and after implantation across polymer coatings and uncoated electrodes on charge density ($F(9,54) = 10.551, p < 0.001$). Pairwise t-tests were conducted on charge density before and after implantation with significant difference for Ppy-pTS ($t = 3.310, p = 0.002$) and uncoated electrodes ($t = 2.061, p = 0.044$) but not for PEDOT-pTS ($t = 1.315, p = 0.194$), Ppy-SO₄ ($t = 0.324, p = 0.747$) or PEDOT-SO₄ ($t = 0.066, p = 0.951$). There was also a significant difference from before and after implantation across polymer coatings and uncoated electrodes on impedance ($F(9,54) = 35.189, p < 0.001$). Pairwise t-tests were conducted on impedance before and after implantation with significant difference for Ppy-SO₄ ($t = 6.250, p < 0.001$), PEDOT-SO₄ ($t = 3.901, p < 0.001$) and Ppy-pTS ($t = 2.460, p = 0.017$) but not PEDOT-pTS ($t = 0.146, p = 0.885$) or uncoated electrodes ($t = 0.076, p = 0.940$). After enzymatic cleaning (figure 7), the charge density on uncoated electrodes returned to near the preimplantation values without altering the impedance value at 1 kHz. The other electrode materials were barely affected by the cleaning process. There was a significant difference from before implantation and after enzymatic cleaning across polymer coatings and uncoated electrodes on charge density ($F(9,54) = 31.161, p < 0.001$). Pairwise t-tests were conducted on charge density before implantation and after enzymatic cleaning with significant difference for Ppy-pTS ($t = 5.998, p < 0.001$) but not for PEDOT-pTS ($t = 1.611, p = 0.113$), Ppy-SO₄ ($t = 1.358, p = 0.180$), uncoated electrodes ($t = 0.869, p = 0.389$) or PEDOT-SO₄ ($t = 0.125, p = 0.901$). There was a significant difference from before implantation and after enzymatic cleaning across polymer coatings and uncoated electrodes on impedance ($F(9,54) = 43.951, p < 0.001$). Pairwise t-tests were conducted on impedance before implantation and after enzymatic cleaning with significant difference for Ppy-SO₄ ($t = 8.567, p < 0.001$) and PEDOT-SO₄ ($t = 5.325, p$

< 0.001) but not uncoated electrodes ($t = 1.909$, $p = 0.062$), Ppy-pTS ($t = 1.180$, $p = 0.243$) or PEDOT-pTS ($t = 0.099$, $p = 0.921$). Overall, both PEDOT-SO₄ and Ppy-SO₄ had relatively stable charge densities after implantation and cleaning, but had an increase in impedance. Ppy-pTS had a large decrease in charge density after implantation and cleaning, however the impedance value was reasonably stable. Uncoated and PEDOT-pTS coated electrodes appeared stable in both charge density and impedance. No overall visible changes in voltammetry or EIS response of uncoated electrodes from before implantation to after enzymatic cleaning indicates little-to-no corrosion or other changes in electrode properties occurred when using a 24 hour cleaning process.

Electrophysiology of different doped conducting polymers on neural electrodes

The conducting polymer-coated electrodes were implanted into the IC of rats and then driven further into the IC in 200 μm steps while recording the neural response to various sound pressure levels of white noise. A typical plot of the SNR at 70 dB at each position in the IC of one electrode is shown in figure 8a. At the beginning of the experiment, this electrode is not in the IC, with a SNR of around 1. After 600 μm drive into the IC, the SNR increases to ~ 3.1 ($\text{RMS}_{\text{bgd}} = 8.3$ mV, $\text{RMS}_{\text{stim}} = 25.8$ mV). The electrode response then remains near this level ($\sim 2.6 - 4.1$) when driven for another 1.6 mm into the IC and when retracted to the same position before decreasing towards 1 again. This response curve is consistent across all other electrodes tested. Furthermore, the depth at which the electrodes display an increase in SNR correlates with the position of each electrode on the probe shanks. The relatively constant response at each electrode at different depths in the IC indicates little neural or electrode damage is occurring during the insertion and retraction process. The slight variation between electrode depths is therefore more likely due to different numbers of neurons within recording distance of the electrode. To reduce this effect, the neural response was averaged from all points where the electrode was considered “in” the IC as described in the methods section.

When plotted against impedance at 1 kHz (figure 8b), the RMS_{bgd} revealed a trend ($R = 0.61$) of increased background noise with increasing impedance (table 1). The RMS_{stim} had an inverse correlation with impedance at 1 kHz, increasing in gradient with larger sound pressure levels, although the distribution was larger than the background noise due to the more random nature of neuron excitation (biological noise). This resulted in an inverse correlation ($R = -0.59$) between the SNR and impedance at 1 kHz (figure 8c). The trend in SNR was PEDOT-pTS $>$ PEDOT-SO₄ $>$ uncoated $\approx$ Ppy-SO₄ $>$ Ppy-pTS (table 1). As expected, this trend ($R = -0.56$) was also seen for the spike count data (figure 8d). No correlations were seen between the maximum peak-to-peak amplitude outside the acoustic stimulation period and impedance. Nor were any correlations visible between RMS, spike count or peak amplitude response with the electrode charge density.

Electrochemistry of different PEDOT-pTS thicknesses on neural electrodes

To assess the effect of conducting polymer thickness on neural recording, the most promising doped conducting polymer was chosen for further investigation. PEDOT-pTS possessed the largest charge density, lowest impedance over the widest range of frequencies, most stable electrochemical response after implantation and the largest SNR. Therefore, 4 different deposition times were used to create a range of polymer thicknesses (figure 1). With a 15 s deposition time, a light blue film formed, and was confined to the electrode surface. With increased deposition times, the film became darker and spread onto the silicon substrate, increasing the electrode area (approximately 0%, 17%, 48% and 70% increase in geometric area with 15, 30, 45 and 60 s deposition times).

The electrochemical response was consistent with the initial PEDOT-pTS coated electrodes detailed above. Increased deposition times increased the electrode capacitance, the reduction process around -0.5 V became more pronounced and the oxidation potential window moved to less positive potentials (figure 9a). Longer deposition times also increased the frequency range with constant impedance, and reduced the impedance magnitude at 1 kHz (figure 9b). The peak in the phase angle also became less negative and began shifting to higher frequencies (figure 9c). The radius of the semicircle in the Z' versus Z'' curves decreased, with the vertical response at low frequencies only visible at long deposition times, a 45° Warburg process was instead present at the shorter deposition times (Figure 9d).

The plot of charge density versus impedance at 1 kHz (figure 10) shifts from a high impedance, low charge density for uncoated electrodes to low impedance and high charge density for thicker PEDOT-pTS coatings. However, there is a reasonably large variation in impedance values from these two probes, which are from a different batch than the two used above (table 1). Comparison of electrochemical response before implantation and after implantation with enzymatic cleaning doesn't reveal any trends with deposition times.

Electrophysiology of different PEDOT-pTS thicknesses on neural electrodes

The electrophysiological response when driving the electrode into the IC was similar to the response detailed above. The RMS_{bgd} was approximately 12 μV for all PEDOT-pTS thicknesses (figure 11a). Despite the larger standard deviation from these probes than seen in the previous dataset, a weak correlation of increase in SNR ($R = -0.35$) and spike count ($R = -0.31$) during acoustic stimulation with decreasing impedance at 1 kHz was seen with PEDOT-pTS deposition times up to 45 s (figure 11b-c and table 1). Once again maximum peak-to-peak amplitude outside the acoustic stimulation period displayed no trends with impedance at 1 kHz and no electrophysiological parameter correlated

with charge density when varying the PEDOT-pTS thickness. The variation in RMS_{stim} was larger than RMS_{bgd} , again highlighting the effect of biological noise.

Discussion

The electrochemistry of iridium and OCP-coated electrodes has been investigated in great detail, and their potential application as neural recording electrode surfaces regularly discussed in contemporary literature albeit on a largely singular non-comparative basis. Consequently, systematic comparison of the performance of specific materials' suitability for application as functional electrode surfaces remains lacking. Whilst it is true that a number of different OCPs have been characterised in depth, with the following references just representative of this [19, 20, 22-24, 32, 34, 36-43], operator-dependant variation of experimental impedance measurement renders integrated comparison of results from different publications and laboratories of limited value. Furthermore, understanding of how these novel conducting polymer coatings behave during and after implantation is also limited.

Microscopy and cyclic voltammetry of the coated electrodes showed an increase in electroactive area after conducting polymer deposition, which also increases with longer deposition times. The plot of charge density vs impedance at 1 kHz indicates that electrode surface modification can give rise to a wide range of electrochemical properties (figures 3 and 10). On this basis, judicious choice of conducting polymer, dopant ion and deposition time potentially facilitates tailoring of impedance and charge density, however certain coatings seem to possess larger response variability and consequent limitation of value in this regard.

Cyclic voltammetry indicated very different behaviour for each polymer. Both Ppy-pTS and Ppy-SO₄ have prominent redox peaks while doped PEDOT is almost featureless over the entire potential window (figure 2a). The presence of large Faradaic processes with Ppy coated electrodes is linked to the poor lifetime of this conducting polymer when repeatedly undergoing potential cycling, due to oxidation of the polymer backbone, while the stability of PEDOT coatings is significantly higher [44, 45]. A general rule thus arises that unless the Faradaic process associated with an electrode material is fully electrochemically and chemically reversible, a material which produces a voltammetric response without Faradaic peaks is recommended for a more stable electrode during chronic implantation and when used for electrical stimulation.

The charge density of both Ppy and PEDOT is greater when pTS is used as the dopant ion rather than SO₄²⁻, and is significantly larger than uncoated electrodes. This is mainly due to the large increase in electroactive area when coating the electrode surface and differences in morphology enabled by using different dopants [28, 32], but is also affected by the presence of surface-confined and diffusion-controlled Faradaic reactions and ion transfer between the polymer and the surrounding medium. No

correlations between charge density and neural recording response were seen, as the ability of an electrode material to inject a large charge is not associated with measuring potential; although this ability is important for stimulating electrodes.

The impedance of each electrode material is typically compared at only one frequency (1 kHz) and at this frequency decreased in the order of Ppy-pTS > uncoated > Ppy-SO₄ > PEDOT-SO₄ > PEDOT-pTS (figure 2b and table 1). However, the Z vs f curves for each conducting polymer coated electrode are non-linear and the power spectrum of multi-unit neural recordings extends from below 300 Hz to over 3 kHz [29]. Fortunately, the order of impedance for each electrode material was maintained over the frequency range of approximately 600-5,500 Hz when using the polymer morphologies described in this article, allowing a qualitative comparison of impedance at one frequency. Due to the variation in polymer morphology and subsequent EIS response from slight variations in polymer deposition and EIS measurement setups, for reproducibility, the impedance value at all relevant frequencies should be measured even when using identical conducting polymers and deposition times to those described in this article.

The phase angle of the electrode impedance is usually discussed in terms of electrode capacitance and resistance, with a -90° phase angle indicative of a capacitor inducing a current (voltage) lag from an applied voltage (current). Inspection of the EIS response of different polymer coatings therefore gives a clue to the degree of electrode capacitance present, with the phase angle becoming less negative (more resistive) from uncoated < PEDOT-SO₄ < Ppy-SO₄ < Ppy-pTS < PEDOT-pTS. This trend is similar to the charge density measured from cyclic voltammetry, with increased resistive behaviour correlating with a larger charge density. This trend is also similar to the measured geometric area, with increased area giving less positive phase angle and larger charge density. The trend of increased geometric area is not the same as reduced impedance.

Another method of viewing impedance is the Nyquist plot (figure 2d), which yields a curve from a high frequency response on the left to a low frequency on the right. The Z' value which intersects the axis at high frequency is due to the solution resistance, which was $1.2 \text{ k}\Omega \pm 400 \text{ }\Omega$ for all electrodes tested. The presence of a semi-circle on a Nyquist plot is due to an RC network or time constant, the radius being proportional to the charge transfer resistance or polarisation resistance (a measure of the electrochemical and chemical kinetics). The uncoated electrodes didn't possess a semi-circle while the radius of the semi-circle on the Nyquist plots of coated electrodes decreased from Ppy-pTS > Ppy-SO₄ > PEDOT-SO₄ $\approx$ PEDOT-pTS. This trend in charge transfer resistance is similar to the trend in impedance value at 1 kHz. The low frequency response then displays a 45° line for diffusion controlled behaviour as seen for the uncoated electrodes and Ppy-SO₄. The 90° line is caused by a frequency-independent capacitance, as visible for both PEDOT coatings.

After acute implantation but prior to enzymatic cleaning, the electrode charge density and impedance were re-tested. The implantation procedure didn't appear to damage the electrodes or the coatings (figure 8a), but did change the electrochemical properties (Figures 4-7). The uncoated electrodes displayed a slight increase in charge density (background capacitance), decrease in potential window and an increased uncompensated resistance, with a semi-circle forming in the Nyquist plots. The Faradaic peaks on Ppy-SO₄ and PEDOT-SO₄ were reduced in magnitude with a larger peak separation and the Nyquist plots displayed an increase in the semi-circle diameter or charge transfer resistance. Ppy-pTS had virtually no visible electrochemical response after implantation and a much larger charge transfer resistance. In contrast, PEDOT-pTS showed little effect from implantation leading to this polymer's being the best option of these polymers for use as a potential OCP-based neural electrode surface. These general features of reduced potential window, reduced Faradaic peak current, increased peak splitting, increased uncompensated resistance and increased charge transfer resistance on both SO₄²⁻ doped polymers and the Ppy-pTS are indicative of a fouling layer forming on the electrode surface, slowing or blocking electron transfer [46, 47]. Fouling of an electrode surface should reduce electrode capacitance, therefore the slight increase in background capacitance seen on uncoated electrodes is most likely a visual artifact due to the higher uncompensated resistance or a pseudocapacitance (surface confined Faradaic reaction) such as a redox reaction of the adsorbed fouling layer. Gently cleaning in an enzymatic protein remover was performed and the charge density and impedance tested again. After cleaning, the charge density was reduced on some uncoated electrodes, but remained stable for the coated electrodes, while the impedance response was little affected by the cleaning procedure. This implies that the fouling on uncoated electrodes was mostly protein, while non-protein compounds or a stronger surface association or different orientation of proteinaceous contaminants occurred on both SO₄²⁻ doped polymers and the Ppy-pTS polymer than on the PEDOT-pTS and uncoated electrodes.

The stability of the electrochemical response of the PEDOT-pTS coating after implantation and its largest SNR led to its choice for further investigation at different deposition times (figure 9). Increased polymer deposition times increased the electrode area, electrode charge density and Faradaic peak current, while reducing the potential window, impedance at 1 kHz up to 45 s deposition time and charge transfer resistance; the phase angle also became less negative and the Warburg diffusion profile at low frequencies at short deposition times was more capacitive with longer deposition. The plot of charge density versus impedance at 1 kHz generally shows a decrease in impedance and increase in charge density with longer PEDOT-pTS deposition times (figure 10). Once again, post-implantation electrochemical testing showed little variation, further highlighting the surface fouling aspect of implantation rather than a bulk material impact. The minimum impedance

occurring at 45 s deposition times indicates a trade-off with electrode properties such as electroactive area, morphology and bulk PEDOT-pTS conductivity.

Neural recording with the different electrode coatings and PEDOT-pTS deposition times indicated a correlation between RMS_{bkgd} (background noise) and impedance at 1 kHz (figure 8b and 11a), with an inverse correlation between impedance at 1 kHz and RMS_{stim} , SNR (figure 8c and 11b) and spike count during acoustic stimulation (figure 8d and 11c). A 45 s deposition time for PEDOT-pTS provided the largest SNR and spike count. The RMS_{bkgd} is mostly attributed to various electrical noise sources which are independent of electrode impedance and thermal noise ($V_{\text{rms}}^{\text{th}}$),

$$V_{\text{rms}}^{\text{th}} = \sqrt{k_b T Z \Delta f} \quad (1)$$

where k_b is Boltzmann's constant, T is the absolute temperature, Z is the impedance and Δf is the measuring bandwidth [48]. Typically, impedance at 1 kHz measured in a buffered solution, is used to model the thermal noise measured *in vivo*. However results presented here indicate that not only is the impedance value not constant nor even linear over the frequencies relevant to neural recording, the electrode impedance is also altered by protein adsorption or other fouling species. Therefore, an *in vivo* modelling of electrode-neural interface based on *in vitro* impedance values will likely not be correct. The reduction in RMS_{bkgd} with low electrode impedances allows detection of more spikes (larger biological noise) in multiunit recording, leading to the inverse correlation between RMS_{stim} and SNR with impedance [48, 49]. This should also result in a larger peak-to-peak amplitude [50], however measurement outside the stimulation period showed no correlation with impedance, and suggests that few spikes occurred during this time period. No attempt was made to measure the peak-to-peak amplitude during acoustic stimulation due to the large number of overlapping spikes.

The formation of a fouling layer on electrodes and its implications for neural recording and stimulation is not usually discussed. Electrode fouling occurs very rapidly on contact with blood or tissue, and is usually the first stage in scar formation and a cellular encapsulation process. These results show an increase in impedance and charge transfer resistance from protein fouling would have an effect on neural recording by reducing SNR and increasing the RMS_{bkgd} . Non-uniform fouling could then result in added variation between electrode responses. For stimulating electrodes, increased impedance and charge transfer resistance would increase the required charge needed to induce neural response which in turn would increase the power usage of any medical device. Irregular fouling on a stimulating electrode could then result in a non-uniform current density being applied to the tissue, and may induce damage. However, a large body of work on histology following neural stimulation has shown this to be of minor or no concern.

The differences in electrochemical and electrophysiological response with the different materials provide a useful guide for further studies. Doped Ppy coated electrodes showed larger impedance and RMS_{bkgd} , smaller SNR and spike count and generally had a larger error in response than uncoated and doped PEDOT coated electrodes. pTs as a dopant ion in either Ppy or PEDOT provided a larger charge density than SO_4 doped polymers. By applying the same experimental protocols detailed in this work, a comparison of more dopant ions, polymers and morphologies can be undertaken, proving more insight into the key parameters in controlling *in vivo* performance.

The large standard deviation from the second batch of electrodes does reduce the ability to correlate the electrochemical and electrophysiological responses; a more controlled comparison of the electrode material and electrochemical cell parameters in a single unit, *in vitro* setup may lead to a better understanding of the critical factors controlling the electrode impedance and hence further improve the signal to noise ratio of *in vivo* neural recording, although this would reduce the electrode fouling which is a critical factor during acute and chronic implantation. These studies provide important insights into controlling electrode properties for acute recording experiments; however their relation to chronic implantation and stimulation studies may be limited and requires more investigation [51].

Conclusion

Four differently doped conducting polymers were compared to bare iridium electrodes for use in neural tissue by electrochemical and electrophysiological techniques. A range of charge density and impedance responses were seen between each conducting polymer. The charge density for all coatings was greater than uncoated electrodes while PEDOT-pTS, PEDOT- SO_4 and Ppy- SO_4 possessed lower impedance values at 1 kHz than uncoated electrodes. The Nyquist plots for each electrode coating were also highly dissimilar, with significant differences in charge transfer resistance and capacitance, which provides details on the chemical properties and kinetics of the electrodes. Acute implantation in a rat model and subsequent retesting of electrochemical properties indicated that uncoated electrodes had significant protein adhesion, however cleaning in an enzymatic solution removed a large proportion of this protein. PEDOT-pTS showed little effect on electrochemical response after acute implantation, indicating good biostability. Increased thicknesses of PEDOT-pTS increased the charge density and up to 45 s deposition times reduced the impedance at 1 kHz, and also affected the charge transfer resistance and electrode capacitance.

Implantation of conducting polymer-coated electrodes into a rat model inferior colliculus and recording neural response to white noise bursts showed a decrease in background noise and increase in biological response, signal to noise ratio and spike count with reduced impedance at 1 kHz with different polymer coatings. PEDOT-pTS possessed the highest signal to noise ratio of the tested polymer coatings and a 45 s deposition time of PEDOT-pTS produced the highest signal to noise ratio

and spike count. This study shows that differently-doped polymers vary in their efficacy in application as surface coatings for neuro-functional electrode surfaces. Of the polymers studied here, PEDOT-pTS appeared to possess characteristics most optimal towards this application.

Acknowledgements

The authors acknowledge the support of the Australian Research Council through the Centre of Excellence for Electromaterials Science.

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Figure Captions

Figure 1: (a) Schematic of staggered electrode coatings with 4 different polymer types or thickness. Optical micrograph of (b) four thicknesses of PEDOT-pTS deposited on a NeuroNexus probe and higher magnification micrographs of (c) PPy-SO₄, (d) PPy-pTS, (e) PEDOT-SO₄, (f) PEDOT-pTS and (g-j) PEDOT-pTS at 15, 30, 45 and 60s deposition times.

Figure 2: Representative electrochemical response of uncoated and coated electrodes in 0.3 M Na₂HPO₄. (a) Cyclic voltammetry at a scan rate of 100 mV s⁻¹, (b-d) impedance response at 0 V with a 10 mV amplitude.

Figure 3: Impedance at 1 kHz versus charge density calculated from the oxidation sweep of a cyclic voltammogram and a quoted geometric electrode area of 413 μm² of 2 NeuroNexus probes coated with different conducting polymers.

Figure 4: Cyclic voltammetry before and after acute implantation without enzymatic cleaning in a rat model of (a) uncoated, (b) Ppy-SO₄, (c) Ppy-pTS, (d) PEDOT-SO₄ and (e) PEDOT-pTS coated electrodes in 0.3 M Na₂HPO₄ at a scan rate of 100 mV s⁻¹.

Figure 5: Nyquist plots before and after acute implantation without enzymatic cleaning in a rat model of (a) uncoated, (b) Ppy-SO₄, (c) Ppy-pTS, (d) PEDOT-SO₄ and (e) PEDOT-pTS coated electrodes in 0.3 M Na₂HPO₄ at 0 V with an amplitude of 10 mV.

Figure 6: (a) Charge density and (b) impedance at 1 kHz of different conducting polymers measured before and after implantation without enzymatic cleaning in the rat model IC for acute neural recording. The unity line has been added to aid visualisation.

Figure 7: (a) Charge density and (b) impedance at 1 kHz of different conducting polymers measured before implantation in the rat model IC and after implantation for acute neural recording and subsequent enzymatic cleaning for 24 hours. The unity line has been added to aid visualisation.

Figure 8: (a) SNR at 70 dB white noise bursts of a representative electrode when driven in and out of a rat model IC in 200 μm steps. (b) RMS_{bkgd}, (c) SNR and (d) mean during stimulation spike count versus impedance at 1 kHz of different conducting polymers.

Figure 9: Representative electrochemical response of uncoated electrodes and different PEDOT-pTS thicknesses in 0.3 M Na₂HPO₄. (a) Cyclic voltammetry at a scan rate of 100 mV s⁻¹, (b-d) impedance response at 0 V with a 10 mV amplitude.

Figure 10: Impedance at 1 kHz versus charge density calculated from the oxidation sweep of a cyclic voltammogram and a quoted geometric electrode area of $413 \mu\text{m}^2$ of 2 NeuroNexus probes coated with different PEDOT-pTS thicknesses.

Figure 11: (a) RMS_{bkgd} , (b) SNR and (c) mean during stimulation spike count versus impedance at 1 kHz of different PEDOT-pTs thicknesses.

Figures

Figure 1

CV 2640 – 9 Ir, 11 Ppy-SO₄, 2642-21 Ppy-pTs, 13 PEDOT-SO₄, 10 PEDOT-pTs

Imp 2640 – 14 Ir, 18 Ppy-SO₄, 28 Ppy-pTs, 1 PEDOT-SO₄, 10 PEDOT-pTs

Figure 2

Figure 3

Figure 4

Figure 5

Figure 6

Figure 7

Combined 2640+2642

Impedance vs outside stim, stim dif, Mean During Stimulation Spike Count 70dB

Figure 8

CV 118A – 3 Ir, 6 15s, 19 30s, 11 45s, 1180 8 60s

Imp 118A – 3 Ir, 12 15s, 19 30s, 5 45s, 1 60s

Figure 9

Figure 10

Combined 1180+118A

Impedance vs outside stim, stim dif, Mean During Stimulation Spike Count 70dB

Figure 11

Table 1: Average and standard deviation of oxidation charge density, impedance at 1 kHz and electrophysiological response at 70 dB acoustic stimulation for uncoated and coated electrodes.

Polymer Coating	Charge		Impedance at		RMS _{bkgd} / μV		SNR		Mean during	
	Density /mC cm ⁻²		1 kHz /kOhm						stimulation spike count	
	Ave	S.D.	Ave	S.D.	Ave	S.D.	Ave	S.D.	Ave	S.D.
PEDOT-SO ₄	5.1	0.5	55.5	4.3	8.4	1.0	3.4	1.3	15.7	6.8
Ppy-pTS	63.0	30.5	268.3	53.4	11.4	1.7	1.7	0.3	4.8	3.3
Ppy-SO ₄	45.6	13.7	146.2	76.7	9.0	1.0	3.1	0.5	16.1	3.7
PEDOT-pTS	54.6	31.3	35.0	6.4	7.2	0.4	4.1	0.8	20.3	2.2
Uncoated	3.1	4.5	192.5	10.0	8.3	0.7	3.2	0.5	16.3	3.0
15 s PEDOT-pTS	5.6	0.9	179.5	60.4	13.0	1.2	2.4	0.5	10.8	4.1
30 s PEDOT-pTS	11.7	1.9	140.1	62.0	12.4	0.9	2.6	0.6	12.6	4.3
45 s PEDOT-pTS	25.1	9.4	97.0	57.9	12.0	0.9	2.9	0.6	14.5	4.2
60 s PEDOT-pTS	33.7	3.9	109.6	63.3	12.0	0.7	2.6	0.5	12.2	3.3
Uncoated	3.8	6.2	179.6	67.0	12.5	0.9	2.3	0.5	10.2	4.1