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**Dendrites en polymère électrogénérées
pour du hardware à dépendance
morphologique appliqué au calcul
non-conventionnel**

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to unconventional computing**

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ABSTRACT

Across all domains of science, variability has often been perceived as a drawback. This remains especially true for microelectronics, as industrialization commands a high degree of reproducibility and repeatability. Yet, even though effort has been put into limiting this phenomena, organic electronics remains intrinsically variable by nature. Instead of fighting against it, would it be possible to use it to our advantage? With the limitations imposed by the current architecture of computers and the physical barriers to increasing density on integrated circuits, it appears essential to start exploring new routes to perform computing.

This work reports on the development of a new class of morphology-dependent hardware grown by AC electropolymerization. This easy-to-perform fabrication technique stands out not only as an alternative to traditional fabrication methods but as a novel fabrication route of its own, paving the way for the bottom-up growth of brain-inspired polymer dendrites that demonstrate unique structure-properties relationships.

First, we studied the influence of the electrochemical environment on the morphology and electrical properties of the electrogenerated PEDOT:PSS fibers, defining the conditions in which growth could be observed. We then demonstrated the ability of these unusual devices to perform brain-inspired operations such as short-term plasticity, and highlighted their potential as an electrochemical classifier, in particular for the detection of bursts of voltage. When grown into a network, cross-talking between fibers means that the network influences itself during operation as they share the same electrolytic environment. We took advantage of this behavior to perform unconventional computing with such polymer fiber networks, implementing quasi-linear to nonlinear operations. Due to the complex dynamics that arise in these systems, we demonstrated *in materio* computing abilities that result in the possibility to perform spatiotemporal information processing with these devices. The unusual and asymmetric morphology of polymer fibers, easily achieved through electropolymerization, is key to these properties. In addition, the intrinsic stochasticity of electropolymerization ensures the uniqueness of such systems. Finally, having highlighted the computing and detection possibilities offered by PEDOT:PSS dendrites, we explored the possibility to perform 2D *in vitro* electrophysiology with dendritic electrodes.

This work brings to light the advantages of electrogenerated objects and their unusual morphology in the framework of unconventional computing, paving the way for a new class of brain-inspired and morphology-dependent hardware.

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Abbreviations

AI Artificial Intelligence

ANN Artificial Neural Network

AP Action Potential

BQ Benzoquinone

BSC BioSafety Cabinet

CP Conducting Polymer

DIV Days *In Vitro*

DNN Deep Neural Network

EDL Electrical Double Layer

EEG Electroencephalography

FMRI Functional Magnetic Resonance Imaging

ITO Indium-tin Oxide

LFP Local Field Potential

MAC Multiply-Accumulate

MEA Multielectrode Array

NDR Negative Differential Resistance

OECT Organic Electrochemical Transistor

OMIEC Organic Mixed Ionic-Electronic Conductor

PBS Phosphate Buffered Saline

PEDOT:PSS Poly(3,4-ethylenedioxythiophene):Poly(styrenesulfonate)

PI Polyimide

RC Reservoir Computing

RIE Reactive Ion Etching

SNN Spiking Neural Network

SNR Signal-to-Noise Ratio

STDP Spike-Timing Dependent Plasticity

Part I

Manuscript

1

GENERAL INTRODUCTION AND MOTIVATION

I General introduction and motivation

"If you pour water over your PC, your PC will stop working. This is because very late in the history of computing - which started about 500 million years ago - the PC and other devices for information processing were developed that require a dry environment. But these new devices, consisting of hardware and software, have a disadvantage: they do not work as well as the older and more common computational devices that are called nervous systems, or brains, and which consist of wetware."

With these words, Wolfgang Maass opened his 2001 article entitled *wetware*¹. Why is that so? Because the technology that we developed for computing is essentially software and hardware, and does not support working in an electrolytic environment. Although a very simplistic and almost childish affirmation, this sentence invites us to take a step back and consider our definition of computing. In fact, we tend to forget that the act of computing is almost as old as life itself, and the reason is simple: life does not perform computing by moving electrons through metallic wires and semiconducting materials. Instead, the brains and nervous systems of living creatures actually consist of wetware: they operate in water and move information in the form of ions. As it has been around for many millions of years, wetware has become very efficient at what it does: receiving inputs from our five senses and choosing how to behave accordingly, thus producing an output. In other words, the brain and nervous system never stop computing for all of their lives. Moreover, to satisfy life's never-ending tendency for optimization, they developed the ability to do it very efficiently. Even now, in the wake of a new AI-driven revolution, and in the midst of many technological marvels, the brain remains an endless source of inspiration for researchers all around the globe, as it still keeps many secrets to unravel, and calls for a shift in our perception of what it means to compute towards a more biologically- and brain-inspired paradigm. If the last century was the century of the atom, the 21st century will be the one of the brain.

I.1 And then came the transistor

The invention of the transistor led to a revolution in the realm of electronics. Definitely the most important innovation of the field, it propelled the whole world into

a whole new era: the digital age. This technology had a tremendous impact on the way information was processed and transmitted. All of a sudden, digital electronics became prevalent over analog electronics and widely available, as the transistor slowly replaced the vacuum tube. The invention of the transistor was such a milestone that its inventors, Bardeen, Shockley and Brattain, were awarded the Noble price in 1956. However, most impressive yet is that over 70 years after its invention, the transistor is still at the very heart of today's world. Although the technology has been drastically improved and dramatically miniaturized, the silicon-based three-legged transistor remains the basic building block of integrated circuits.

Yet, digital computing has not always been the norm. For a long time, analog machines were used to perform calculations, and they did so pretty well. Already in ancient Greece, the Antikythera mechanism is thought to have been used for astronomical calculations². Compared to their digital counterparts, analog computers even have a few advantages: they excel at computing complex differential equations, allow real-time simulation and continuous representation of physical variables, and are often less energy demanding than digital computers. The major difference is that analog computers can take on a whole array of states/outputs.

At the end of World War II, digital computing slowly installed itself as the norm, eclipsing its analog cousin, until becoming the sprawling industry it now is. Yet, as the quantity of information generated daily spirals out of proportions and the solutions to keep improving our hardware become scarce, it might be time to reconsider our very idea of computing.

I.2 Reinventing technology

Interestingly enough, given the pace at which technology is advancing, the core ideas at the very foundations of electronics and modern computing remain surprisingly untouched. The transistor, the building block behind all of today's technology, dates back to 1947. Most of the effort led by the industry in order to increase the computational power of our computers has been to find new ways to downsize transistors in order to increase the density of components within integrated circuits. For some time, that proved fairly beneficial, as the number of components per integrated circuit doubled every two years: this is the infamous Moore's law. But that prediction turned into a bane as it became a self-fulfilling prophecy. As industry leaders were obsessed with the idea of increasing the density of components in integrated circuits, innovation in the field of electronics were let aside so that that all the effort could go into making smaller and smaller devices. Soon enough, Moore's law was coming to an end, as it became physically more and more difficult to bring down the size of transistors. As of today, transistors are of the scale of the atom^{3,4}, reaching the very limits of what it seems possible to achieve according to the laws of physics.

Another notably important idea that has not been challenged since it came around and still rules the way our computers are designed is the Von Neumann architecture, which was born around the same time as the transistor, in 1945. Within the framework of the Von Neumann architecture, storage and computing units are separated, which causes the data to move back and forth between the processing

and the memory nodes. Due to the increasing amount of data to be processed by energy-consuming algorithms, this results in a limitation of the performances of the system (due to the overloading of the communication buses) known as the Von Neumann bottleneck.

I.3 Bioinspiration

As everyone can tell from personal experience, learning comes with experimentation. In this regard, nature has many things to teach. For the last 3.5 billion years or so, nature has imposed a hard pressure on every living creature to adapt, and kept only the most fitted individuals through natural selection. As a consequence, nature abounds with the most ingenious adaptations to a whole array of hostile environments. For some time now, these adaptations have driven technological innovation.

Among the marvels to be found in nature, the brain stands out as a computing wonder. But more impressive yet is the ability of the brain to learn and adapt: even when some of its most important areas are damaged, it can rewire to be able to transfer the computing tasks that used to be performed by the damaged zone to another brain region. This plasticity makes it a unique tool for computing, along with its ability to massively parallelize and perform information processing at a low energy cost (it is often said that, in a resting state, the brain consumes about 20 W, which is less than a standard computer). Moreover, and maybe its most impressive feature yet, the brain is not specialized for a unique processing task: unlike IBM's Deep Blue that could only play chess or Deep Mind's AlphaGo which was designed to be the best at the go game, the brain regulates an immense number of vital functions in the body, and it not only integrates internal information but also stimuli from the external world captured by our senses. So far, we still have to invent a robot that could learn to play both basketball and the guitar. Drawing inspiration from the brain thus appears to be an exciting and promising prospect when it comes to finding new ways of computing.

I.4 The future of electronics and computing

How can we compute differently? That was one of the questions that led the research behind this manuscript. As technology advances, we are now facing exciting challenges that require to delve into novel ideas. These challenges will ask us to try peering into the future to anticipate the needs of tomorrow, and to explore new ways to envision electronics and perform computing.

Has the reign of all-mighty silicon come to an end? Although it has been the cornerstone of the semiconductor industry for as long as one can remember, new materials are now emerging that could challenge its previously uncontested superiority by opening up new possibilities. In the future, electronics could be smarter, flexible and seamlessly integrated to our body. Instead of being discrete blocks of rigid material, the electronics of tomorrow could be integrated to every aspect of our lives, with the ability to evolve and learn depending on the task to be performed. This enhanced electronics could have tremendous applications in biomedicine and play a major role in the development of personalized medicine: delivering drugs as

close to the target as possible in the body and in the most efficient way for the treatment, analyzing patterns in biomarkers to monitor one's health and prevent diseases instead of the cure-oriented approach of today's western civilization, new generations of prosthetics establishing bidirectional and closed-loop communication with the nervous system. . .

All of this will generate a massive amount of data that will have to be processed and transferred between nodes in this gigantic web of interconnected smart devices. Under the current model, the movement of data between integrated and wearable devices and processing nodes will require incommensurable energy resources, without even mentioning the cost of implanting the infrastructure to support the demand for this ever-increasing interconnection. With the climate crisis looming, this sweet dream almost feels anachronical. Yet, the deployment of 5G has not even finished that 6G is already being discussed. . . Silicon cannot be at the heart of this next wave of technology: it is a rigid material, with highly energy-consuming processing routes which are not compatible with a more bottom-up, flexible and on-the-go approach, and which extraction comes at a high cost for the environment. Thus, a new unified framework is required for the development of an all-compassing electronics, and organic electronics might be an interesting starting point to dig into.

What can we hope for in the future? In its way, this thesis aims at contributing to that discussion by offering its own original take on the matter by answering the following question: how can the bottom-up approach enabled by unusual electrogenerated objects such as polymer fibers be used for the development of a new generation of brain-inspired organic and metamorphic devices?

I The emergence of a biologically-inspired computing

I.1 Introduction

The invention of the transistor in 1947 fostered the era of digital electronics⁵. All of a sudden, it became possible to modulate an electrical current via an electrical parameter (in which case a potential). The transistor quickly came to replace the electromechanical relay and the vacuum tube⁶, as it was quicker to operate, more energy efficient and had more potential for miniaturization. Downsizing the transistor means that the density of components on integrated circuits could be increased and soon enough, billions of them were assembled into Graphic and Central Processing Units (GPUs and CPUs), developing an ever more powerful silicon-based technology. Yet, in a world where the quantity of data generated spirals out of control while the need for more efficient electronics keeps pressing, two main issues arise. First, transistors are starting to reach a critical size and the race for smaller devices seems to have reached a dead end: Moore's law has come to an end⁷. Second, information processing is now limited by the cost of moving information between the memory and the processing units, which is a consequence of the architecture of modern computers⁸. This so-called Von Neumann bottleneck thereby invites us to rethink about the way computers are built, and more generally about what it means to perform computing.

I.2 The Von Neumann architecture

The Von Neumann architecture, first described in 1945, is a model for the design architecture of digital electronics⁹. Within this model, the processing and memory units are distinct, but they share a common bus for the transmission of data. This leads to the limitation of the transfer rate known as the Von Neumann bottleneck, as the CPU must constantly wait for information to be moved back and forth between the memory of the system and its arithmetic logic unit.

Given the pace at which CPUs are now able to perform calculation, effective processing speed is thus limited by this throughput. In addition, the rate of im-

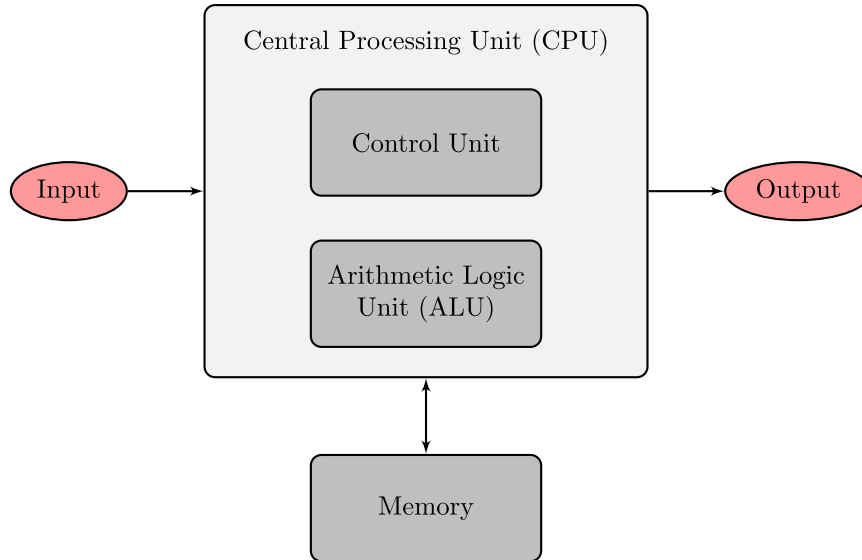


Figure 2.1: Simplified representation of the Von Neumann architecture.

provement in microprocessors has for long increased that of memory speed, resulting in what has been coined the memory wall¹⁰: a large amount of energy and time is dedicated to data transport. As data is moved in the form of an electrical current, power loss can be calculated by Joules' law: $P_{Joules} = RI^2$. This power is actually transformed into heat, which in turn limits the speed and precision of the system. Some studies showed that energy consumption could increase by 5% with every 10-12°C increment¹¹. After a certain threshold (typically around 100°C), the system deliberately reduces power to avoid damaging the CPU and reducing its lifespan, which in practice translates to limitation of the clock frequency and performance loss.

Although the Von Neumann architecture has proved more than useful for the development of modern technology, the bottleneck described in the previous paragraphs nevertheless invites us to reconsider the way we think about computing. In fact, digital computing is a very recent trend in the history of information processing. Before the invention of modern electronics, analog computers were the norm. For a very long time, analog machines were used to perform a whole array of tasks, such as predicting astronomical positions and eclipses, evaluating mathematical operations (multiplications, logarithms, trigonometry ...) or even solving differential equations. Still today, mechanical and automatic watches are analog computers used to keep track of time, some of them even including complex complications like a perpetual calendar. There are many alternatives to digital computing that would be deemed unconventional, even though some of them might be anterior to the invention of the computer. From a larger perspective, processing massive amounts of data requires the ability to parallelize the computing tasks. In this regard, leaving behind the Von Neumann architecture to find new ways to perform information processing is definitely needed.

I.3 Unconventional computing

Taking a step back, one might ask a seemingly naive question: what does computing mean? In its most usual definition, a computing system establishes a deterministic relationship between an input and an output. Yet, all computing systems do not comply with this definition, as some may involve randomness or uncertainty, especially brain-inspired approaches^{12,13}. This probabilistic behavior is related to the physics of the substrate chosen for computation: although silicon has become the cornerstone of microelectronics, the biological world exploits the resources available in its environment to perform computing tasks, and for an astonishing result^{1,14}. If on paper, the slow movement of ions in a noisy aqueous electrolyte is not the best fit for effective computing, in reality the operation of the brain and its efficiency are still a mystery.

The field of unconventional computing abounds with fascinating ideas that challenge the traditional digital computing paradigm. Under the umbrella of unconventional computing, many different substrates and physical effects have been explored, from biochemical reactions to optical or spintronic properties. Already in 2002, van Noort and colleagues proposed to use DNA computing to determine the largest number of fully interconnected nodes in a graph¹⁵. Other algorithmic problems, such as finding the shortest path, have been solved using the flow of current induced by a pH gradient¹⁶, or even microorganisms such as slime mold¹⁷. Exploiting the quantic properties of the electron also appears to be an efficient way to solve specific and complex problems¹⁸. Using a more classical substrate and building upon the already existing CMOS technology, Tour *et al.*^{19,20} proposed to take advantage of the randomness observed in self-assembled approaches of molecular electronics by operating a post-fabrication configuration of their NanoCell, effectively demonstrating that their device could be efficiently used as a memory and perform logic operation despite the disordered organization of the material. By doing so, they showed that precise patterning was not always necessary to perform computing applications.

Apart from physical implementation, the world of computing can also benefit from a redesign from an algorithmic point of view. In automata theory, a cellular automaton consists of a grid of cells, each of which can take a finite number of states. These cells can evolve in time and change state, depending on their previous state as well as the state of their neighbors. One fascinating observation about cellular automata is that, even if the local rules guiding the evolution of the system is extremely simple, after a few steps the global network thus created can exhibit a deeply rich and complex behavior (Figure 2.2), so much so that it inspired Stephen Wolfram a complete and fundamental theory of physics and the universe²¹. Such patterns can also be found in nature, for instance on the shell of some mollusks (Figure 2.2b).

The most famous cellular automata is the Game of Life, designed by John Conway in 1970²³. Despite its apparent simplicity, the Game of Life has been demonstrated to be Turing-complete²⁴, meaning that it has the ability to solve any problem that has an algorithmic solution and proving at the same time the computational power of otherwise overly simplistic rules.

Within the large framework of unconventional computing, neuromorphic com-

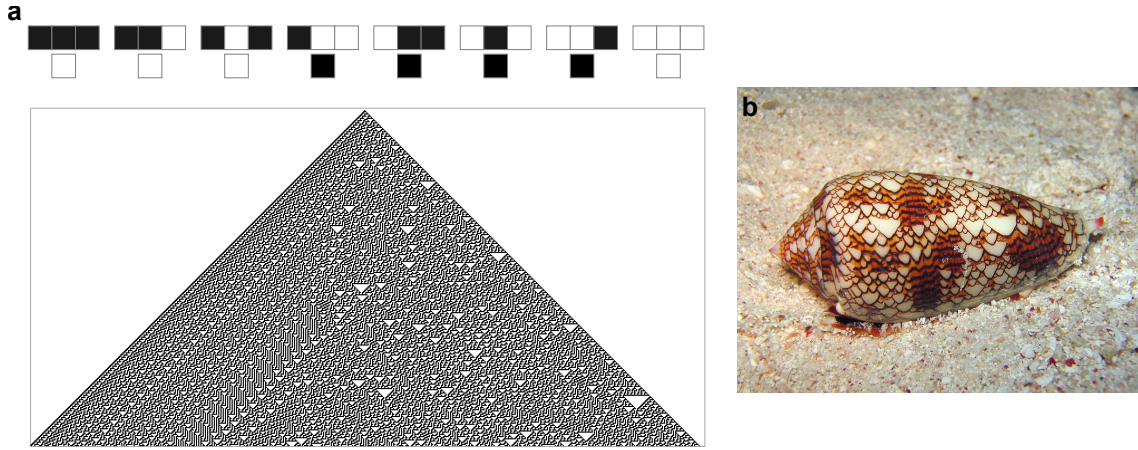


Figure 2.2: Cellular automata. a) Evolution of a cellular automaton following Rule 30 (shown at the top of the image) after the first 300 steps, starting from a single black cell. Time is thus represented along the y-axis. This visual representation evidences the complexity of the global pattern, showing ordered (left) as well as random (right) organization. Reproduced from [21]. b) Conus textile shell, showing a pattern similar to Rule 30. Reproduced from [22].

puting draws inspiration from the structure and function of the brain, taking advantage of analog values and stochasticity, enlarging the scope of the binary and deterministic world of digital computing.

I.4 A new brain-inspired paradigm

The brain is an astonishing source of marvel and inspiration, due to its ability to massively parallelize information processing at an impressively low energetic cost. Contrary to the fixed architecture of silicon-based technology, the brain has an innate ability for adapting and learning. These features have led to the development of neuromorphic engineering, an emerging field of research that aims at mimicking and emulating the operation of biological neural networks.

Neuromorphic electronic systems

Carver Mead was one of the pioneers of neuromorphic computing, looking for inspiration in the biological realm and exploring the reasons behind the energy-efficiency gap that existed between the brain and the digital technology of the time. Doing so, he drew attention to the similarity between the curve showing the current through the cell membrane of a neuron as a function of voltage across the membrane and the current flowing through a transistor as a function of the gate voltage (see Figure 2.3)²⁵.

Mead became an early advocate of the development of biologically-inspired technology²⁶. At the same time, bioinspired concepts had already begun emerging in software implementations, where their benefits were demonstrated before being applied at the hardware level.

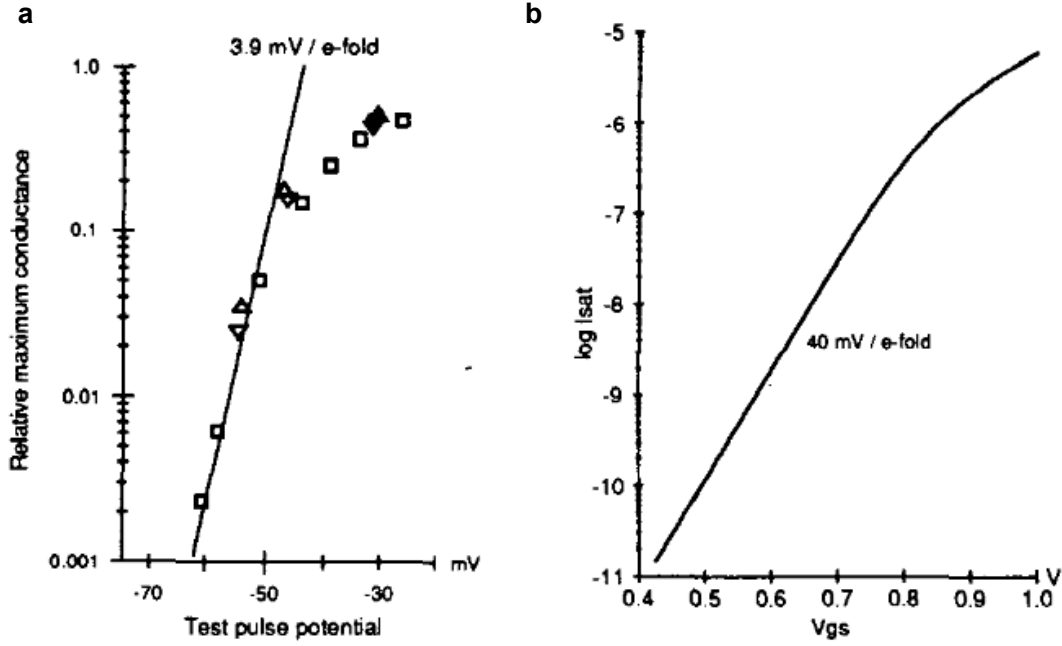


Figure 2.3: Current-voltage curves across a) a cell membrane and b) a transistor calls for comparison. The most important observation to be made is that in both cases the current increases exponentially with the voltage. Adapted from [25].

The perceptron

Artificial Intelligence (AI) and Artificial Neural Networks (ANNs) are without a doubt the most famous brain-inspired computing paradigm today. ANNs consist of a succession of artificial neurons (the nodes) connected together through edges meant to represent biological synapses. An artificial neuron can be represented as shown in Figure 2.4.

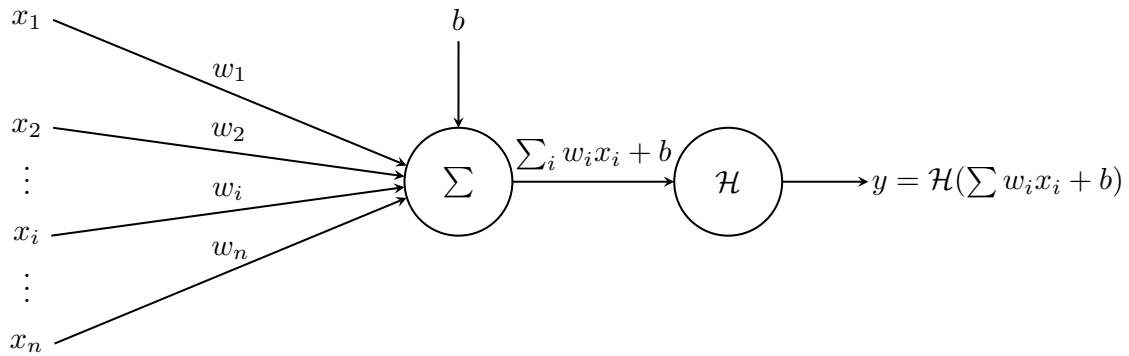


Figure 2.4: Structure of an artificial neuron.

Each neuron is presented with a series of inputs that are weighted and summed before entering an activation function. This activation function is actually a threshold: if the weighted sum of the input crosses that threshold (the value of b in the present example), then the output of the neuron will be ‘1’. Conversely, if the threshold is not met, the output of the neuron will be ‘0’. Formally, the perceptron can be described by the following equation, using the Heaviside step function as the activation function, such as represented in Figure 2.4:

$$y(x) = \mathcal{H}(w \cdot x + b)$$

where

$$\mathcal{H}(x) = \begin{cases} 1, & \text{if } x \geq 0 \\ 0, & \text{otherwise} \end{cases}$$

The simplest binary classifier can be made from a single artificial neuron, and is called a perceptron. A single perceptron is a linear classifier, meaning that the classification decision is based on the evaluation of a weighted sum of the inputs of the system. In practice, this limits the usefulness of a single perceptron, as more complex and non-linear functions cannot be apprehended by such as simple network. From a Boolean logic perspective, a perceptron can implement basic functions such as AND, OR and NOT, but is unable to perform the exclusive-OR (XOR) operation as it is not linearly separable. A single perceptron is therefore not functionally complete. This issue can be solved by implementing artificial networks with several layers of neurons.

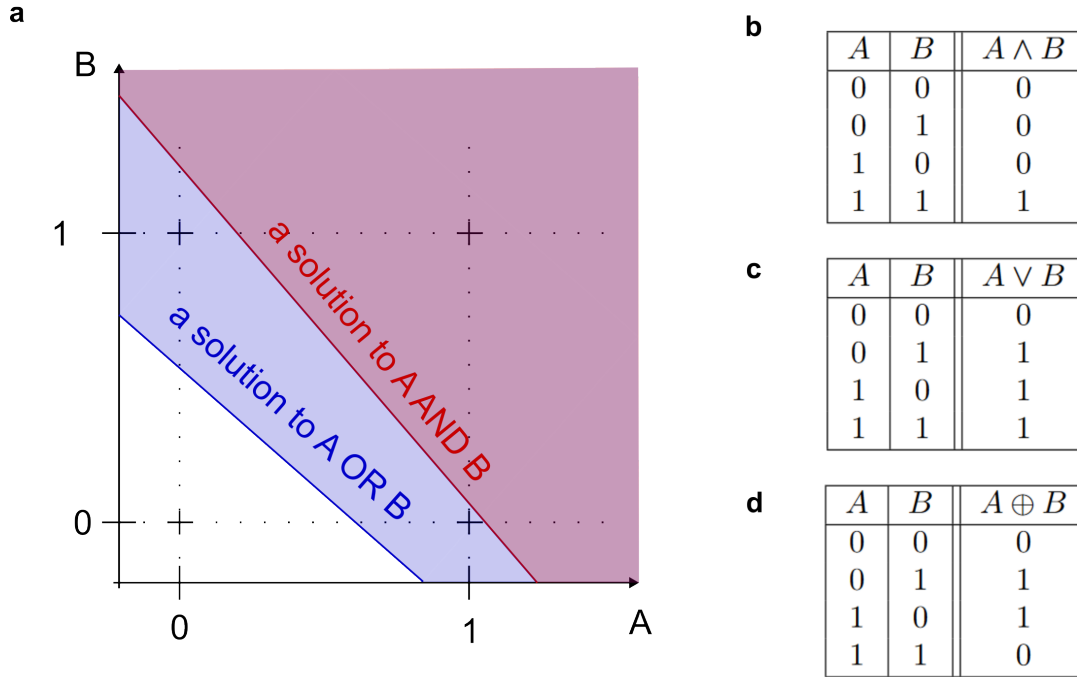


Figure 2.5: Boolean logic with a single perceptron. a) Examples of linear solutions to the AND and OR problems. Truth tables of the b) AND, c) OR and d) XOR functions. The single perceptron is not functionally complete as it cannot implement a XOR function.

I.5 Artificial neural networks

Today, Deep Neural Networks (DNNs) are built on many hidden layers of artificial neurons, such as the perceptron presented in the previous section. The latest deep learning revolution has drastically advanced fields such as computer vision or natural language processing²⁷. Yet, deep neural networks necessitate vast amounts of data to be trained on due to their gigantic number of parameters (GPT-3 has

175 billion parameters, a number that should be significantly higher for GPT-4). Consequently, the training of these models is extremely energy-intensive.

In contrast, biological systems have evolved to optimize energy consumption, so much so that bioinspired strategies have already been successfully implemented by researchers to address many challenges. For instance, evolutionary algorithms are a class of optimization algorithm that use an initial set of candidate solutions to an optimization problem, and make that population evolve through selection, crossover and mutation, until reaching an optimal solution. Within this class, genetic algorithms might be the most popular subset, and the first population-based stochastic algorithm to have been explored historically²⁸. Drawing inspiration from Darwin’s theory of evolution and natural selection, genetic algorithms randomly choose the fittest candidates within the population to create the next generation of solutions according to their fitness score. Solutions with high fitness values are more likely to be selected. This example highlights the benefits of incorporating stochasticity into optimization problems. Although DNNs are inspired by the operation of the brain, their structure is fixed *a priori*, and they lack the ability to evolve and adapt that has made biological systems so resilient and efficient.

Reservoir Computing (RC) is another brain-inspired computing paradigm that has attracted a lot of attention within the deep learning community²⁹. In a reservoir, the input is non-linearly projected into a high dimensional space before being mapped by the readout layer. The popularity of reservoir computing is mainly due to two factors. First, the reservoir acts as black box, of which only the readout layer has to be trained, greatly diminishing the time and energy required to train the system. Second, this kind of paradigm can virtually exploit any physical system, provided that the dynamics of the system present nonlinear characteristics. This computing paradigm started stimulating the interest of the scientific community after the publication of two seminal papers in 2001 and 2002, the first one by Jaeger presenting the “echo state network”³⁰, and the second one introducing Maass’ “liquid state machine”³¹.

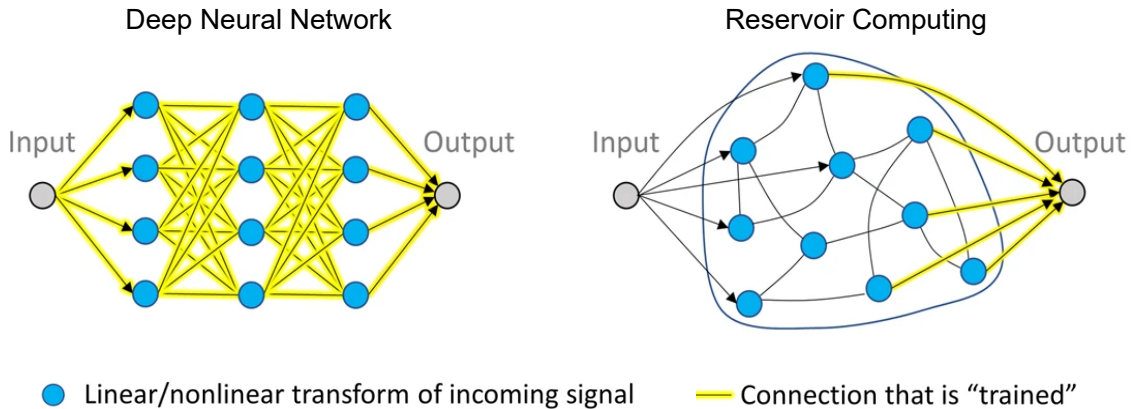


Figure 2.6: Deep Neural Network vs Reservoir Computing. In DNN, all the weights of the network have to be trained, versus only the readout in RC. Adapted from [32].

One of the most iconic representations of a reservoir was proposed by Natschläger

and colleagues³³, in which the reservoir was compared to water surface. This idea was later physically implemented, showing that it was indeed possible to exploit the patterns created by the waves on the surface of the liquid to be used as a reservoir³⁴. Many physical systems have since then been explored for the development of physical reservoirs, exploiting the nonlinear properties of electronic circuits³⁵, quantum systems³⁶, spintronic materials^{18,37}, or even biological substrates³⁸, amongst many others^{39–41}.

In the context of the current energy crisis, it seems highly necessary to explore other options to perform computing differently. In this regard, brain-inspired computing paradigms are an interesting approach, as they combine classic machine learning techniques with a search for novel materials with nonlinear properties, all the while looking for inspiration in the biological world⁴². On the hardware level, this path that has been followed by the neuromorphic community, drawing inspiration from the function and structure of the brain.

I.6 Neuromorphic hardware*

In an attempt to make deep neural networks more energy efficient and biologically relevant, Spiking Neural Network (SNN) were developed^{43–46}. Contrary to DNNs, SNNs use discrete spikes to transmit information between the neurons of the network and are event-driven instead of synchronous. SNNs were introduced to overcome the limitations imposed by the Von Neumann architecture and to improve in pattern recognition tasks (something that the brain is very good at)⁴⁷, but also because they possess some advantages over traditional ANNs when it comes to hardware implementation. Notably, Maass demonstrated that any computation performed with an ANN could also be performed with a SNN, while requiring fewer computing element⁴³. The sparsity of spike trains also makes SNNs less energy-intensive than other deep learning networks, an advantage that is translated to hardware and reduces power consumption⁴⁸. Hardware implementations have thus followed the development of SNNs to propose a large-scale integration of artificial neurons on Complementary Metal-Oxide (CMOS) technology^{49–52}. The potential of this approach is important, as demonstrated by the unmatched energy consumption obtained by silicon artificial neurons such as the one developed at IEMN by Sourikopoulos and coworkers⁵³, which energy consumption amounts to 4 fJ/spike. Being implemented on a traditional CMOS technology means that neuromorphic chips optimized for SNNs are ready to be industrialized. Two of the major tech companies in the world, Intel and IBM, have already commercialized their own neuromorphic chips, namely Loihi⁵⁴ and TrueNorth⁵⁵.

Apart from these neuromorphic chips, neuromorphic hardware has also been used to implement sensors mimicking the functions of biological sensory organs such as the retina^{56,57}, the cochlea⁵⁸ or the skin⁵⁹. On the component level however, one very

*In the literature, the terms “neuromorphic” and “brain-inspired” are often used interchangeably. In this manuscript though, I will mainly use “brain-inspired” to refer to any computing paradigm or concept inspired by the function of the brain, while “neuromorphic” will be restricted to hardware considerations. In this sense, neuromorphic computing becomes a subset of brain-inspired computing.

special electronic component for neuromorphic hardware that is worth mentioning here is the memristor.

The memristor

Synapses are extremely important elements in biological neural networks, yet for a long time there was no simple electronic equivalent to them. The memristor, conceptualized by Leon Chua in 1971⁶⁰, was only physically implemented in 2008⁶¹. A memristor is actually a two-terminal variable resistance which conductance state depends on the voltage applied across its terminals. In that sense, a memristor acts as a non-volatile memory, given that its resistivity depends on past events. This effect is the result of a pinched hysteresis, such as the one presented in Figure 2.7.

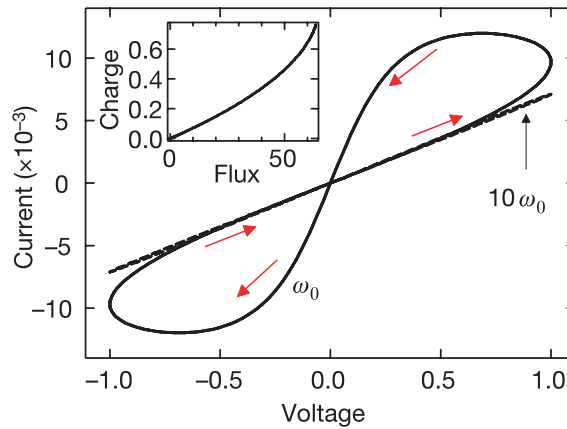


Figure 2.7: I-V curve of a theoretical memristor, showing the pinched hysteresis. Adapted from [61].

In addition, memory in these devices is embedded in their physical state, and can consequently be stored indefinitely, contrary to conventional electronics. The discovery of the memristor reignited interest in the development of artificial neural networks and neuromorphic hardware, as an electronic component could now emulate a synaptic connection and the corresponding synaptic plasticity^{62–65}.

Limitations of neuromorphic hardware

If the hardware implementation of spiking neural networks delivers on the promise to cut energy consumption compared to general-purpose computers limited by the von Neumann architecture, there is still a series of limitations regarding neuromorphic computing that need to be addressed⁶⁶. In hardware learning is still a challenge for neuromorphic chips. If backpropagation has been adapted to function with deep SNNs⁶⁷, learning in the brain is shaped by an impressive number of local synaptic rules that neuromorphic hardware should aim at replicating⁶⁸. On an architectural level, adding neuromorphic building blocks on top of a preexisting von Neumann basis means that the limitations of such a computer architecture still hold true (see I.2). An ideal architecture should therefore increase the degree of parallelism with a high interconnectivity between the distinct neuronal units, each of which should be able to perform computing and to implement unsupervised learning rules locally⁴². In addition, current neuromorphic hardware still suffers the same drawbacks as CMOS

and would benefit from the emergence of new technologies and materials. Notably, silicon-based technology lacks in flexibility, and while silicon itself is not cytotoxic, it might trigger immune response or degrade over time, requiring any Si-based device to be adequately packaged to avoid long-term concerns. It also does not show high biological realism, which hinders the development of brain-computer interfaces. Moreover, this technology is highly energy-demanding, and still subjected to Moore’s law, which end has already been commented by many observers.

Brain-inspired computing resides at the intersection of many areas of research that are tightly interwoven, as schematized in Figure 2.8.

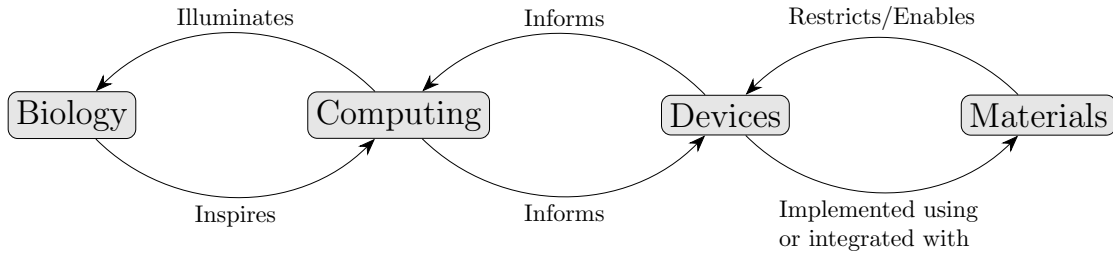


Figure 2.8: Areas of research involved in brain-inspired computing. Reproduced from [69].

Solving the previously mentioned issues related with neuromorphic hardware might therefore come from two different approaches:

- Diving into the biological world to deepen our understanding of how computation is performed in living creatures.
- Exploring how emerging devices and materials could enable new computing abilities. The field of neuromorphic hardware will necessitate novel materials and a deep understanding of their physical properties to perform biologically-plausible computing⁷⁰.

The next section of this manuscript will therefore attempt to provide the reader with a sense of the rules that govern electrophysiology in the brain, while the last part of this chapter will be dedicated to showing how organic polymers might be an interesting class of materials for the development of a new generation of neuromorphic hardware.

II Computing in the biological world

Understanding how the brain works may be one of the biggest intellectual challenges of the 21st century. Despite the recent technological and scientific advances made in many fields, the brain remains to this day mostly a mystery and many questions are still left unanswered: how does it process information in such an efficient way? How do we store memories? What is consciousness and how does it emerge from electrical activity?

Our comprehension of quantum physics, which study started barely a century ago, seems to develop faster than our knowledge of the brain, as it is now possible

to use that theory to realize computing tasks. It is fascinating to think that in little over a lifetime, the understanding we have of the atom has enabled us to master nuclear energy and to create the most powerful computers ever, while the brain, a massive aggregation of about 1.4 kg of matter that has been there for as long as humanity, does still not unravel all its secrets.

II.1 The structure of the brain

The brain is a fascinating computing machine. With around 86 billion neurons and 10^{16} synapses to form interconnections⁷¹, the brain presents an unmatched and innate ability for highly parallelized signal processing, due to its unique structure. It is made of a gigantic network of interconnected cells, where each neuron is estimated to communicate with about 10^4 other neurons. If the way neurons work individually has been thoroughly studied over the past decades and is now understood in considerable detail, how colossal amounts of them cooperate at the scale of neural networks is yet to be fully comprehended. In addition, the brain is not only made of neurons, but is rather a complex entanglement of many types of cells (microglia, astrocytes...) that all have a role to play in its operation, increasing the overall complexity of the structure.

The neuron: the basic unit of the brain

The neuron is the basic structural and computational unit of the brain, composed of a cell body (or soma) from where dendrites originate and collect information from surrounding neurons. Information is then transmitted in the form of action potentials (all-or-nothing electrical phenomena that can be thought of as ‘0’ and ‘1’) through the axon and synapses that establish connections with between neurons.

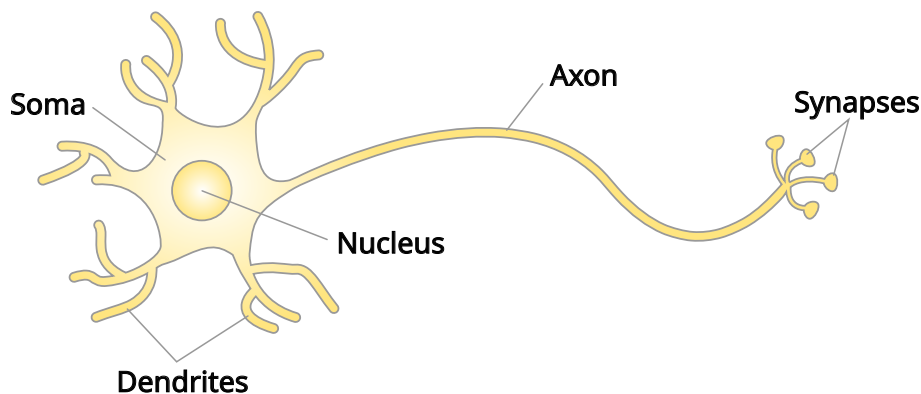


Figure 2.9: Basic structure of a neuron.

The rules governing computing at a biological level operate on very different principles from those of silicon-based technology. The brain is a highly parallelized system, working in an inherently noisy environment. Yet, it remains a very powerful, resilient and reliant computing machine, with an ability to learn, evolve and adapt over its whole lifespan. Contrary to traditional electronics, the brain is asynchronous, meaning that information is not pushed forward at a pace imposed by an external clock, but rather computed as it comes and when needed. Biological neural networks demonstrate unrivalled prowess when it comes to processing ill-conditioned

input data, all the while being slow from a conventional point of view: the brain uses ions in an aqueous environment instead of free electrons in a conductor to transmit information, and action potentials are the fastest events recorded in the brain at about 1 kHz, far from the GHz of the most advanced processors today. In addition to that, the energy consumption of the brain is orders of magnitude lower than that of any supercomputer, 20 W being the most recurrent number for the resting brain⁷².

The synapse

Within the nervous system, nerve cells communicate with each other via synapses. A synapse is a junction between two neighboring neurons (namely the presynaptic and the postsynaptic neurons) or between a neuron and an effector cell. Its role is to assure the continuity of the electrochemical signaling between cells. At the end of the axon of the presynaptic neuron, the electrical signals, transmitted in the form of electrical impulses known as action potentials, are passed to the postsynaptic cell by a synapse. There are two main modalities of synaptic transmission:

- In a chemical synapse, the electrical activity of the presynaptic neuron is converted into chemical signaling through the exocytosis of neurotransmitters (e.g., dopamine, GABA, acetylcholine, glutamic acid or norepinephrine). Those molecules cross the synaptic cleft and bind to the receptors of the postsynaptic cell membrane, initiating a response. This type of synapses is the most common one.
- In electrical synapses, the pre- and postsynaptic cells are connected via a gap junction that allows the electric current to flow directly from one cell to the next, therefore increasing the speed of transmission compared to chemical synapses.

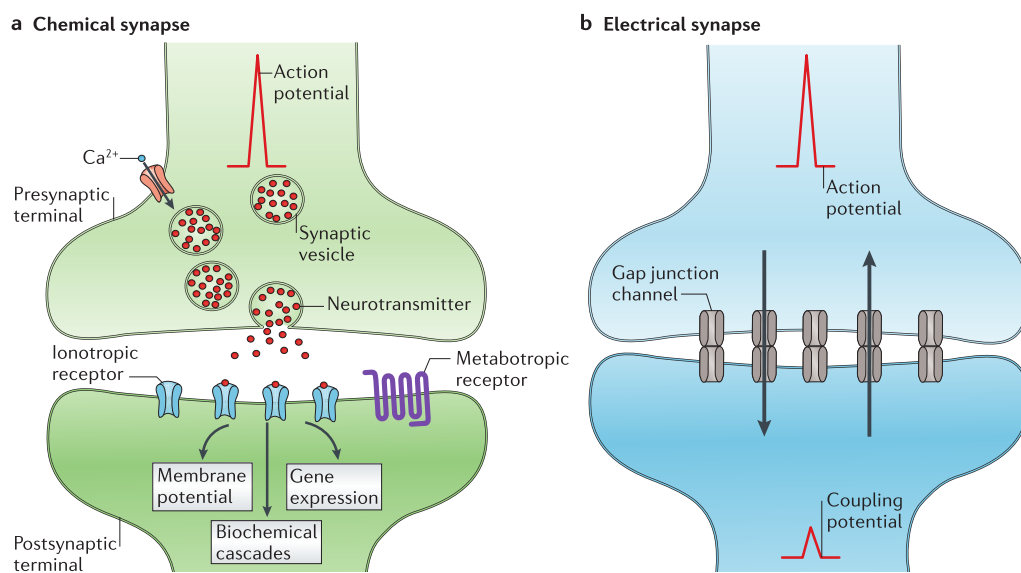


Figure 2.10: The two modalities of synaptic transmission. a) A chemical synapse. b) An electrical synapse. Reproduced from [73].

Apart from signal transmission, synapses are also critical structures that have an important role to play in processes such as learning and memory⁷⁴.

II.2 The learning brain

One of the main advantages of living matter over traditional electronics is its innate ability to adapt and evolve through a process called learning. In the brain, learning is implemented through synaptic plasticity, which is defined as the ability of synapses to modulate the efficacy of the transmission of a given signal. There exists a great diversity of local learning rules governing synaptic plasticity. However, these rules can be classified into two main forms of synaptic plasticity: functional and structural plasticity.

Functional plasticity

Functional plasticity refers to the alteration of the function of synapses and neural circuits without any physical modification of the network itself. Functional plasticity often reflects immediate changes in synaptic weights.

The Hebbian learning rule might be the most famous attempt to explain synaptic plasticity. Hebb's rule stipulates that, when two adjacent neurons are active simultaneously, the weight of the connection between these neural cells is increased, meaning that their connection is strengthened. This behavior, observed by Hebb in 1949, is famously referred to as the Hebbian learning rule, and summarized as follows: "cells that fire together, wire together"⁷⁵. This might happen though a change in the number of receptors on the postsynaptic membrane, via the modulation of the released quantity of the neurotransmitter, or because of a change in the sensibility of the receptors to said neurotransmitter. The concept of Hebbian learning and the temporal causation that comes with it has been observed in a synapse-specific manner in what is known as Spike-Timing Dependent Plasticity (STDP). In addition to the observation made by Hebb about the synaptic weight potentiation when the presynaptic cell fires just prior to the postsynaptic cell, STDP stipulates that when the activity of two neighboring neurons is negatively correlated, the connection between them is depressed⁷⁶. This process is intimately correlated with the activity-dependent development of the nervous systems and is one of the drivers of long-term potentiation and depression.

Structural plasticity

Structural plasticity involves physical modifications of the neural network and is usually associated with long-term memory⁷⁷. If structural plasticity has been observed in the adult brain, similar mechanisms have been found to underpin neuronal development in the early stages of life. In the absence of sensory experience, external stimuli cannot be the cause of synaptic plasticity in the brain. In this case, growth appears to be governed by gradients of guidance molecules⁷⁸ and spontaneous spike activity⁷⁹. The growth of a fiber however is compensated by a decrease in the synaptic weight of another connection. These concepts underlie deeply interesting ideas. First, the neural network is able to self-organize with no external stimuli and no supervised learning. Second, the brain has to deal with a limited quantity of available material, and thus has to optimize its resources. This is achieved via synaptic pruning, a phase during which unsolicited synapses are eliminated. Pruning is associated with learning, but might also be related to energy saving as maintaining

synapses consumes nutrients.

II.3 Bioelectricity

Resting membrane potential

If one were to measure the difference of potential between the inside and the outside of a neuron, putting the positive input of a multimeter into the soma of that cell and its negative lead into the extracellular medium, the voltmeter would show a result of around -70 mV.



Figure 2.11: Resting potential of a neuron.

In other words, at rest, the potential of the intracellular medium should be around 70 mV lower than the potential of the extracellular medium. Why is that, and where is it coming from?

The membrane of a neuronal cell is actually a lipid bilayer impermeable to ions, scattered with highly specific ion channels and pumps that have the ability to open or close. They can therefore drive ions in and out of the cell, thus allowing electrical conduction through the membrane.

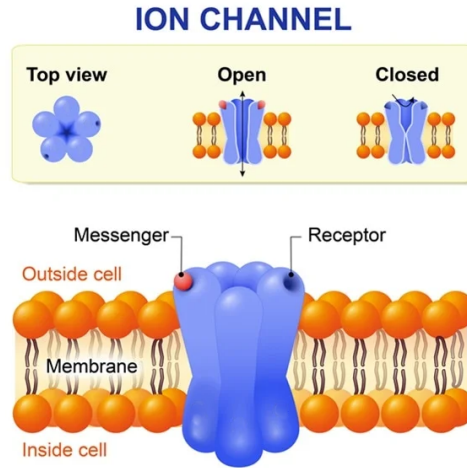


Figure 2.12: Schematic of an ion channel. Adapted from Designua / Shutterstock.

The difference between ion channels and pumps lies in the fact that channels do not require energy to be activated, as they rely on the equilibrium potential to move the ions, whereas pumps actively transport ions through the plasma membrane using the energy provided by ATP (the energy source of the cells) so that they can go against the electrical gradient or gradients of concentration. Indeed, the concentration of the different types of ions present in the medium are different on both sides of the plasma membrane. The following table summarizes these concentrations for the main ions. Note that these values are given for information only and may vary between different individuals or types of cell.

	Intracellular Medium	Extracellular Medium
[Na ⁺]	10 mM	150 mM
[Cl ⁻]	4 mM	115 mM
[K ⁺]	145 mM	4 mM
[Ca ²⁺]	10 ⁻⁴ mM	2 mM

Table 2.1: Ion concentrations in intracellular and extracellular media.

Consequently, ion movement is driven by two main mechanisms: the electric field due to the resting membrane potential and diffusion due to the gradients of concentration.

These ionic movement are used to implement electrical signaling in the brain, in the form of action potentials.

The action potential

An Action Potential (AP), also called a spike, is the propagation of the depolarization of the plasma membrane that originates from the axon hillock of the cell and travels down the axon. Action potentials are “all or nothing” phenomena that have an explicit shape relative to the type of the excitable cell from which they emanate.

They can be compared to the zeros and ones of the digital world. As a result, information is contained not in the shape of action potentials, but in the frequency at which they are fired by a cell.

Figure 2.13 shows the typical shape of such an action potential for a neuron.

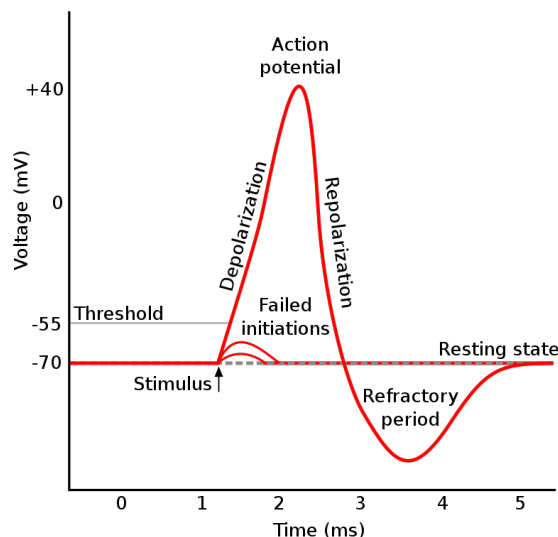


Figure 2.13: Shape of a typical action potential. Reproduced from [80].

An action potential comprises several phases:

1. In order to evoke an action potential, the cell membrane must receive a stimulus important enough to reach the threshold potential, considered to be around -55 mV. If not enough depolarization is accumulated, initiation fails.
2. If the threshold potential is crossed, fast voltage-activated sodium channels open and let Na^+ ions flow into the cell, further depolarizing the membrane.
3. At this point, slow voltage-activated potassium channels start to balance the flow of current into the cell by allowing potassium cations to leave the cell. At the peak of the action potential, sodium ion channels close whereas potassium keeps leaving the intracellular medium, causing repolarization and hyperpolarization.
4. Finally, equilibrium is reestablished. During this refractory period, no other action potential can be evoked as the ion channels are inactivated while the membrane voltage returns to its resting potential.

An action potential is thus a local phenomenon which is able to sustain itself and spread via positive feedback loops: voltage-activated channels control the membrane potential, which in turn controls voltage-activated channels.

When the action potential reaches the end of the axon, it is either converted into a chemical signal mediated by neurotransmitters by a chemical synapse, or passed on directly by an electric synapse, as shown in Figure 2.10, eliciting a stimulus in

the postsynaptic cell. The mechanisms at the origin of the formation of the action potential have been extensively studied by Hodgkin and Huxley in the early 1950s⁸¹, a work that will earn them the Nobel prize in 1963⁸².

Recording the electrical activity of electrogenic cells can be performed using different methodologies⁸³. Intracellular recordings, such as patch-clamp, consist of introducing an electrode within the cytoplasm of a cell in order to record individual action potentials, often at the cost of cellular death. On the other end of the spectrum, the global activity of the brain can be monitored using techniques such as Functional Magnetic Resonance Imaging (fMRI) or Electroencephalography (EEG). These techniques are non-invasive, but have poor spatial resolution as they only allow to record the activity of large populations of neurons (see Figure 2.14). Somewhere in between these two extremes, extracellular recordings allow the non-invasive observation and stimulation of the activity of large neural populations through the detection of Local Field Potentials (LFPs correspond to extracellular field potentials, which are made up of the spatial and temporal integrations of the electrical activity of the surrounding cells^{herrerias'local'2016}). Although the recorded signal might be distorted, it is still possible to theoretically reconstruct the individual sources of the signal, using spike sorting algorithms for example⁸⁴. The following of this section will cover the basics of electrophysiological recordings using Multielectrode Arrays (MEAs).

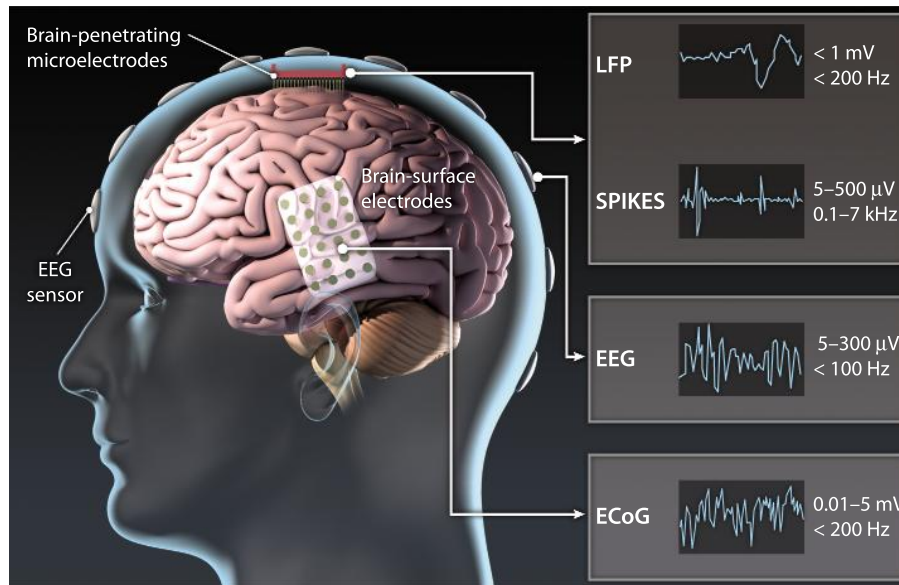


Figure 2.14: Recording the activity of the brain. Different methods exist to record the activity of the brain. Some of them are non-invasive (electroencephalogram, or EEG, recorded from the scalp), while other might be invasive and non-penetrating (Electrocorticogram, or ECoG, from the surface of the brain) or invasive and penetrating (brain-penetrating electrodes). Depending on the technology, different types of signals might be acquired. Reproduced from [83].

Equivalent circuit of the neuron/electrode interface

One important assumption that can be made when studying the neuron/electrode interface is that the system can be divided into two, with the metal electrode acting as an insulator between these subparts:

- the extracellular environment
- the metal/electrolyte interface

The extracellular environment: In a biological environment, the activity of a great diversity of electrogenic cells participates in the disturbance of the extracellular potential and the propagation of voltage deflection in the extracellular fluid. The transmembrane currents that are observed when an action potential is initiated give rise to voltage deflections that can be detected by nearby electrodes. Assuming the homogeneity, isotropy and ohmicity of the extracellular medium (an assumption that might find itself hard to satisfy in the brain⁸⁵) the potential at the surface of electrode in such as environment can be written as

$$V_e = \frac{I}{4\pi\sigma r}$$

where I is the current flowing from a source point, σ is the conductivity of the electrolyte and r is the source-electrode distance. The potential distribution around a point source is shown Figure 2.15.

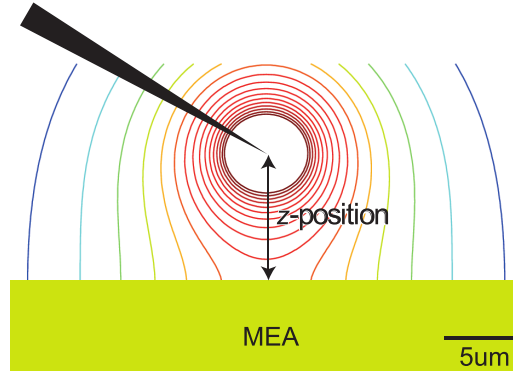


Figure 2.15: Potential distribution around a point source. Adapted from [86].

If there are several point sources to be considered, their contributions have to be summed:

$$V_e = \frac{1}{4\pi\sigma} \sum \frac{I_n}{r_n}$$

If large enough, the electrode cannot be considered as a single point. The potential detected by the electrode can then be computed as the average V_e solved at different locations of the electrode. The distance between the source of current and the electrode and the orientation of the cell might affect the shape and amplitude of the detected potentials, as shown in Figure 2.16.

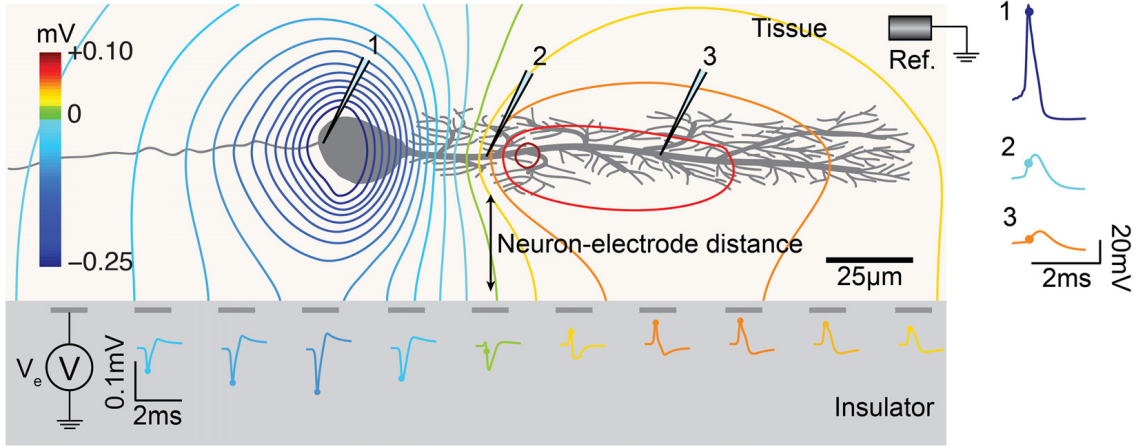


Figure 2.16: The "fluid"-side of the neuron/electrode interface. The orientation of the cell might influence the shape and amplitude of the recorded signal. Adapted from [85].

The properties of the interface then influence the recorded voltage.

The metal-electrolyte interface: The equivalent circuit of the electrode-electrolyte interface is shown in Figure 2.17, where V_e is a low-impedance voltage source, Z'_e is the electrode effective impedance and Z'_a is the effective amplifier input impedance.

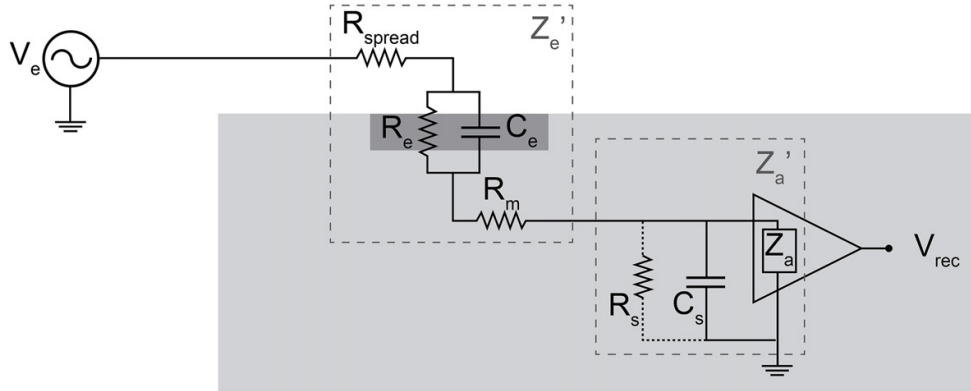


Figure 2.17: The 'metal'-side of the neuron/electrode interface. Adapted from [85].

The impedance at the electrode/electrolyte interface comprises a resistance and capacitance, namely R_e and C_e which correspond to a simplified model of the electric double layer that forms at the interface (see MATERIALS AND METHODS for more details) as well as a spreading resistance R_{spread} , which value can be neglected if the conductivity of the medium is high enough. The resistance of the metallic wires R_m can also be neglected as a consequence of the high conductivity of metals.

On the amplifier side, the effective input impedance Z'_a includes the actual input impedance of the amplifier Z_a in addition to the shunt resistance R_s and shunt capacitance C_s . Due to its low value, R_s can often be neglected. However, C_s , which represents the parasitic coupling of all capacitances of connectors and wires

with the electrolyte through the dielectric insulator, has to be taken into account as its value is critical. Amplifiers are designed to have a high input impedance, so that most of the voltage drop would happen at the amplifier if the ratio $Z'_e/Z'_a \ll 1$. This condition is usually met when using microelectrodes, as the size of the electrodes limits the value of C_e . In fact, this might be one of the reasons why it has been observed that decreasing the impedance of the electrodes does not always correlate with better recordings⁸⁷.

Point-contact model: The full system, including the neuron, electrode and the cleft between them, as well as the amplification circuitry can be modeled using the point-contact model⁸⁸ represented in Figure 2.18.

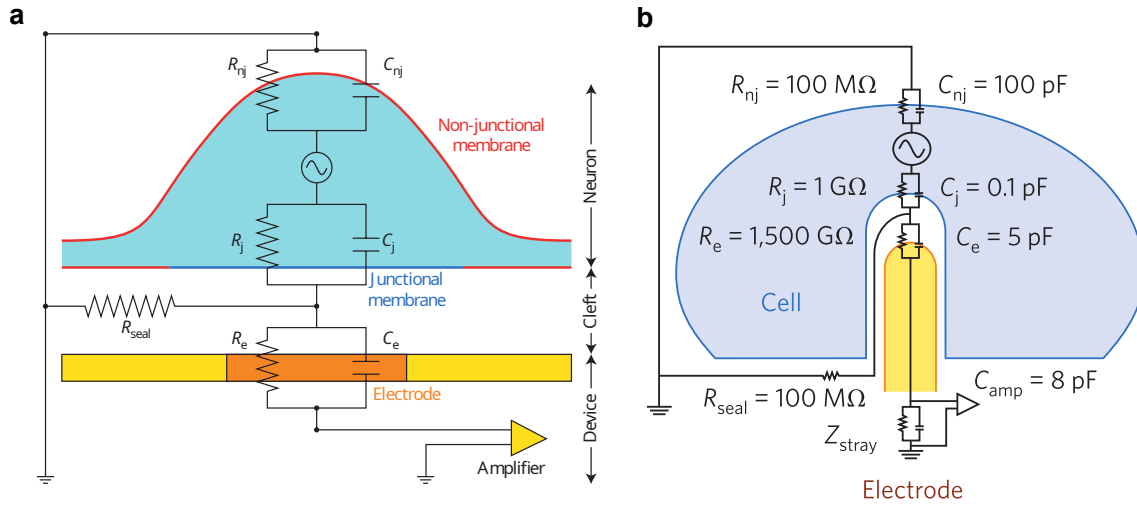


Figure 2.18: Equivalent model of the electrode/neuron interface. a) Point-contact model of the neuron/electrode interface. This representation works both for a passive electrode as well as an active device. b) Order of magnitude of the values of the different parameters of the model. Adapted from [89].

In this model, the neural cell is viewed as a voltage generator, which membrane is subdivided into a junctional and a non-junctional part. The junctional membrane faces the electrode, and is represented by a junctional resistance R_j in parallel with a junctional capacitance C_j . In turn, the non-junctional membrane is represented by a non-junctional resistance R_{nj} in parallel with a non-junctional capacitance C_{nj} . The electrode/electrolyte interface is also represented by a parallel RC, as discussed previously. The important parameter that has been added to this model is the seal resistance R_{seal} , which represents the current leak at the cleft.

This seal resistance is an indicator of the adhesion between the cell and the electrode. The closer the neuron to the surface of the electrode, the lower the magnitude of the leak currents returning to the medium, and therefore the higher the value of R_{seal} . In turn, increasing the seal resistance is synonym with a better coupling between the neuron and the device. Consequently, different design strategies have tried to maximize the value of R_{seal} .

II.4 Electrophysiology with multielectrode arrays

Monitoring the nervous system with a single-cell precision is anticipated to be an important first step to improve our understanding of how the brain works⁹⁰. If techniques such as calcium imaging achieve that level of precision⁹¹, yet temporal resolution does not allow precise spike timing determination. In contrast, direct monitoring of the electrical activity of neural cell has the potential to circumvent this issue using matrices of microelectrodes. The challenge then becomes to increase spatial resolution. Miniaturization of the electrodes and important integration efforts with CMOS technology have resulted in the development of high density probes, with up to 65 536 simultaneously recording and stimulating electrodes⁹².

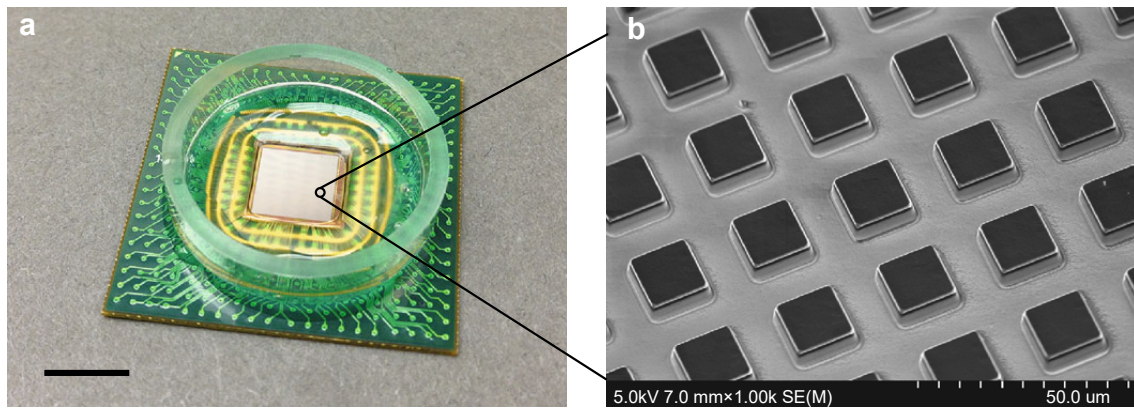


Figure 2.19: A high-density multi-electrode array. a) Photo of the packaged integrated circuit. b) SEM image of the $14 \times 14 \mu\text{m}$ microelectrodes. Adapted from [92].

However, increasing spatial resolution in MEAs in order to get to a single neuron activity induces a decrease of the surface area of the electrodes. This comes at the cost of an increase of the electrode input impedance, and consequently a higher Signal-to-Noise Ratio (SNR) for the recordings. The main strategy implemented to oppose this effect has been to increase the effective surface area using nanostructuration.

Nanostructuration of electrodes

To compensate for the loss of surface in contact with the cell that necessarily comes with the miniaturization of electrodes, 3D nanostructures such as nanotubes^{93–95}, nanowires^{96–99}, or nanopillars^{100,101} have been used. In addition to electrical sensing, these structures hold great potential for biomedical applications, such as intracellular drug delivery¹⁰².

This approach effectively reduces the impedance of the electrodes, although no real improvement of the signal-to-noise ratio have been reported, as the seal resistance remains quite low. These structures typically allow extracellular recordings, but some works reported successful intracellular recording of action potentials as well as synaptic potentials with a MEA^{102,103}. Most of them perform these measurements thanks to the perforation of the neural membrane by the nanostructures, sometimes with the help of electroporation. Nevertheless, mushroom-like electrodes such as the one presented in Figure 2.21 have also been developed to perform intra-

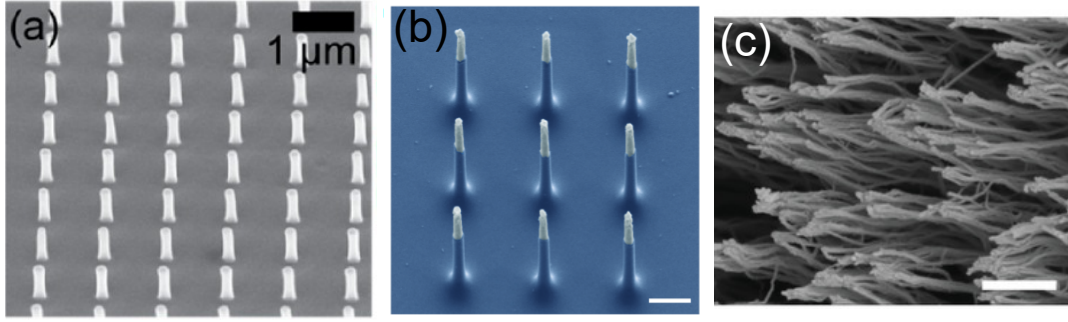


Figure 2.20: 3D nanostructuring of microelectrodes. Examples of a) nanopillars¹⁰¹, b) nanowires⁹⁹ and c) nanotubes⁹⁵.

cellular recordings without damaging the cell¹⁰⁴, this time efficiently improving the cell-electrode coupling by letting the neuron engulf the electrode.

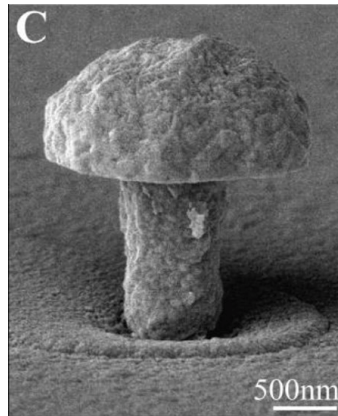


Figure 2.21: Mushroom-shaped microelectrode. Reproduced from [104].

Nanostructuring has also been achieved through the coating of the electrodes with various materials to increase the effective surface area of otherwise relatively small electrodes. To this end, gold nanoflakes¹⁰⁵ or porous materials such as black platinum^{106,107} have been used.

Electrode materials

Apart from metal, other materials have been studied for the development of microelectrodes¹⁰⁸. Due to the ease of integration, high sensitivity and well-developed fabrication methods, inorganic semiconductors and especially silicon-based technologies, often combined with nanostructuring, have been proposed as a material of choice^{109–111}. Indium-tin Oxide (ITO) is another inorganic semiconductor that has attracted some attention, notably due to its optical transparency in the visible spectrum¹¹².

Optical transparency is also to be found in graphene, along with other outstanding properties for neural interface applications, including high conductivity as well as high biocompatibility^{113,114}. More generally, there appears to be a growing interest in carbon-based materials for the development of neural probes. Glassy carbon has for example been investigated because of its inertness, long-term stability and

electrochemical properties^{115–117}. For the same reasons, in addition to a wide potential stability that is very welcome for neural stimulation and functionalization possibilities, diamond has also emerged as a promising electrode material^{118,119}.

Nevertheless, one of the major drawbacks of most previously mentioned materials is their high Young’s modulus, which causes a mismatch with much softer biological tissues, especially for *in vivo* implantation. Additionally, they induce a clear separation at the interface between the electrode and the biological environment due to fundamentally different conduction mechanisms and distinct charge carriers. To address these issues, the field of organic electronics has emerged, offering to bridge the gap between electronics and the biological realm.

III Organic electronics

Conventional electronics is based on inorganic materials, especially metals and semiconductors, among which silicon is to this day the most critical element of the periodic table for the fabrication of integrated circuits. Yet, if semiconductors appear to be irreplaceable in today’s industry, some carbon-based molecules show highly desirable electrical properties. In the 1970s, the work led by Alan J. Heeger, Alan G. MacDiarmid and Hideki Shirakawa proved that plastic material - polymers made of long chains of small molecules called monomers - could be made conductive. For the “discovery and development of conductive polymers”, they were conjointly awarded the Nobel Prize in Chemistry in 2000¹²⁰. Since then, the field of organic electronics has greatly developed, notably with the discovery that conductive polymers could act as a semiconductor material, which paved the way for the creation of organic diodes and organic transistors in the 1980s. Overall, organic semiconductors present many advantageous properties (including their ease of processing, low cost, biocompatibility, stretchability and flexibility...) making them a promising alternative to silicon-based technology.

III.1 Conducting polymers

Structure of conductive polymers

Conducting Polymers (CPs) are long chains of monomer units that have the ability to conduct electrical current thanks to $\pi - \sigma - \pi$ delocalization.

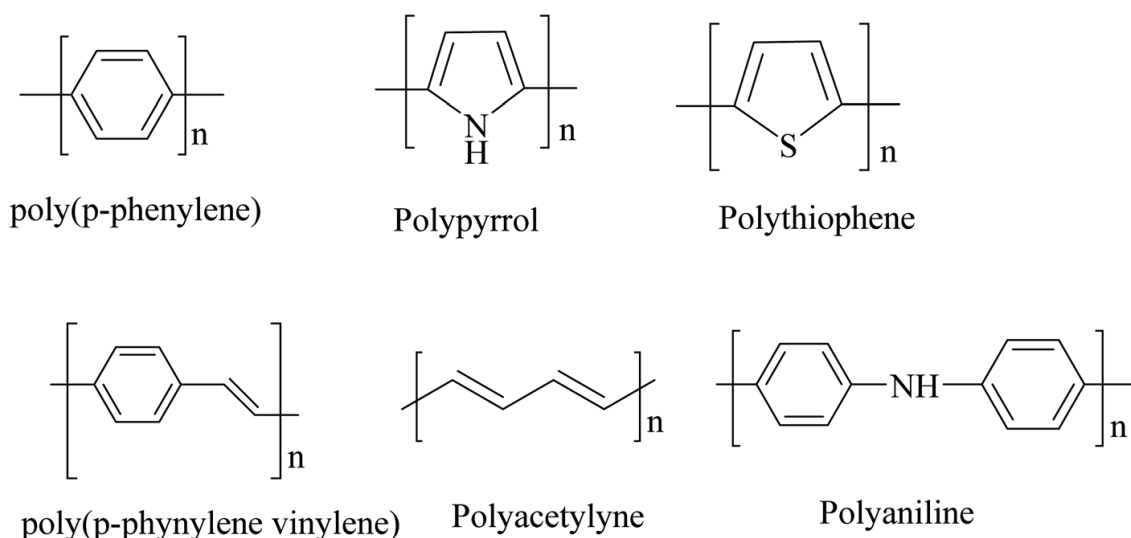


Figure 2.22: Structure of some of the most usual conducting polymers. Reproduced from [121].

Conductive polymers possess a conjugated carbon-based backbone with alternating single and double bonds. Each of these bonds contains a strong σ -bond, while an additional and weaker π -bond is present in each double bond. These π -bonds can be transferred from one double bond to the next single bond, therefore allowing electrons to move along the carbon skeleton of the molecule, delocalizing them over multiples atoms. The delocalized electrons thus created facilitate the conduction of electrical current¹²².

If conjugation provides a way for electrons to move along the backbone of the polymer chain, it is not sufficient to make the material conductive, as these delocalized electrons are not free charges: charge carriers must also be available to support the flow of current. Together with the conjugated bonds of the backbone, they allow charges to move and conduction to happen in CPs¹²³.

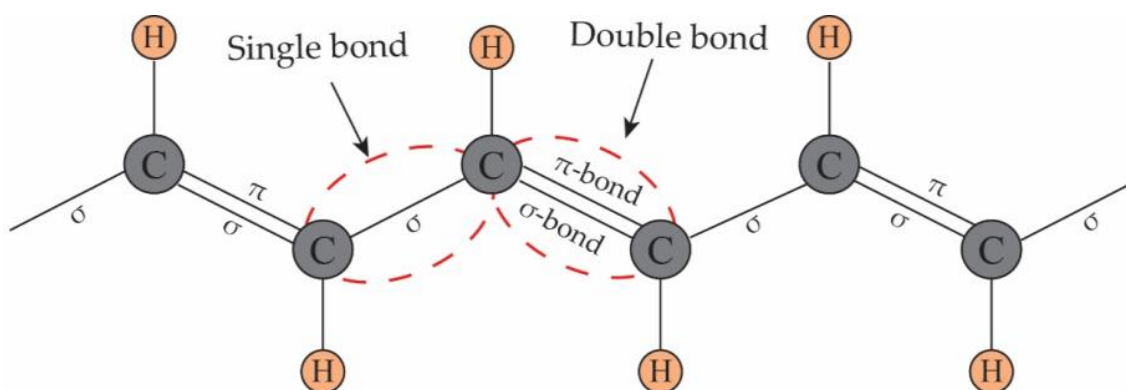


Figure 2.23: Conjugated structure of polyacetylyne, showing the alternating double and single bonds. Reproduced from [122].

Doping in conducting polymers

In their undoped form, the conductivity of conductive polymers can reach that of insulating materials^{122,124}. The process of increasing the conductivity of conductive polymers has been coined doping by analogy with the doping of inorganic semiconductors, but the term might be misleading as the two processes are physically completely different. In CPs, doping is often referred to in the literature as a redox reaction between the polymer and an electron-accepting or electron-donating species^{122,124}, although it can also be achieved without a complete charge transfer¹²⁵. Conversely to what happens in traditional semiconductors, the polymer can undergo partial oxidation in the case of p-doping (therefore adding mobile holes to the semiconductor), or partial reduction for n-doping (in which case mobile electrons are considered to be the charge carriers). In each case, a counterion is required to respect electroneutrality, respectively a counter-anion or a counter-cation. In conductive polymers, doping can be either chemically-, electrochemically- or photo-induced¹²².

Processing

One of the major advantages of organic polymers is their ease of processing. Although, contrary to other conventional polymers, conductive polymers are usually not thermoplastic, they still remain easy to work with. Even when not soluble, most conjugated polymers can be made into a dispersion, enabling solution processing on various substrate materials. Consequently, a variety of well-controlled, low-cost and scalable deposition methods exist that are compatible with conductive polymers, such as spin-coating^{126,127}, dip-coating¹²⁸, inkjet printing^{129,130} or roll-to-roll processes^{126,131}. Polymers can also be processed through additive manufacturing¹³².

Although electroplating is a common electrochemical process used in the industry to produce metal coatings, electrochemical polymerization, also called electropolymerization, is still to this day less conventional in chip manufacturing. Yet, it is an easy-to-perform deposition technique that shows many benefits, as it will be discussed in the following section of this chapter and demonstrated all along this thesis.

III.2 Electropolymerization as an alternative to traditional fabrication routes

Electropolymerization refers to the technique of synthesizing polymers on a conductive surface through the activation of an electric field. It is an interesting alternative to other fabrication methods traditionally employed with organic polymers. When strong enough, an electric field can initiate redox reactions in the presence of electroactive species.

Although this thesis is mainly focused on the applications derived from AC-bipolar electropolymerization, DC electropolymerization remains another important technique for the deposition of polymer thin films. .

DC electropolymerization

Direct-Current electropolymerization is a very straightforward process that requires few materials and equipment, granting it a great advantage over more traditional energy-intensive fabrication methods of the semiconductor industry that often necessitate a lot of resources, including access to cleanroom facilities.

Overall, DC-electropolymerization is a promising route for the design of functional surfaces^{133–136}. The relative ease of processing of conducting polymers allows to chemically tune the properties of the film through material engineering. In our group, Ghazal *et al.*¹³⁷ showed that the addition of a glycolated side-chain to EDOT monomers conferred hydrophilic functions to the resulting electrodeposited thin films used to coat microelectrodes. The film integrating the glycolated EDOT derivative displayed a lower Young’s modulus than usual PEDOT:PSS thin films, thus yielding a better interfacing with neural cultures. In turn, a better interface allows to perform high quality electrophysiological recordings, which translates to a higher degree of precision when trying to localize the cells¹³⁸.

DC electropolymerization has been especially exploited for the development of efficient electrochemical biosensors^{139,140}, but this strategy has also proven useful for the fabrication of gas sensors^{141,142} or solar cells¹⁴³. It has emerged as an alternative to photolithographic patterning for the *in situ* deposition of the channel in organic electrochemical transistors^{144–146}. Additionally, we showed that electropolymerization could be used to tune the properties of electrochemical transistors (especially their capacitance and transconductance) via a control over microdomain organization of PEDOT:PSS films¹²⁷.

AC electropolymerization

While AC electropolymerization is the core deposition method that will be explored in the body of this thesis, it remains nonetheless a niche technique. Yet, it provides advantages over DC electropolymerization. In addition to the formation of a polymer film at low frequency, AC electropolymerization results in the field-directed growth of polymer fibers with a unique morphology when the frequency of the applied waveform is high enough. Polymer deposition is therefore no longer confined to conductive surfaces and their whereabouts, as the spatial reach of polymer fibers becomes more important than that of DC-deposited films. The technique can thus be used to create electrodes with a higher degree of spatial extension, but can also serve to interconnect electrodes over longer distances.

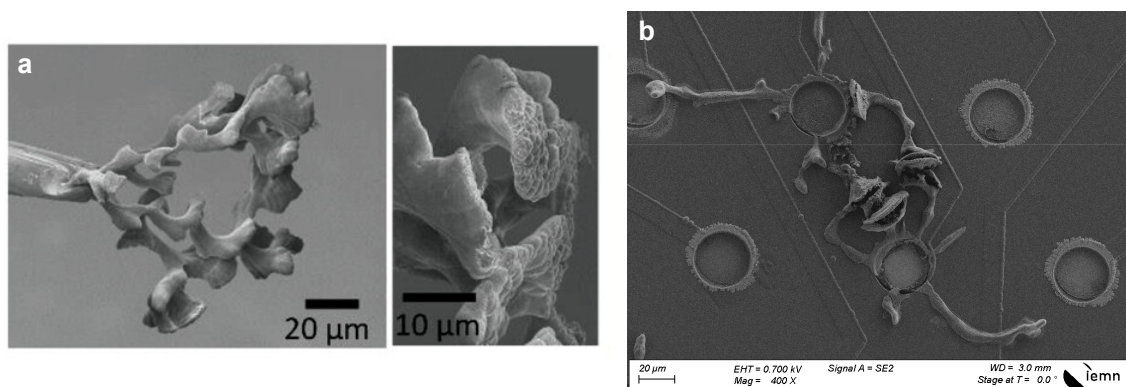


Figure 2.24: SEM image of PEDOT:PSS dendrites, either a) freestanding in a 3D environment (adapted from [147]) or b) on a 2D substrate.

Another benefit in favor of AC electropolymerization is that the polymer fibers thus grown display a transistor-like behavior, showing a modulation effect in their output current as an external potential is applied to the system. Especially, Janzakova *et al.*¹⁴⁷ demonstrated how powerful of a tool AC electropolymerization was regarding the development of multielectrode organic electrochemical transistors (see The Organic Electrochemical Transistor), with a clear structure-properties relationship arising in these emerging materials and devices (see Figure 2.25).

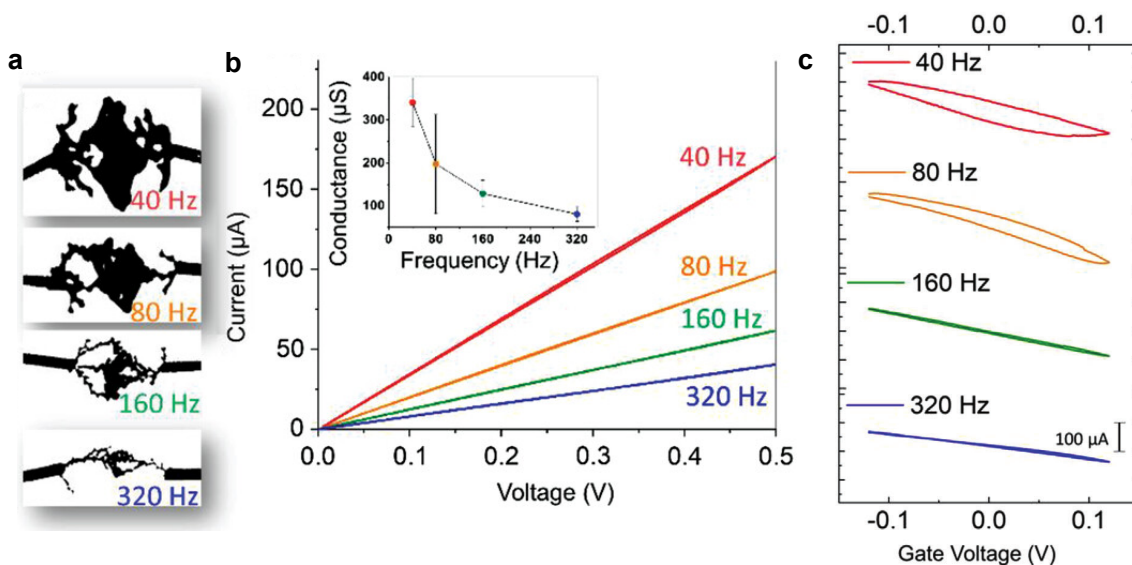


Figure 2.25: Dendritic PEDOT:PSS organic electrochemical transistors. a) Different growth frequencies result in different morphologies. **b)** Output characteristics and **c)** transfer characteristics of the dendritic OECTs. Adapted from [147].

Using AC electropolymerization (that will be referred to simply as electropolymerization from now on, unless specified otherwise) increases the degree of freedom introduced during polymer growth. Indeed, many different parameters appear to influence the morphology of the fibers such as the chemistry of the monomer and its concentration, the choice of supporting electrolyte and solvent, or the substrate on which growth happens^{148–150}. In our team, Janzakova *et al.*¹⁵¹ demonstrated the

primary importance of waveform parameters such as the applied frequency, duty factor and voltage amplitude. Indeed, there exists an undeniable relationship between electrical parameters and the morphology of the fibers, as illustrated in Figure 2.26.

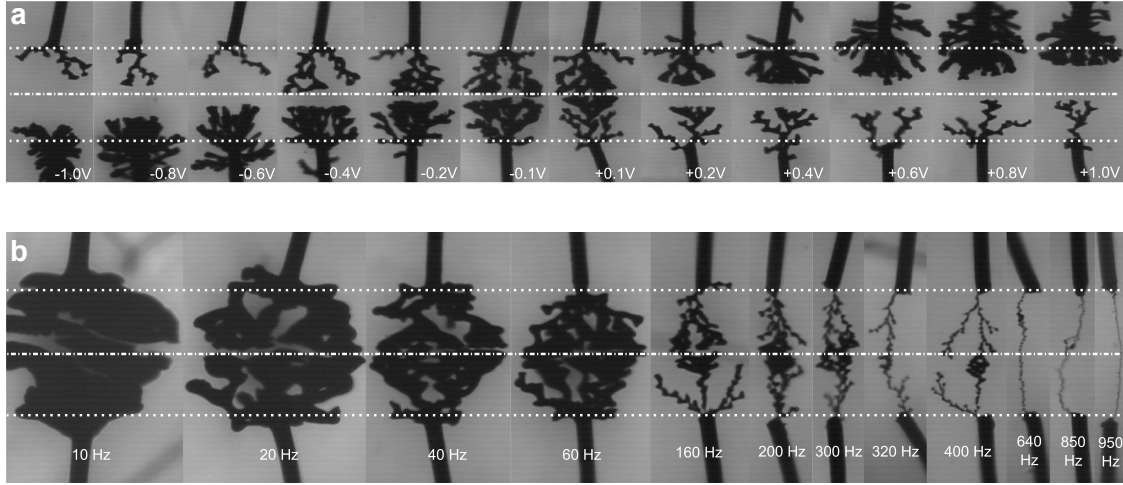


Figure 2.26: Influence of some of the waveform parameters on the morphology of electrogenerated polymer dendrites. a) Influence of the offset voltage. b) Influence of the growth frequency. Adapted from [151].

As the electropolymerization process is field-dependent, the shape of the fibers also depends on the distance between the working and counter electrodes¹⁵². It appears that it is the accumulation of counter-anions at the working electrode that drives the growth along the electric field lines. In the case of DC electropolymerization, growth therefore happens isotropically and results in the deposition of a film on the working electrode of the system, whereas under AC electropolymerization the intensity of the electric field is greater at the tip of the fibers and results in directed growth. As the oxidation potential of many conductive polymers are of the order of one volt, low voltage supplies (compatible with 5V logic) are perfectly suited for the job^{122,137,153}. In addition, electropolymerization has the potential to be scaled. Although a model for the growth of dendritic networks has been proposed by Ciccone *et al.*¹⁵⁴, electropolymerization remains overall a very stochastic process, a property that will be explored in chapter 6.

To the best of our knowledge, the first mention of conducting polymer nanowires grown between in-plane electrodes through AC electropolymerization in the literature dates back to a conference proceeding published in 2014 by Dang *et al.*¹⁵⁵. Yet, the first article to be found on the formation of PEDOT fibers via electropolymerization seems to have been published only two years later by the team of Shinsuke Inagi¹⁴⁸. Although the experimental setup is not conventional (PEDOT is grown between two conductive pieces of metal immersed into an electrolyte through the application of an external electric field), this paper paved the way for the study of the formation of polymer fibers^{149,150,156}, also called dendrites in analogy with their biological counterpart.

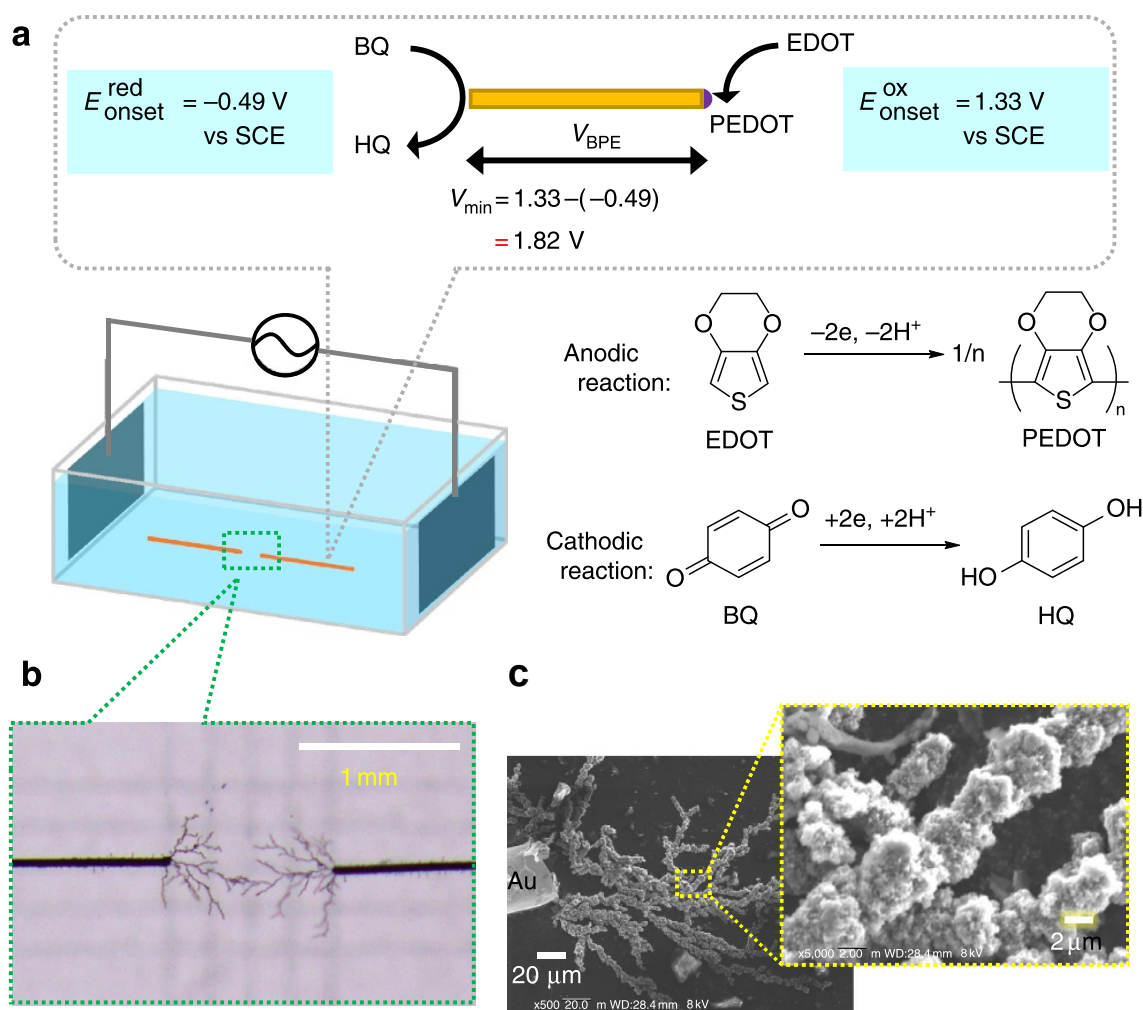


Figure 2.27: Experimental setup for the AC-bipolar electropolymerization of EDOT. **a)** When a strong enough external electric field is applied, EDOT is oxidized into PEDOT at the anode, while benzoquinone is reduced into hydroquinone as a sacrificial reaction at the cathode. **b)** Microscopic image of the PEDOT fibers. **c)** SEM image of the PEDOT fibers. Reproduced from [148].

III.3 Organic Mixed Ionic Electronic Conductors (OMIECs)

There are many CPs available^{121,157}, but recently an emerging class of organic material, able to efficiently transport and couple ionic and electronic charges, has attracted a lot of attention: the Organic Mixed Ionic-Electronic Conductors (OMIECs).

As the name suggests, OMIECs are a class of conducting polymers that possesses ion conducting properties in addition to electronic transport. These materials present the unusual ability to operate in an electrolytic environment, from where ions can be collected as a way to maintain charge neutrality when charges are injected into the polymer, which provides OMIECs with an extensive ionic-electronic coupling. These properties have made OMIECs very attractive for many domains of applications, including energy storage^{158,159}, supercapacitors¹⁶⁰, light-emitting devices¹⁶¹, bioelectronics^{162–165} or neuromorphic computing (see section III.5) and work as ion sensors^{166,167}.

There exists a great diversity of OMIECs, along with an even greater number of material properties that has helped the field to develop, as reviewed by Paulsen and coworkers¹⁶⁸. However, poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate), better known as PEDOT:PSS, might well be the most prototypical OMIEC representative.

The workhorse of OMIECs: PEDOT:PSS

PEDOT:PSS has been praised for its numerous advantageous electronic, optical as well as mechanical properties. It is indeed a good electrical conductor (with a conductivity in the range of 10^{-4} - 10^3 S.cm⁻¹), visually transparent in the visible spectrum and flexible^{169,170}. Moreover, it has a good physical as well as chemical stability in ambient conditions¹⁷¹. Its biocompatibility, low electrochemical impedance and ionic-electronic coupling make PEDOT:PSS a material of choice for the fabrication of bioelectronic devices^{162,164}, but it has also been employed in a variety of applications, including energy storage¹⁵⁹, electrochromic devices¹⁷², transistors¹⁷³, solar cells¹⁷⁴ or OLEDs¹⁶¹, among many others. Another non negligible advantage of PEDOT:PSS is that it is commercially available at a low cost (for example from Heraeus CleviosTM or Agfa OrgaconTM).

Chemical structure: PEDOT:PSS is an heterogeneous blend of an electronically conducting conjugated polymer (PEDOT) with an ion-conducting polyelectrolyte (PSS). As PEDOT does not solubilize well by itself, it has to be stabilized using a template to create aqueous dispersions (commonly PSS, although other polyelectrolytes have been used). In addition, PSS acts as a dopant for the hole conducting PEDOT, its SO₃⁻ sulfonate anions balancing the positively charged PEDOT chain, as shown in Figure 2.28.

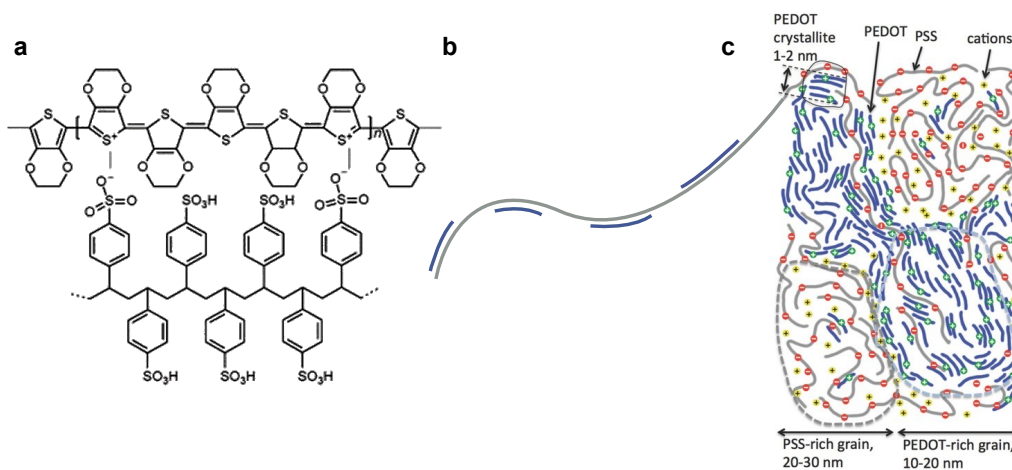


Figure 2.28: Chemical structure of PEDOT:PSS. a) Oxidized PEDOT molecules and sulfonate groups from the PSS are bound through Coulomb interactions¹⁷⁵. b) PSS backbone with PEDOT segments, adapted from [176]. c) Schematic diagram of the morphology of a PEDOT:PSS film, demonstrating a two-phase system with PEDOT-rich particles embedded in a PSS-rich matrix¹⁷⁷.

When formed into thin films, PEDOT:PSS presents a granular structure, with conducting and hydrophobic PEDOT-rich domains supporting hole conductivity, connected in a hydrophilic and insulating PSS-rich matrix that allows ions to move within the bulk of the material^{176,177}.

Conduction mechanism: The PEDOT-to-PSS content is an important parameter that influences the conductivity of the polymer, with a theoretical optimum found at one dopant per three monomer units. The commercially available Clevious™ PH1000 has a PSS content 2.5 times superior to that of PEDOT. Yet, the doping level is not the only important factor when it comes to improving conductivity. Simply put, in the presence of a local electric field E (V.m^{-1}) in a conductive material, Ohm's law states that:

$$\vec{J} = \sigma \vec{E}$$

with J the current density (expressed in A.m^{-2}) and σ the total conductivity (expressed in S.m^{-1}).

As discussed previously, the number of charge carriers available in the material is directly proportional to the doping level, but the conductivity also depends strongly on the mobility of these carriers. In OMIECs, this parameter tends to be the one limiting the overall movement of the charge carriers. In the case of PEDOT:PSS, the disordered organization of the polymer makes it impossible for electrons to be delocalized over extended atomic chains (see Figure 2.28c). As a consequence of this morphology, the main obstacle for the flow of electronic current is to pass from one PEDOT-rich grain, with higher intrinsic conductivity, to the next. Charge mobility is therefore limited by the movement of holes from one PEDOT chain to another through PSS-rich boundaries (inter-chain mobility, as opposed to intra-chain mobility). Apart from primary doping, the conductivity of PEDOT:PSS can be further enhanced through secondary doping methods, and various physical and chemical approaches^{178–180}. In fact, these treatments mostly aim at increasing the degree of order in the polymer and reorganizing the PEDOT chains in order to favor intra-chain mobility, thus enhancing hole conductivity often at the expense of ion mobility. Post-fabrication acid treatment has for instance been demonstrated to lower the quantity of PSS in the polymer films, resulting in a rearrangement of the three-dimensional network and an increase in conductivity. The most common recipe to prepare PEDOT:PSS dispersions include ethylene glycol to drive phase separation between PEDOT and PSS, and (3glycidylxypropyl)trimethoxysilane (GOPS) to function as a crosslinker to increase stability¹⁸¹.

Due to its ability to solvate ions in the whole bulk of the material, PEDOT:PSS possesses a volumetric capacitance, a feature that cannot be easily obtained with classes of materials other than OMIECs and that has proved essential for the development of a new class of transistors: the Organic Electrochemical Transistor, or OECT.

III.4 The Organic Electrochemical Transistor

As mentioned earlier, the properties of OMIECs (and especially their biocompatibility and high iono-electronic coupling) have made this class of materials very

attractive for bioelectronics. In particular, the OECT has shown great promise for the development of the field, because of a number of advantages that the device possesses over traditional electrodes, notably its high voltage to current transduction that allows high quality recordings through local amplification.

Operation principle

The basic structure of an OECT is shown Figure 2.29. Just like every other transistor, the OECT is a three-terminal device, with a source, drain and gate electrode. The organic semiconductor film that forms the channel of the transistor is immersed in a liquid electrolyte. The conductivity of the channel is controlled by the potential applied at the gate, referenced with respect to the source potential. This voltage controls the injection of ions into the channel of the device, which results in electrochemical doping/dedoping of the polymer.

Although many materials have been investigated^{173,182–184}, PEDOT:PSS may be the most typical material for the fabrication of organic electrochemical transistors. Because PEDOT can be highly doped, the resulting device is often operated in depletion mode, as presented in Figure 2.29b. PEDOT:PSS is indeed intrinsically doped, meaning that, in the absence of gate potential, the current flowing through the channel will be proportional to the number of charge carriers in the channel. However, when a positive voltage is applied at the gate, cations are injected into the PEDOT:PSS channel and dedope the polymer as the holes are extracted at the source, leading to a decrease in conductivity and therefore a drop in the drain current.

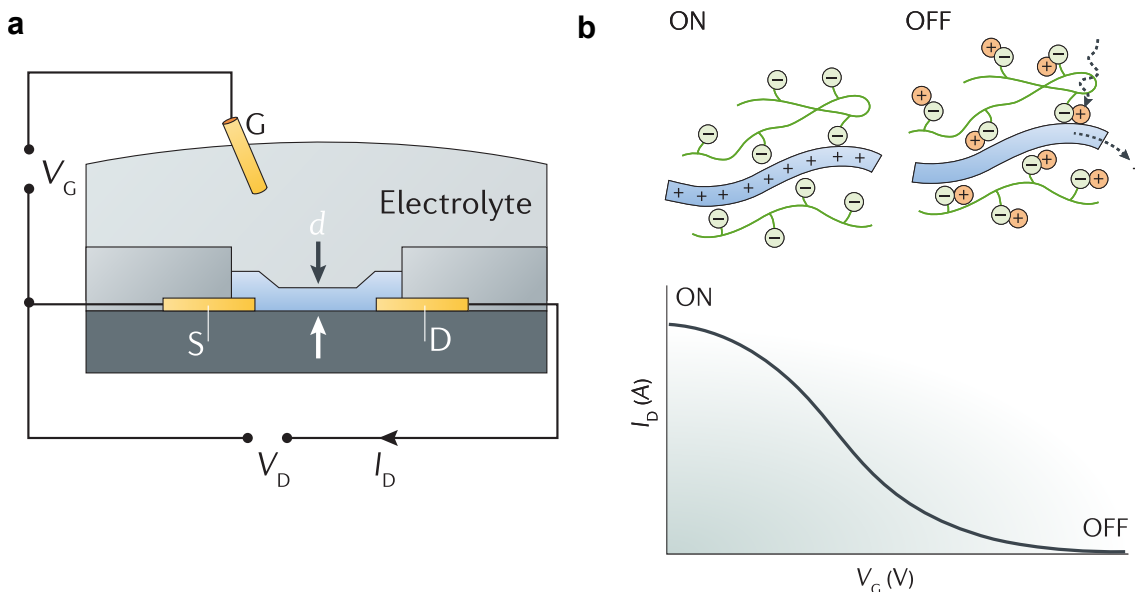


Figure 2.29: The OECT. a) Structure of the transistor, showing the source (S), drain (D) and gate (G) electrodes along with the polymer channel (in blue) and the electrolyte. b) Operation principle of a depletion-mode transistor. When a positive gate potential is applied, cations drift into the bulk of the channel, dedoping the polymer. Adapted from [173].

What makes the OECT different from other classes of transistors is the volumetric capacitance of OMIECs, which yields the material with a high iono-electronic

coupling. As a result, small variations of the gate potential lead to large modulations in the drain current (see Figure 2.31). This confers OEETs a great advantage over other types of transistors, such as the field-effect transistor (FET) or the electrolyte-gated FET, in which the dedoping effect is limited to a thin interfacial region.

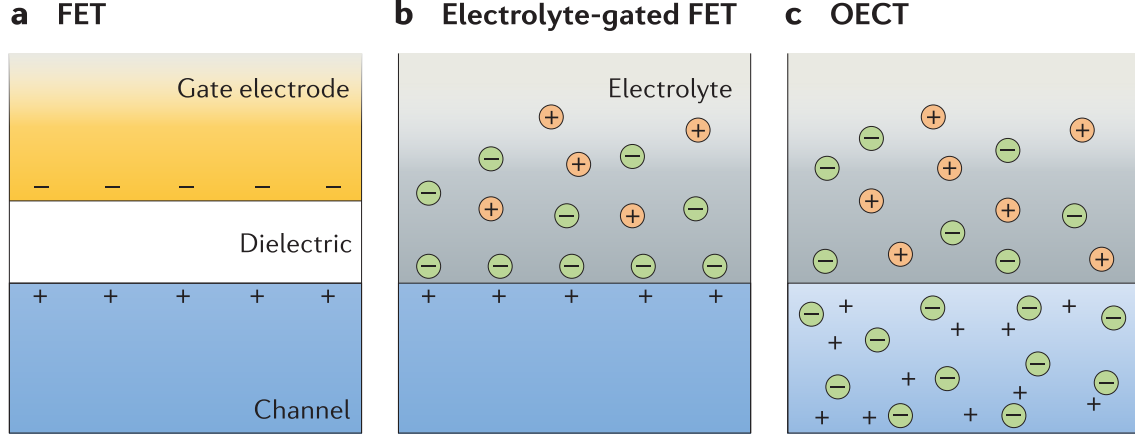


Figure 2.30: Gating effect in a) field-effect transistor, b) electrolyte-gated transistor and c) OEET. Reproduced from [173].

Therefore, the most important figure of merit of the OEET is its transconductance, which is a measure of the modulation of the drain current in response to a variation of the gate potential. The transconductance g_m is defined as follows¹⁸⁵:

$$g_m = \frac{\partial I_D}{\partial V_G} = \frac{W}{L} d\mu C^* (V_{TH} - V_G)$$

Where W , L and d are the width, length and thickness of the polymer channel, μ is the charge carrier mobility, C^* the volumetric capacitance and V_{TH} represents the threshold voltage.

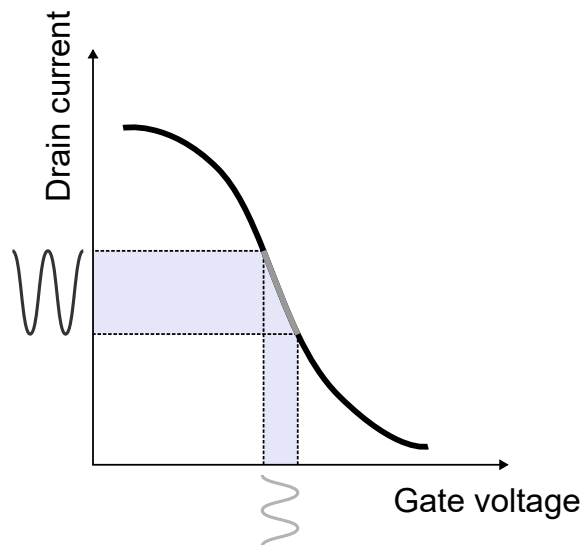


Figure 2.31: Local amplification of the OEET. Due to its high transconductance, the OEET is an efficient signal transducer, as a small perturbation of its gate voltage translates to a larger perturbation of the output current.

As any other transistor, the OECT can be used either for digital or analog electronics, in which case it makes for an efficient amplifier due to its high transconductance.

Modeling the OECT

Bernards and Malliaras were the first to propose a simple model to reproduce the steady-state and transient behavior of the OECT. In their seminal paper¹⁸⁶, they segregated the model into an electronic circuit to account for hole transportation in the semiconductor channel of the transistor, and an ionic circuit that describes the behavior of ionic charge carriers in the electrolyte. According to this model, the electronic circuit consists in a resistor. As a result, the current in the channel can be described by solving Ohm's law.

The ionic circuit of this model is made of a resistance in series with a capacitance, to represent the resistance of the electrolyte and the electrical double layer that forms at the interface with the semiconductor channel, but also takes into account the gate-electrolyte interface.

The Bernards model fits rather well to the output characteristics of OECTs and provides a good starting point to understand the device physics. Yet, this model can be further improved. It considers a purely capacitive coupling between the electrolyte and the bulk of the polymer, without taking into account charge transfer and works only in a non-Faradaic regime. It also considers that the injection of anions into the bulk of the polymer does not lead to increased charge carrier concentration. Some models have thus tried to expand the Bernards model to improve the understanding of the OECT operation¹⁸⁷, including non-uniform charge mobility and a charge-density dependent conductivity¹⁸⁸. However, in all of these models, ionic motion is limited to one dimension, vertically between the gate and the channel. Kaphle *et al.*¹⁸⁹ therefore used a 2D drift-diffusion model to take into account lateral ions current and show that ion accumulation at the drain electrode alters the transistor steady-state behavior. But the operation behind the OECT cannot be fully understood without looking at the chemistry of the system. To this end, Tybrandt *et al.*¹⁹⁰ proposed a two-phase model, based on the assumption that the volumetric capacitance of PEDOT:PSS arises from the electrical double layer formed at the interface between the PEDOT and the PSS phase¹⁷⁷. More recently, Cucchi *et al.*¹⁹¹ offered a unified framework using a thermodynamic approach.

Applications of the OECT

The OECT has emerged as a building block of organic bioelectronics due to its enhanced performances in terms of biocompatibility, stability in aqueous environment and ability to amplify efficiently biological signals through local amplification due to the large transconductance of some CPs such as PEDOT:PSS^{163,173,192}.

Malliaras was one of the pioneers in the development of the field, as his work established the advantage of the OECT over other devices in electrophysiological recordings, notably due to the absence of a dielectric at the polymer/electrolyte interface that allow the response of the system to be determined by the whole volume

of the channel rather than by a surface effect¹⁹³. In 2013, his team was the first to report on the *in vivo* use of an OECT to record brain activity¹⁹⁴, observing signal that was previously hard to detect with passive electrodes. It was also shown that such a device was successful in recording the electrocardiogram, electrooculogram and brain activity in humans¹⁹⁵. The OECT therefore acts as a high fidelity transducer for biological signals¹⁹². An effort toward integration later led Khodagholy and coworkers to develop internal ion-gated OECTs to allow individualized transistor gating^{196,197}.

If the OECT indeed demonstrated great potential for electrophysiological recordings when interfaced with electrogenic cells^{194,197,198}, it has also shown very useful for a wide variety of applications. Fabricated on a flexible polymer substrate, an OECT array was for example used to record electrocardiogram when placed on the surface of the human skin¹⁹⁹. OECT-based circuits have also been developed, such as inverters for biosignal amplification²⁰⁰ based on an ambipolar OMIEC. Through impedimetric measurements, the OECT has served as a diagnostic tool for biological barrier layers²⁰¹, or the attachment of a cell to a substrate²⁰². Another major benefit of organic polymers is that they can easily be functionalized to tune their properties according to the tasks for which there are intended. As such, OECTs can thus perform signal transduction for the detection of analytes and metabolites²⁰³ such as glucose²⁰⁴, lactate^{205–207}, dopamine^{208,209} or even multi-analyte screening from saliva¹⁶⁷ and work as ion sensors^{166,210}.

As seen with all these examples, the OECT is a versatile device, which holds great promise for electrophysiological and biosensing applications. But apart from bioelectronics, the OECT, and more generally conductive polymers, have contributed to the recent development of a new class of organic neuromorphic devices, as it will be explored in the following section of this manuscript.

III.5 Neuromorphic organic devices

A highly sought-after property for the realization of neuromorphic devices is the ability to implement adaptative behavior as a function of past activity, in order to mimic short- or long-term memory. In this direction, the work of Gkoupidenis really laid the foundation of the neuromorphic organic community. The first mention of neuromorphic functions in a PEDOT:PSS OECT was made by Gkoupidenis *et al.*²¹¹, who demonstrated synaptic plasticity functions, such as paired-pulse depression, adaptation and dynamic filtering (Figure 2.32).

Although these functions implement short-term plasticity, the memory of the device was made non-volatile by using another PEDOT derivative undergoing conformational change under high potential²¹². Building on this work, long-term plasticity was displayed in an OECT²¹³, leaving behind the von Neumann framework.

These results paved the way for the development of organic-based neuromorphic circuits and more advanced neuromorphic functions. Using multi-gated OECT, Gkoupidenis *et al.*²¹⁴ demonstrated orientation-selectivity in the fashion of the mammalian visual system. In 2017, van de Burgt *et al.*²¹⁵ demonstrated a non-volatile

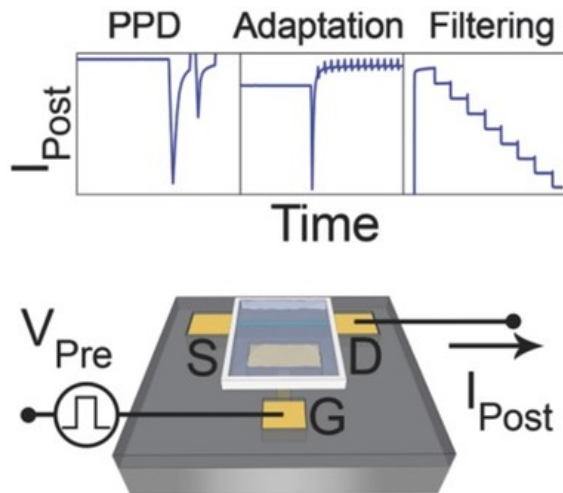


Figure 2.32: Short-term plasticity functions in an OEET. Reproduced from the graphical abstract of [211].

electrochemical neuromorphic organic device with over 500 programmable conductance states in a narrow voltage range (about 1 V), which behaves as an artificial synapse with low-energy consumption. Other artificial synapses later came up²¹⁶, showing for example neurotransmitter-mediated plasticity²¹⁷. The first organic electrochemical neuron with ion-modulated spiking was then developed by Harikesh *et al.*²¹⁸ Recently, Gkoupidenis *et al.*²¹⁹ proposed a biohybrid interface with an organic artificial spiking neuron, while Gerasimov *et al.*²²⁰ produced the first example of a biological interface to include a hardware-based pattern classifier.

Electropolymerization has started to emerge as an alternative fabrication method, offering new possibilities for the devices. Long-term plasticity was for example achieved by increasing the conductance of an OEET by using electropolymerization as a post-fabrication tool to create evolvable devices¹⁴⁶, or by mimicking dendritic morphogenesis through the growth of PEDOT:PSS dendritic interconnections¹⁵¹. Using AC electropolymerization, Janzakova *et al.*²²¹ implemented structural plasticity by the growth of PEDOT:PSS fibers, showing that this strategy can follow Hebbian learning principles. The same idea allowed Hagiwara *et al.*²²² to develop networks of conductive polymer fibers into a 3D space. In terms of computing, the variability induced by the electropolymerization process allowed Pecqueur *et al.*¹⁴⁴ to generate structurally different active materials on lithographically patterned devices in order to enrich the dynamics of a reservoir, thus facilitating the classification of dynamic information. The same concept later inspired Cucchi *et al.*²²³ to grow a network of polymer fibers to perform classification through a reservoir computing framework, although Akai-Kasaya *et al.*²²⁴ had previously demonstrated the use of such dendritic interconnects for the physical implementation of machine learning.

IV Conclusion

Throughout this introductory chapter, we saw that the current computing paradigm and architecture face challenges that require rethinking our conception of computing. In that regard, the brain appears to be a fascinating source of inspiration, as it

is able to perform parallel information processing at a very low energy cost. Even more intriguing is that it performs such a feat with materials that do not seem fitted at all for the job: wetware that depends on slow ionic movement to transfer and process information in a salty aqueous solution. A new class of materials, OMIECs, paves the way for the implementation of a new type of brain-inspired computing hardware, mainly due to their iono-electronic coupling and biocompatibility. In addition, these materials allow the development of new fabrication methods, such as electropolymerization that gives birth to novel devices with a unique morphology that arises from the stochastic nature of the process.

As these emerging devices and materials bring about fascinating and exciting opportunities, they remain relatively new concepts that require further investigation to assess their practical potential. This thesis will notably aim at answering the following questions:

- How does the electrochemical environment alter fiber growth? As we saw that electrical parameters exert an important influence on the morphology of the fibers, how does that translate to the composition of the surrounding electrolyte? What are the ideal and boundary conditions for the formation of polymer fibers, and what is the impact of the environment over their properties and operation? These questions are of primary importance, as a better understanding of the relationship between electrolyte composition and growth could help tuning the properties of polymer fibers by engineering their environment. In addition, depending on the use case the chemical composition of the environment might be imposed (e.g., when in contact with living tissue). A fine understanding of the parameters controlling polymer deposition therefore becomes essential.
- What are the computational possibilities offered by such devices? Since OMIEC materials have been demonstrated to implement various neuromorphic functions, can we expect the same behavior from polymer dendrites? Does the unique and complex morphology conferred by the electropolymerization framework to these objects results in the emergence of functions that could not be implemented following more traditional fabrication methods? Drawing inspiration from the operation of the brain, we will explore the abilities of polymer dendrites from an unconventional computing point of view.
- Can these polymer fibers lay the foundation for the next generation of brain machine interfaces? In this first chapter, we have explored how conducting polymers have been praised for their biocompatibility and for the realization of efficient interfaces with electrogenic cells. Using polymer fiber networks, would it be possible to create a mesh of interconnected electrodes with greater spatial extension and *in memory* computing abilities? This approach could enable *in materio* computing directly at the junction with biological systems. Do artificial dendrites enhance signal detection compared to usual metal or polymer-coated electrodes? These innovations could mark a first step toward autonomous interfaces with *in operando* evolving abilities and advanced *in materio* signal processing.



Figure 2.33: Artificial polymer dendrites and biological neural networks merging together. In the future, polymer fiber networks could serve as the basis of the next generation of brain/machine interfaces, integrating seamlessly with the cells of the nervous system and forming efficient bidirectional interfaces with *in memory* computing abilities. This image was generated with Adobe Firefly.

In the following chapters of this manuscript, we will explore electropolymerization as a powerful tool for the development of brain-inspired and biologically interfaced hardware. Chapter MATERIALS AND METHODS will provide details about the materials and methods of importance for this thesis. In chapter EXPLORING THE ELECTROCHEMICAL ENVIRONMENT, the electrochemical conditions for the growth of polymer fibers and their repercussions on the morphology and electrical properties of the dendrites will be explored. In chapter FUNCTIONAL PLASTICITY IN POLYMER DENDRITES, I will make use of these objects to perform signal detection and classification. Chapter DENDRITIC NETWORKS FOR MORPHOLOGY-DEPENDENT COMPUTING HARDWARE will take advantage of the intrinsic stochasticity of electropolymerization to demonstrate brain-inspired conductive polymer dendrites for morphology-dependent computing hardware. In chapter DENDRITIC ELECTRODES FOR IN VITRO ELECTROPHYSIOLOGY, polymer dendrites networks will be investigated as a biohybrid interface. Finally, chapter DISCUSSION AND OUTLOOK will offer some future perspectives on this work.

3

MATERIALS AND METHODS

This chapter will provide details and descriptions regarding the main experimental methods, materials and equipment that were used in the context of this thesis.

I Electrochemical Impedance Spectroscopy

Electrochemical Impedance Spectroscopy (EIS) provides insights on the dynamics of an electrochemical system by measuring the impedance of a system, or the opposition to the flow of current when an alternating voltage is applied to its terminals. During EIS measurement, a small sinusoidal perturbation is applied over a large range of frequencies to a system in equilibrium, while the output current is recorded by an analyzer. EIS thus allows to study the transfer function of an electrochemical system, which is to say the deformation of the input in terms of amplitude and phase shift. For the analysis to be correct, the system must be linear (a sinusoid input has to give rise to a sinusoid output). As a large range of frequencies are swept, EIS highlights processes that have different timescales, from very slow to very fast.

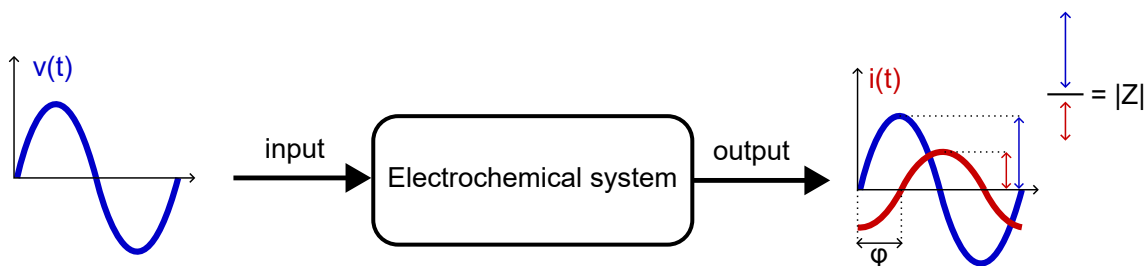


Figure 3.1: Electrochemical Impedance Spectroscopy, showing a small voltage sinusoidal (blue) as the input and the corresponding output current (red), with φ the phase shift and $|Z|$ the impedance modulus.

I.1 The electrochemical cell

The usual 3-electrode electrochemical cell, with the reference, counter and working electrodes is shown in Figure 3.2. In this configuration, the reaction to be monitored takes place on the surface of the working electrode, and a potentiostat measures the potential difference between the working and the reference electrode, that thus have

to be placed as close as possible to each other to minimize potential drop in the electrolytic solution. The value of this potential difference is corrected by flowing a current between the counter and the working electrodes.

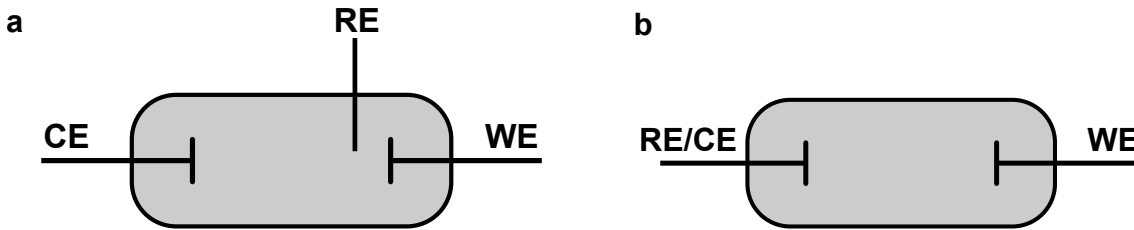


Figure 3.2: The electrochemical cell, showing the working (WE), counter (CE) and reference electrodes (RE). **a)** Usual three-electrode setup. **b)** Two-electrode setup where the reference and counter electrodes are connected. Adapted from [225].

In the case of a 2-electrode setup, such as the one that will be used in this thesis, the reference electrode and counter electrode are connected, limiting the precision of the potential applied between the working and the counter electrodes. Therefore, the impedance of the cell is due to:

1. The interface between the counter electrode and the electrolytic solution.
2. The electrolytic solution.
3. The interface between the working electrode and the electrolytic solution.

These different parts can be electrically modeled.

The electrolytic solution

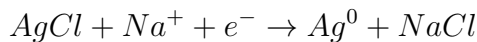
An electrolytic solution is a liquid medium containing electrolytes. Due to the presence of these mobile ions, the electrolyte is a conductive medium, which conductivity depends on ionic concentration and ionic mobility. Consequently, an electrolyte can be electrically represented as an ohmic resistance $R_{electrolyte}$.

The electrode-electrolyte interface

When a potential difference is applied between two electrodes immersed in an electrolyte, an ionic drift is induced in the solution, as the cations move towards the negative electrode and the anions towards the positive electrode. Two phenomena can then happen at the electrode-electrolyte interface:

1. In the absence of redox reaction, current cannot flow through the interface and ions accumulate at the electrode interface, resulting in a layer of charges on one side of the interface and a layer of opposite charges on the other side. This electrical double layer (EDL) causes the voltage to drop dramatically at the interface. Inert materials, such as gold, are good representatives of these ideal polarizable electrodes. Electrically speaking, the EDL is represented as a capacitance C_{DL} , with an impedance $Z_C = \frac{1}{jC\omega}$ where j is the imaginary unit and ω the pulsation related to the frequency f such that $\omega = 2\pi f$.

2. In the presence of redox couples, charge-transfer reactions can happen, resulting in a faradaic current flowing through the interface. The electrode-electrolyte interface is then best represented as short circuit. As a consequence, most of the voltage drop in the system is due to the electrolytic resistance. Ag/AgCl electrodes immersed in NaCl solution are a good example of ideal non-polarizable electrodes, given that:



However, the flow of current is always limited at the interface of the electrode by either one of two processes, namely the kinetics of the redox reactions or the diffusion of the redox species to the electrode surface. The interface is therefore represented as a charge transfer resistance R_{CT} in series with a Warburg element Z_W , such that $Z_F = R_{CT} + Z_W$.

In real conditions, electrodes can exhibit behaviors of both ideal polarizable and non-polarizable electrodes.

I.2 Electrical model of the electrochemical cell

Taking into account two metal-electrolyte interfaces and the resistance of the electrolytic solution as previously described, the electrical model of the electrochemical cell becomes as represented in Figure 3.3. Yet, the model can still be simplified. Indeed, the surface of the counter electrode is usually much bigger than that of the working electrode. Accordingly, its large capacitance can be neglected compared to $C_{DL,WE}$. Additionally accounting for no faradaic reaction on the counter electrode (a hypothesis that might hold in case of a gold electrode for example), the system can now be represented by a Randles circuit. Assuming that diffusion is not the limiting process in that situation, the Warburg impedance can be removed from the electrical model²²⁶.

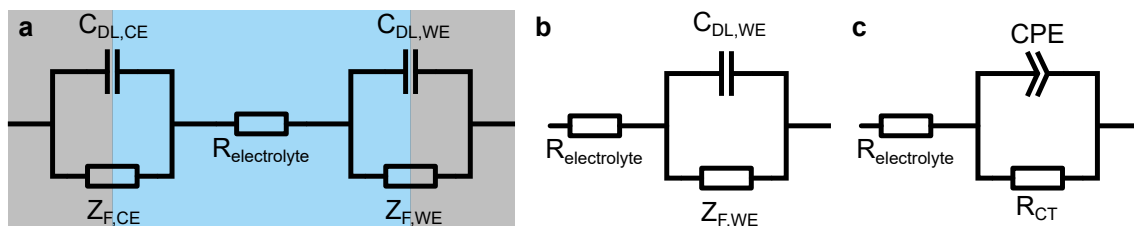


Figure 3.3: Electrical model of the two-electrode electrochemical cell. a) Equivalent model considering equivalent faradaic and capacitive behavior on both the working and counter electrode. b) As the CE is usually much bigger than the WE, the capacitance of the counter electrode can be neglected. If no charge transfer is to be accounted for at the counter electrode (a reasonable assumption if the material is inert), the equivalent circuit can then be represented as a Randles circuit. c) If diffusion is not the limiting process at the WE, the Warburg element can be removed, further simplifying the model. However, in order to account for non-ideal capacitive behavior, a Constant Phase Element (CPE) is used instead of a capacitor.

At high frequencies, the double layer capacitance does not have time to charge and behaves as a short circuit, thus screening the charge-transfer resistance. Theoretically, the impedance measurement therefore shows the electrolytic resistance. In

practice, $R_{electrolyte}$ will also include the impedance of the wires, but in most cases their contribution remains insignificant. Conversely, at low frequencies, C_{DL} can be replaced by an open circuit. The equivalent circuit then become the sum of all the series resistances.

So far, we have considered the case of an ideal flat electrode surface. Yet, verifying this assumption under practical conditions is challenging, and often even impossible due to the roughness of solid electrodes. In turn, this causes an uneven distribution of properties at the electrode-electrolyte interface. As a consequence, a single capacitance is not enough to model correctly the experimental data recorded from real-world EIS measurements. A new element must thus be introduced: the Constant Phase Element, or CPE, which impedance can be written as

$$Z_{CPE} = \frac{1}{Q(j\omega)^n}$$

with $0 \leq n \leq 1$. For $n = 0$, the CPE will behave like a resistance, while a value of n close to 1 indicates that the system operates almost like a capacitance. For $n = 0.5$, the impedance is that of a Warburg element. The electrical model of an electrochemical cell can finally be as represented in Figure 3.3c.

I.3 EIS setup

The impedance measurements shown in this manuscript were performed using a Solartron Analytical Ametek impedance analyzer.

I.4 Impedance modeling

Circuit impedance modeling was performed using the open source EIS Spectrum Analyzer*. The different parameters were manually adjusted to fit the experimental raw data as best as possible.

II MEA fabrication

All the multielectrode arrays used during the experiments presented in this manuscript were microfabricated in the cleanroom of IEMN.

A glass substrate (5 x 5 cm) was immersed successively in water, acetone and isopropanol and placed in an ultrasonic bath, before undergoing O_2 plasma (1000 W, 15 min). The substrate was then put on a hot plate to dehydrate for 5 min (110°C). HMDS was first spin-coated to act as an adhesion promoter (closed cover, 2000 rpm during 20 s) before a positive UV resist was deposited (AR-P5320 50/50, open cover at 1500 rpm for 40s). Electrode patterning was then performed using a mechanical mask (5 s of exposure). Following the exposure, the resist was developed for 1 min (2:3 ARP300-26:DIW), before metallization (10 nm of Ti and 190 nm of gold, by evaporation). After lift-off in an ultrasonic bath with acetone,

*<http://www.abc.chemistry.bsu.by/vi/analyser/>

the substrate was put under UV-ozone for 10 min before being chemically functionalized to prepare for the deposition of the insulating layer of parylene-C (200 mL n-hexane : dichloromethane (7:3) + 0.2 mL of acetic acid + 0.2 mL of [3-(Methacryloyloxy)propyl]trimethoxysilane).

A 1 μm thick layer of parylene-C was then evaporated onto the substrate, before the openings were patterned via UV-photolithography (AZ40 XT (50/12), open cover, 2500 rpm for 15 s followed by 20 s of insolation). After development (3min30s in AZ 726 MIF developer), the parylene layer was etched via reactive ion etching (O_2 plasma, 150 W, 100 mT and 30 sccm for about 6 min). The remaining of the protective resist layer was finally striped off in acetone ultrasonic bath. The fabrication protocol is summarized in Figure 3.4. A culture chamber could be glued to the substrate if necessary, using a biocompatible epoxy.

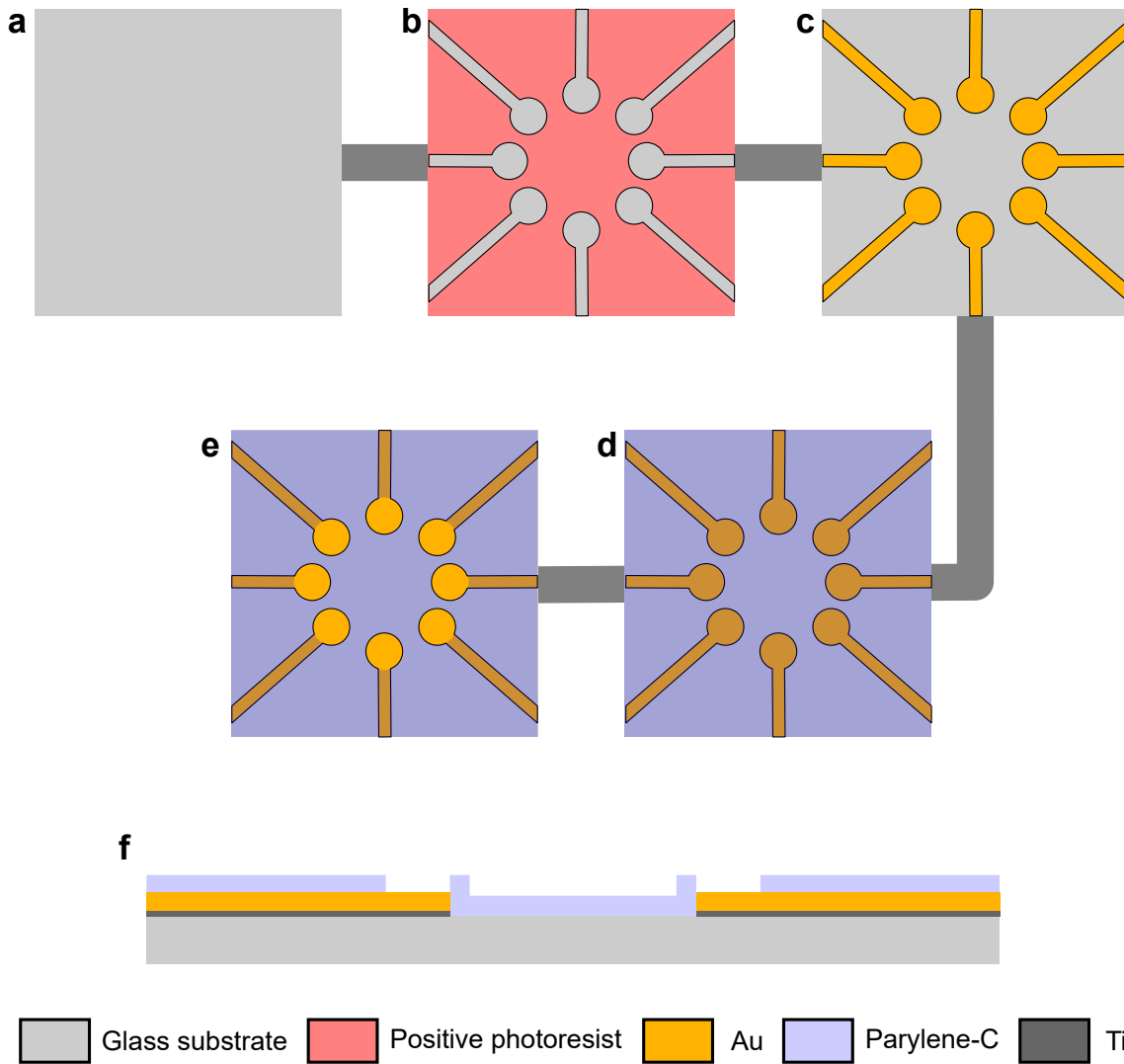


Figure 3.4: Fabrication of a MEA. a) Cleaning of the glass substrate. b) Electrode patterning through UV-photolithography. c) Gold evaporation and lift-off. d) Chemical functionalization and Parylene-C deposition. e) Patterning through UV-photolithography, opening with reactive ion etching and strip-off in acetone. f) Sectional view of the final device.

A picture of the final device is presented in Figure 3.5.

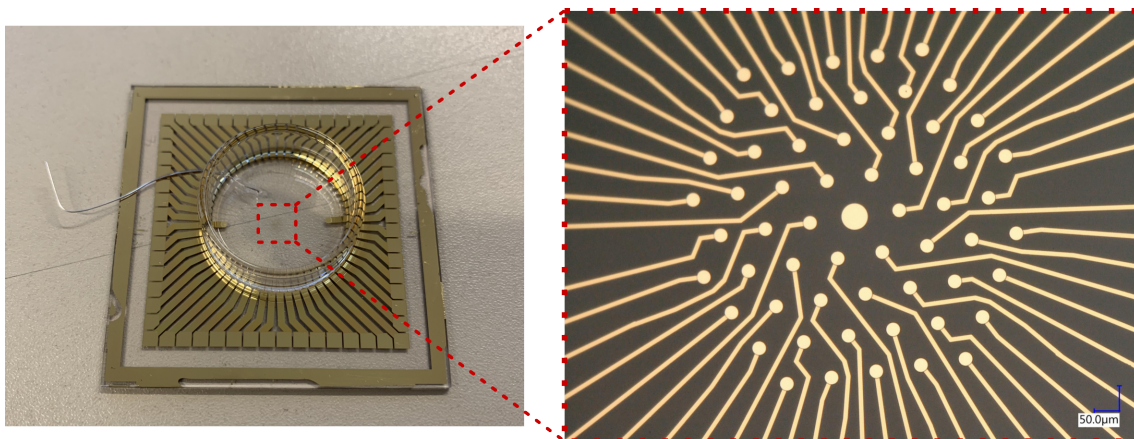


Figure 3.5: A microfabricated MEA. The left picture shows a macroscopic photo of the device where the culture chamber as well as a silver wire to serve as an external reference can be appreciated. The right microscopic photo shows the electrodes of the MEA.

III Dendritic growth

Dendritic growth was performed using bipolar AC electropolymerization in a drop of aqueous solution containing 10 mM of EDOT, 10 mM of benzoquinone (BQ) and 1 mM of NaPSS, unless specifically mentioned otherwise. Square-wave signals were generated using either a Keysight 33600A Trueform waveform generator or a 50 MS s⁻¹ dual-channel arbitrary waveform generator from Tabor Electronics. Only the frequency and peak-to-peak amplitude were occasionally varied (the values are given in the text of this manuscript when they matter), while the other electrical parameters were always kept unchanged (notably the duty cycle at 50%, and no offset voltage).

During this thesis work, two different types electrogenerated objects were studied: freestanding dendrites and dendrites grown on a 2D substrate.

III.1 Freestanding dendrites

The denomination “freestanding dendrites” refer to dendrites that were grown between two electrodes in a drop of the above-described electrolyte, at a controlled height from a parylene-C covered substrate. The two electrodes consist in two 25 μm diameter gold wires (purchased from GoodFellow, Cambridge, UK), separated by a distance of about 240 μm and operating sequentially as the working and counter electrodes of the system. The usual setup for the growth of freestanding fibers is shown in Figure 3.6.

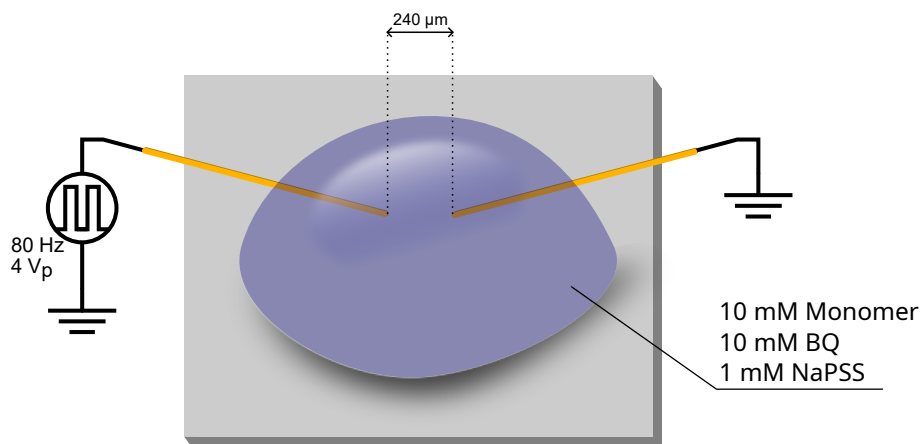


Figure 3.6: Usual experimental setup for the electropolymerization of free-standing polymer fibers. The values of the different parameters are given as an indication but are prone to modification during the different experiments.

III.2 Dendrites on a 2D substrate

Other dendrites were grown onto microfabricated multielectrode arrays fabricated in the cleanroom of IEMN following the protocol described in MEA fabrication. The situation is very similar to that described in the previous subsection. The usual setup for the formation of these objects is shown in Figure 3.7.

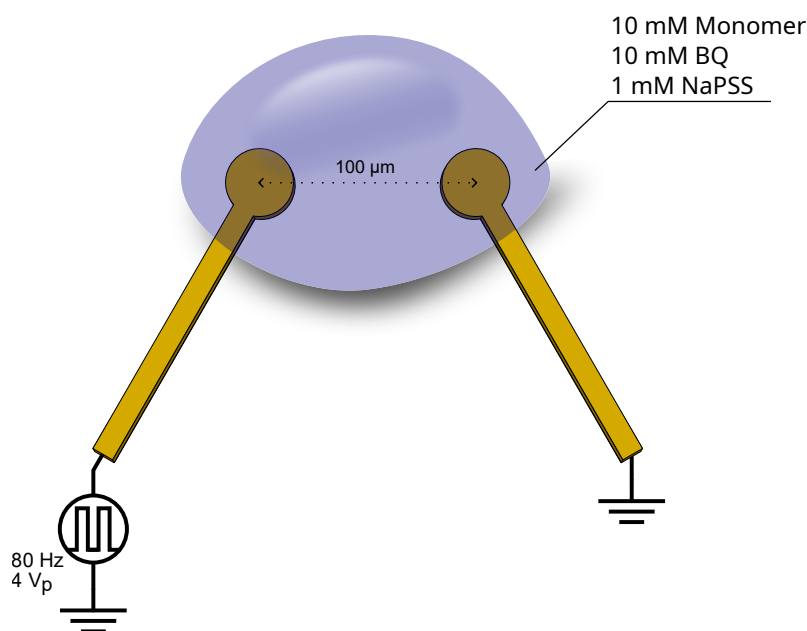


Figure 3.7: Usual experimental setup for electropolymerization on a 2D substrate. The values of the different parameters are given as an indication but are prone to modification during the different experiments.

III.3 Image processing

In all cases, growth was monitored through a VGA CCD color Camera (HITACHI Kokusai Electric Inc).

IV DC characterization

Throughout this work, electrical characterizations were carried out using an Agilent B1500A Semiconductor Device Analyzer coupled with B2201A Switching Matrix.

V Scanning electron microscopy

SEM images were obtained using an electron microscope ULTRA 55 (Zeiss) equipped with two detectors for secondary electrons (InLens and Everhart Thornley) and a low voltage/high voltage BSE detector.

VI *In vitro* electrophysiology

In vitro experiments were performed by Camille Lefebvre at Lille Neuroscience & Cognition Research Centre, University of Lille, INSERM U1172, Lille, France under the supervision of Dr Sophie Halliez, member of the “Alzheimer & tauopathies” team led by Dr Luc Buée. The EOPS facility (PLBS UAR 2014 - US 41) took care of the animals. Experimental research was performed with the approval of an ethics committee (agreement APAFIS #43474-2023050714441306 v6) and follows European guidelines for the use of animals.

VI.1 Sterilization with ethanol (EtOH)

MEAs were sterilized using a 70% ethanol (EtOH) solution. 1 mL of the EtOH solution was introduced into the cell culture chamber of each MEA for 10 minutes in ambient conditions. Following sterilization, the ethanol was aspirated, and the chambers were rinsed three times with sterile water. The MEAs were then allowed to air dry in a biosafety cabinet (BSC).

VI.2 Coating

Cells or slices were plated in poly-D-lysine (0.5 mg/mL, Sigma) and laminin (10 µg/mL, Sigma)-coated MEA to promote cell adhesion. A small drop of coating solution was deposited onto the electrodes of each MEA, and the arrays were placed in a BSC for 2 hours. Excess coating solution was then removed, and the MEAs were rinsed twice with sterile water. The coated MEAs were dried in a BSC before being used for cell culture. Neural cells or brain slices could then be plated on the in-house developed MEAs.

VI.3 Primary Embryonic Neuronal Culture

Cell culture preparation

Mouse primary cortical cells were prepared from 15-days-old C57BL/6JRj mouse embryos. Gestating females (Janvier Labs) were housed in a temperature-controlled (20 - 22°C) room maintained on a 12 h day/night cycle with food and water provided

ad libitum. Cortical cells were prepared as previously described²²⁷. For each sample, one drop of a cellular suspension at 4000 cells/ μ L in cell culture medium (Neurobasal supplemented with B-27, Antibiotic-Antimycotic and L-glutamine, all from Gibco) was seeded on the MEA recording area. Ten minutes after plating, 1 mL of fresh medium was added and the MEA culture chambers were sealed with a removable hydrophobic semipermeable membrane cover (fluorinated ethylene-propylene copolymer, 12.5 microns thick, from Sigma Aldrich). Cultures were kept in a cell incubator (ThermoScientific) at 37°C in a 5% CO₂ and 95% humidity atmosphere. The culture medium was never changed but small amounts of fresh medium were added once a week to compensate for medium evaporation (never more than 10% of the initial volume).

Cell viability assessment

For the live cell imaging, 0.1 μ M of NeuroFluorTM NeuO (Stemcell technologies) was added in the cell cultures. After 10 minutes, fluorescent pictures were then taken using an inverted microscope (Leica DM IL LED) and a camera (MC120 HD). The images were modified using Adobe Lightroom’s auto-correct mode to enhance contrast and visibility.

VI.4 Organotypic culture

Slice preparation

Entorhinal cortex slices were prepared from 7-days-old C57BL/6JRj mice as previously described²²⁸ and kept in an incubator at 37°C (5% CO₂ and 95% humidity). The culture medium was changed daily.

Cell death assessment

1% propidium iodide was added to the culture medium to assess cell viability. The samples were then rested 1 hour in incubator at 37°C (95% humidity and 5% CO₂). Pictures were taken with a LEICA MC120 HD camera, and later modified using auto-modification in Adobe Lightroom to improve contrast and increase visibility.

VI.5 *In vitro* electrophysiological recordings

In vitro electrophysiological recordings were performed using a MEA2100-system (Multi Channel Systems) experimental setup with a 60 channels headstage (Multi Channel Systems) that amplifies and digitizes the signals, and an interface board (MCS-IFB, Multi Channel Systems) connected with GUI on a computer for real-time monitoring, signal filtering, spike detection and recording using the Multi Channel Experimenter software. A temperature controller (TC02, Multi Channel Systems) was used to ensure a constant temperature of 37°C for the cell cultures during the recordings. After leaving the incubator, MEAs were kept 15 minutes in the experimental setup before starting the recordings.

VI.6 *In vitro* electrical characterization of the devices

In vitro characterizations of the polymer dendrites were performed using two portable B2901BL Source/Measure Units (SMUs) from Keysight Technologies. The SMUs were controlled using either the EasyEXPERT software or the Quick IV Measurement software. A TC02 temperature controller from Multi Channel Systems was used to keep cell cultures at a constant temperature during the experiments.

VI.7 Electron microscopy

Preparation of the samples

After removing culture medium, cells were fixed overnight with a solution of 1% glutaraldehyde in 0.1 M Sodium Cacodylate Buffer (Delta Microscopies). The next day, the MEAs were rinsed thrice with PBS and osmium tetroxide was added as a contrast agent²²⁹.

Imaging

SEM images of cell cultures were taken with the help of Dr Nicolas Barois at Institut Pasteur, in Lille, using a Zeiss SEM Merlin Compact VP.

VI.8 Signal processing of electrophysiological data

Spiking and bursting events were detected through the Multi Channel Experimenter software by amplitude thresholding. A threshold was set for each recording electrode as 10 times the standard deviation of the baseline noise level after a software band pass filter from 200 Hz to 4 kHz. Raw and software-filtered recordings were then exported to the NEX format using the Multi Channel DataManager software (Multi Channel Systems). All the analysis were done through the Neuroexplorer software (Nex Technologies). The analysis was performed over the total recording time of five minutes.

Spike sorting

Spike sorting was performed with the help of Alexis Melot and Dr. Pierre Yger from Institut de la Vision and Lille Neuroscience & Cognition Research Centre.

Raw extracellular recordings were processed with SpyKING CIRCUS⁸⁴, through the Python SpikeInterface module. Default parameters were used, and by default all the templates found by the software were kept for further analysis. The temporal width of the templates is fixed to 3 ms. By default, every channel detects the peaks higher than 6 Median Absolute Deviations (MAD) of its noise level.

$$MAD = \frac{1}{n} \sum |x_i - \mu|$$

Signal-to-Noise Ratio (SNR) First, raw data coming from the channels are band-pass filtered with a third order Butterworth filter between 300 Hz and 4750 Hz to analyze spiking activity. A whitening is then performed to remove spurious

spatial correlations between nearby recording electrodes. SNR was calculated for each electrode as the ratio between its maximum spike amplitude and the MAD of its background after setting the threshold by 6xMAD.

Number of neurons detected per electrode After running spike sorting, each computed waveform was then attributed to one of the electrodes of the array based on its amplitude. The detected neuron would be related to the electrode where the waveform amplitude is the highest.

4

EXPLORING THE ELECTROCHEMICAL ENVIRONMENT

I Introduction

In the first chapter of this manuscript, polymer dendrites were presented as quite a new concept that has so far mostly been employed for the implementation of artificial synapses for neuromorphic hardware. To better comprehend their growth and control their properties, Janzakova and coworkers led a thorough investigation on the correlation between growth parameters, fiber morphology and electrical properties, pointing at a structure-property relationship¹⁵¹. In the continuation of this work, and in a quest to refine our understanding of the underlying mechanisms governing dendritic growth, this chapter will focus on the impact of the chemical composition of the aqueous electrolyte on the final morphology and properties of polymer dendrites.

As presented above, electropolymerization is a process influenced by numerous parameters. Although some studies have worked to uncover its underlying mechanisms^{154,230,231}, the inherent stochasticity governing material morphology hinders reliable and consistent control, and requires a statistic approach (similar to biological studies). Apart from the aforementioned electrical parameters, electropolymerization appears to strongly depend on its chemical environment. Already, the influence of the monomer and the supporting electrolyte on the morphology of the fibers was evidenced by different studies^{148,150}. Still, the electroactive solution used to grow the dendrites is a complex system, composed of different electroactive species with distinct properties, such as the redox potentials. The choice of solvent is also crucial, as it will define the electrochemical window in which to operate, in addition to the solubility of the monomer as well as the counterion used as a dopant.

In this study, we focus on the influence of three different variables, namely the concentration of NaPSS, the EDOT/BQ ratio and the concentration of monomer in the solution. Lastly, the chemistry of the monomer and its impact on the morphology and electrical properties of the final structure will be assessed.

II Experimental framework

With these considerations in mind, and in order to keep building on the previous work performed in the framework of the IONOS project while minimizing the undesirable influence of unstudied factors in the present study¹⁵¹, the following parameters are fixed:

- Water is used as a solvent. Although this decision shortens the electrochemical window and limits the solubility of EDOT (2.0 g/L at room temperature²³², or about 14 mM), water is non-toxic, polar, volatile, naturally abundant and low cost. It is also an appropriate medium to interface living organisms.
- The setup will comprise two 25 μm diameter gold wires, 240 μm apart, immersed in a 60 μL drop of the electrolyte.
- The input signal consists of a square waveform with a frequency of 80 Hz, a peak amplitude of 4 V and a duty cycle of 50%.

The experimental setup for the electropolymerization of freestanding dendrites in a 3D environment is depicted in Figure 4.1.

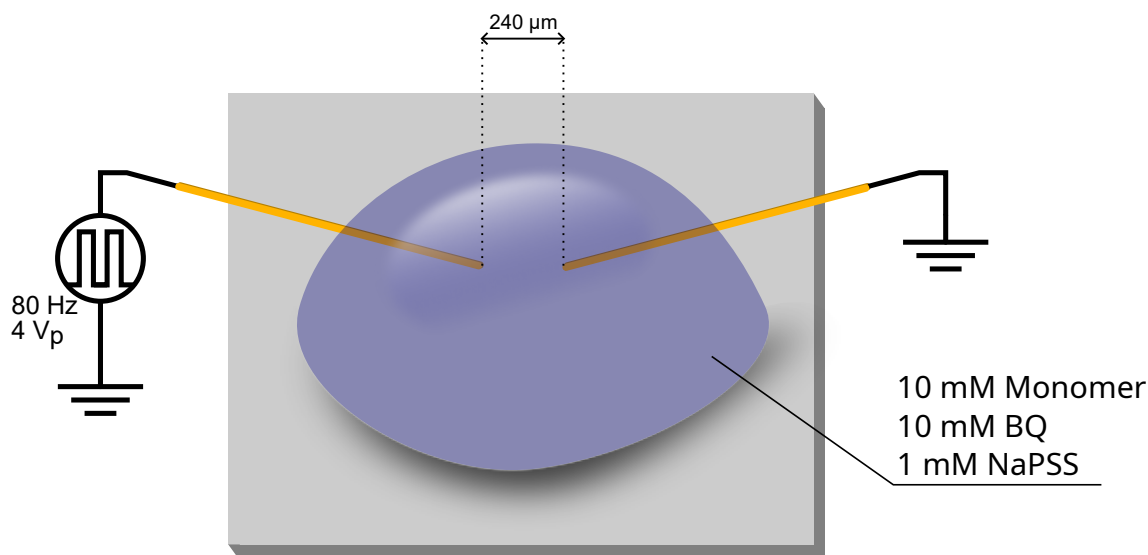


Figure 4.1: Experimental setup for the electropolymerization of freestanding polymer fibers.

More details regarding electropolymerization are to be found in the MATERIALS AND METHODS chapter.

III Influence of the concentration of supporting electrolyte

When it comes to the electropolymerization of conductive polymers, the choice of supporting electrolyte is not trivial. Indeed, this latter plays different important roles. First, it helps stabilizing the dispersion of monomers in a continuous liquid phase. Indeed, EDOT being poorly soluble in water, adding PSS as surfactant allows

to help with the preparation of the solution. Second, NaPSS is used as the dopant and is necessary to the electropolymerization process as a supporting electrolyte, as described in AC electropolymerization. If these last two roles had already been made clear in the first chapter of this manuscript, salt also makes deionized water conductive through the addition of ionic species to a solvent that would otherwise act as a dielectric.

Considering a symmetrical process on the two gold electrodes, which both alternate between anodic and cathodic behavior, one can make the assumption that the electropolymerization rate is similar on both sides of the system. As a first approximation, the equivalent electrical circuit of the system during growth can thus be modeled as shown in Figure 4.2a, with a charge-transfer resistance $R/2$, a double layer capacitance $2C$ and $1/G$ the resistance of the electrolytic solution that is directly proportional to the inverse of the ionic concentration.

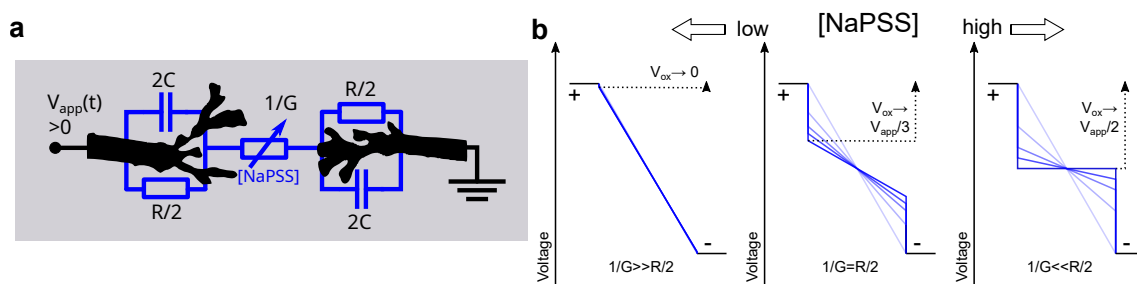


Figure 4.2: Influence of NaPSS on the voltage drops in the system. a) Equivalent electrical model of the system during growth. b) Representation of the drop of voltage in the system for three different ionic concentrations, successively when the resistance of the electrolyte is much greater than, comparable to and negligible compared to the charge transfer resistance at the metal-electrolyte interface.

Figure 4.2b illustrates the importance of ionic concentration in the electropolymerization process, as it determines the voltage drops across the system. When NaPSS concentration is low, the impedance of the solution increases dramatically. Most of the drop of voltage therefore happens within the electrolyte, rather than at the electrode-electrolyte interface, meaning that the oxidation potential is not reached at the electrode and that polymerization cannot happen. Conversely, for high concentrations of salt, the electrolyte becomes highly conductive and voltage drop is maximized at the electrode-electrolyte interfaces, possibly causing water electrolysis if the applied voltage is important enough.

The resistance of the electrolyte therefore appears to be an essential factor. Consequently, the concentration of supporting electrolyte must not be left to chance. In order to access the concentration range within which electropolymerization can happen, dendrites were grown into electrolytic solutions with different concentrations of NaPSS, keeping the other parameters of the system constant as mentioned earlier (growth frequency $f = 80$ Hz, distance between the wires $d = 240$ μm , peak voltage $V_p = 4$ V, $[\text{BQ}] = [\text{EDOT}] = 10$ mM). Every set of conditions was repeated twice. Figure 4.3 presents the final morphology as well as the conductance and completion time values of the resulting dendrites.

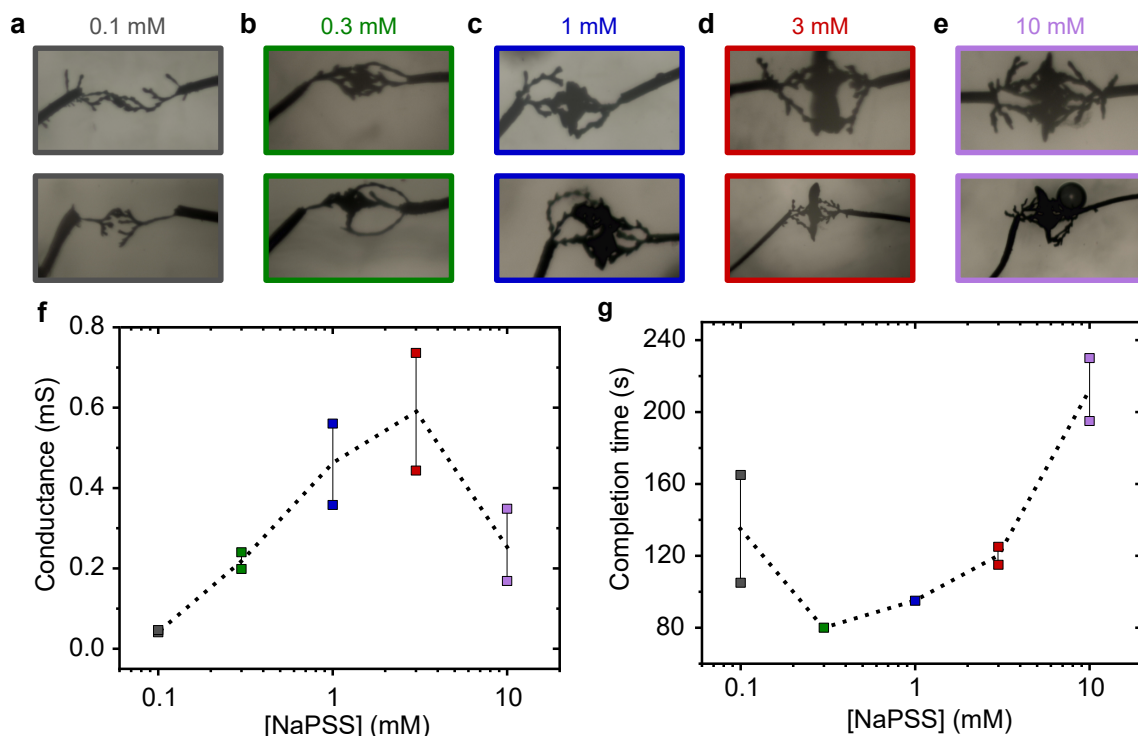


Figure 4.3: Influence of the concentration of NaPSS on the a-e) morphology, f) conductance and g) completion time of the dendrites.

Growth did happen from salt concentrations ranging from 0.1 mM to 10 mM, as shown in the pictures in Figure 4.3a-e. Higher and lower concentrations of salt did not lead to dendritic growth experimentally, although it can be noted that, at 100 mM, a dark veil appeared around the electrodes and some material was deposited onto the gold wires, but no dendrite appeared and bubbles were formed due to water electrolysis. At these concentrations, PEDOT might be too soluble to stick to the gold electrodes.

Electrolyte concentration does not appear to have an important influence over the morphology of the fibers. Even if at 0.1 mM the dendrites tend to be more linear, for higher concentrations the impact is visually insignificant. This is in accordance with other studies pointing at the low impact of the supporting electrolyte on the morphology of electrodeposited polymer films²³³. However, Figure 4.3f highlights a positive correlation between salt concentration and conductance up to 3 mM. In the meantime, the completion time of the dendrites (defined as the duration between the start of electropolymerization and the first contact between both tips of the dendrites originating from the two electrodes) follows a V-shaped curve with a minimum reached for 0.3 mM. A trade-off thus has to be found between conduction and completion time.

The increase of conductance with salt concentration could originate from the fact that NaPSS is used as a dopant for the PEDOT. Therefore, increasing its concentration might indeed help to increase the conductivity of the polymer through the doping of more units. Yet, some studies suggest that the composition of electrodeposited PEDOT:PSS does not correlate with anion concentration in the elec-

trolyte²³⁴. Morphology is also a relevant parameter, and the changes in conductance could be due to a higher quantity of material participating in the conduction of current, rather than to an increased conductivity.

As addressed earlier, the concentration of salt in the electrolytic solution controls the voltage drops in the medium. When the resistance of the electrolyte is negligible compared to the charge-transfer resistance at the interface, the voltage drops mainly take place at the interfaces between the electrodes and the electrolyte. The oxidation voltage V_{ox} is then almost equal to half of the voltage applied at the terminals of the circuit, leading to the electrolysis of water and the formation of H_2 (g) as it is the case for $[NaPSS] = 100$ mM. Even if the voltage drop is not sufficient to produce gas, it might still be enough to overoxidize PEDOT, thus decreasing its conductivity^{235,236}. On the other hand, if the resistance of the electrolyte becomes comparable to the charge-transfer resistance at the interface, the drop of potential happens evenly in the circuit, meaning that V_{ox} now amounts to approximately a third of the applied voltage. This might limit the kinetics of the reaction, or even prevent it altogether from happening if the oxidation potential is not reached.

Increasing salt concentration also means that the quantity of ions in the solution increases, possibly making it harder for EDOT to oxidize at the anode as more anions could accumulate at the surface of the electrode^{154,230}, effectively reducing the surface accessible to the monomer and increasing completion time. In addition, PSS is not only used as a dopant, but also as a surfactant, which could impact the formation of the polymer in the solution²³⁷. Anionic solubilizing agents are known to lower the oxidation potential of EDOT²³². This could have two opposite effects: increasing charge carrier density in the conducting polymer as well as its conjugation length (and thus the conductivity), but also facilitating overoxidation^{146,238}.

Impedance measurements were then performed on the dendrites with the experimental setup presented in the inset of Figure 4.4a, using a small Ag wire ($\varnothing = 25$ μ m) as the output of the system and the shorted dendrite as the input for the sinusoidal perturbation. The results are presented in Figure 4.4.

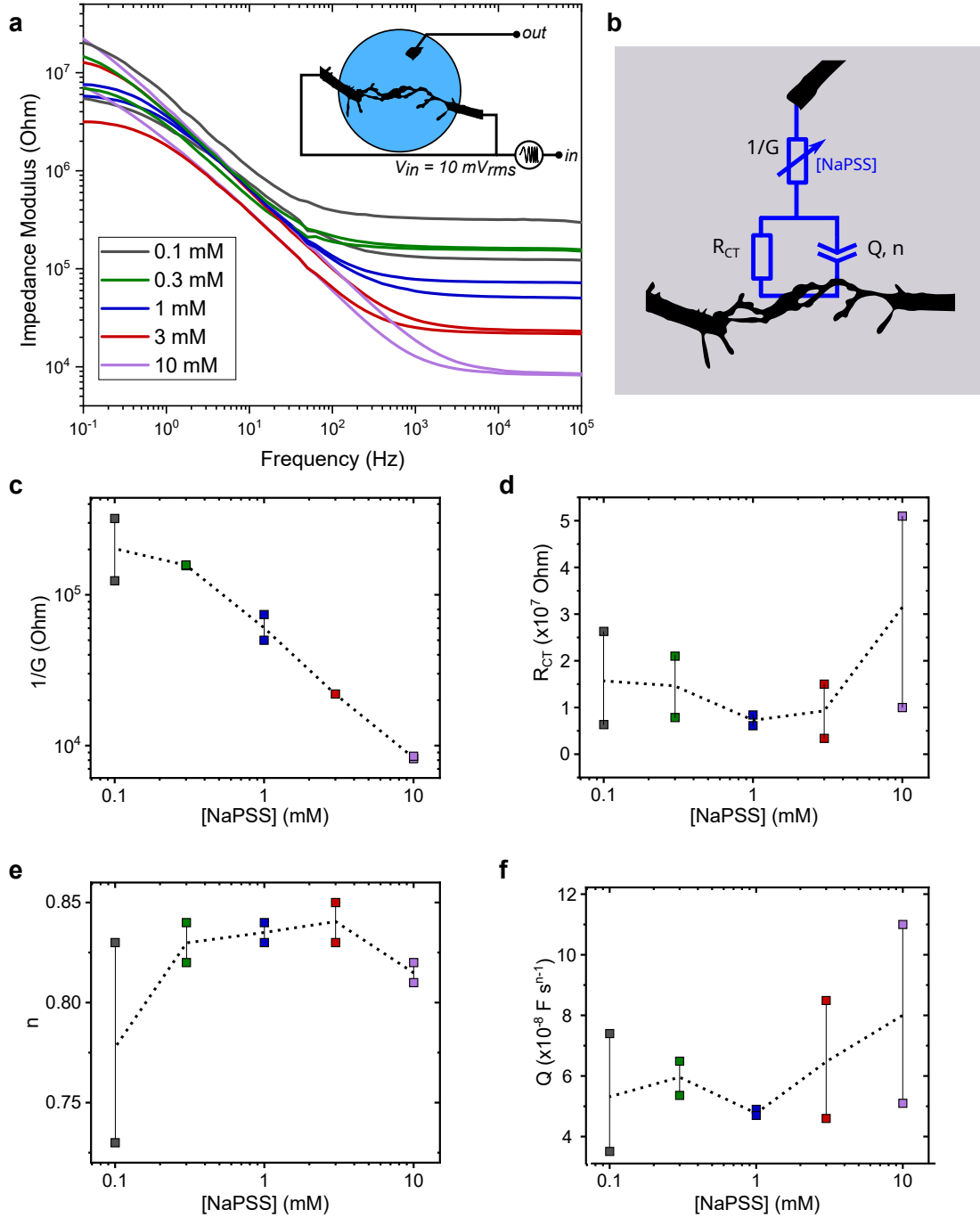


Figure 4.4: Impedance measurements and impedance modeling. **a)** Impedance modulus plot. Inset: experimental setup for the AC electrochemical impedance of the dendritic structure, with the two ends of the dendrite shorted on the input and the silver gate electrode as the output. **b)** Equivalent circuit used for the impedance modeling: a resistance R_0 in series with a resistance R_1 in parallel with a constant phase element (CPE), which impedance is defined as $Z_{CPE} = 1/(Q_0\omega^n) \cdot e^{-i\pi/2n}$. **c-f)** Evolution of the different fitting parameters (respectively $1/G$, R_{CT} , n and Q) as a function of the concentration of salt in the electrolyte.

The quantity of salt in the growth electrolyte does not seem to affect much the impedance of the final dendrites. Figure 4.4a shows the impedance modulus plots acquired at the end of the growth, as well as the value of the different fitted parameters for each NaPSS concentration. As expected, the decrease of $1/G$, related to the resistance of the electrolyte, is inversely proportional to the increase in salt concentration (Figure 4.4c), as shown by the high frequency plateau observed in Figure 4.4a. Conversely, for the other electrochemical parameters (R_{CT} , Q and n), no trend emerged from the modeling, suggesting that the interface between the dendrite and the electrolyte might not depend on the quantity of salt. This is not so surprising for the charge transfer resistance: as no faradaic reaction is to be expected, the value of R_{CT} is quite high, on the order of 10^7 Ohm (Figure 4.4d). The steadiness of the fitted values of Q and n (Figure 4.4e-f) for the Constant Phase Element (CPE) can also be understood in terms of morphology. As it was observed in previous studies^{147,239,240}, the geometry and morphology of the structure is what appears to matter when it comes to estimating the values of the fitting parameters of the equivalent model. In this case, as the morphology of the dendrites does not evolve much with the concentration of salt (Figure 4.3a-e), it is not unsettling to consider similar values for these parameters.

In conclusion, salt concentration seems to mainly affect dendritic formation through its impact on electrolytic resistance. Neither the final morphology of the dendrites nor the conductance measurement allow to conclude with certainty about the role of NaPSS on these parameters. Yet, the supporting electrolyte remains an important factor to take into account when considering electropolymerization, as there is a range out of which no growth was observed. Too low a concentration might prevent growth, while a high concentration might degrade the polymer properties. Further analyses (e.g., XPS) might be necessary to establish the composition and structure of the polymer fibers and correlate the stoichiometry of the growth solution with that of the final product and its doping level.

IV Influence of the concentration of oxidizing agent

Since its reduction consumes two electrons in addition to two protons (exactly what EDOT releases during its oxidation), benzoquinone (BQ) is introduced in the growth electrolyte to balance the redox reaction at the cathode^{148–151}. BQ is an unstable chemical that tends to easily reduce to hydroquinone, mediating electron transfer during the oxidation of EDOT and facilitating the reaction. Modifying the EDOT-to-BQ ratio might thereby have an influence over the kinetics of the electropolymerization process, as it influences the charge-transfer resistances at the interfaces.

Six dendrites were grown at different EDOT:BQ ratios, ranging from a large excess of EDOT molecules (10000:1) to an amount of BQ twice superior to that of EDOT. In each situation, the concentration of NaPSS was set to 1 mM, and the total amount of electroactive material was kept constant throughout the experiments ($[EDOT] + [BQ] = 10$ mM). Indeed, fixing one of the concentrations while varying the second one could have been an option in this situation, as the water solubility of BQ is greater than that of EDOT. Yet, we tried to keep the system

as close as possible to the usual one, to avoid increasing the number of variables that could influence growth. Setting the concentration of NaPSS allowed to fix the conductivity of the electrolyte, while the constant amount of electroactive material throughout the experiments meant that the overall quantity of reactive species in the solution would not modify the electrochemical environment. In practice, this would isolate the effect of ratio changes and allow comparability and reproducibility across experiments. The value was chosen to accommodate with the solubility of EDOT.

Contrary to the concentration of salt in the electrolyte, the amount of BQ in the medium had a clear influence on the morphology of the final dendrite (Figure 4.5a-f). For high ratios (down to 100:1) and despite a high concentration of EDOT, a thin and linear dendrite was formed, in a pearl necklace fashion. Interestingly, these dendrites grew mainly out of one of the two electrodes of the system in a very directed manner, in a fashion that reminds that caused by a voltage offset^{151,154}, instead of the simultaneous growth from both electrodes that can usually be observed. Around equimolarity, the dendrite became thicker with more branches. This evolution of the shape correlates with the conductance of the structure, as thinner dendrites present a lower conductance, while those grown at equimolarity show higher numbers (Figure 4.5g). Let us also note that no growth was observed in the absence of BQ, nor did we manage to grow dendrites with higher contents of BQ in these conditions, most probably due to the low amount of EDOT necessary to keep constant the amount of electroactive material in the system. However, we did achieve to grow a dendrite for a 1:10 EDOT-to-BQ ratio by increasing both the concentrations of EDOT and BQ, which tends to prove that the low amount of monomer was the limiting factor (see Exploring the impact of low EDOT concentration for high EDOT-to-BQ ratios). This parameter will be studied in more depth in the following section.

The amount of BQ in the system appears to govern the time required for the dendrites to form, as the completion time curve presents a V shape and reaches a minimum of about 100 seconds for a 10:1 EDOT-to-BQ ratio (Figure 4.5h). Therefore, a slight excess of EDOT appears to be the perfect condition to minimize the time needed for growth while maximizing conductance.

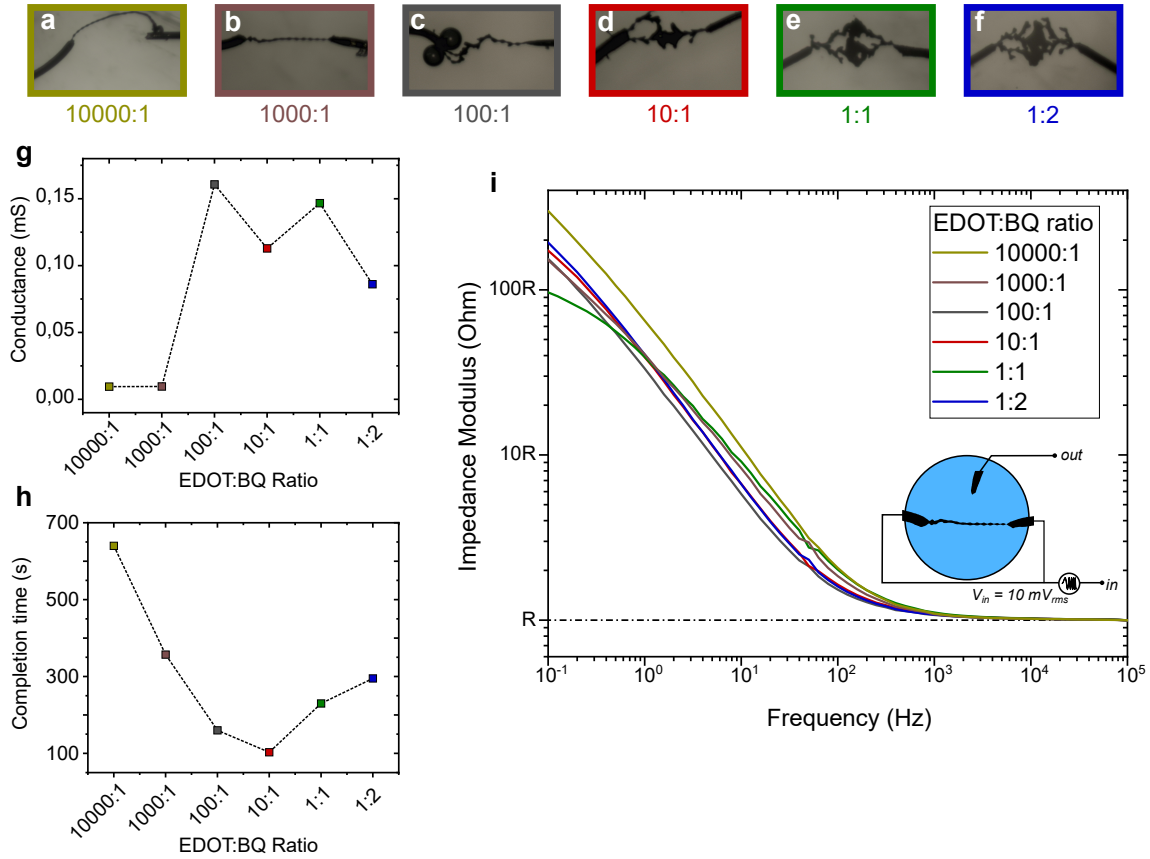


Figure 4.5: Influence of the EDOT:BQ ratio on the growth and properties of a dendrite. a-f) Pictures of the dendrites at the end of the growth, for different EDOT:BQ ratios ranging from 10000:1 to 1:2. g-h) Conductance and completion time of the dendrites as a function of the EDOT:BQ ratio. i) Impedance modulus plot normalized using the high frequency plateau corresponding the impedance of the electrolyte R as the reference. Inset: experimental setup for the AC electrochemical impedance measurement of the dendritic structure, with the two ends of the dendrite shorted on the input and the silver gate electrode as the output. The curves were normalized to the high frequency resistance plateau to help with the comparison.

Although the morphology of the dendrite was this time visibly affected by the EDOT:BQ ratio, their impedance on the other hand seems to remain mostly unchanged (see Figure 4.5i). If one were to refer to the equivalent model discussed in Figure 3.1, the fitting parameters of the different dendrites must be similar: as the slope and cutoff frequency are very close, so should be the values of n and Q for the CPE, given that the resistance of the electrolyte (not visible here as the curves are normalized) does not fluctuate much from one dendrite to another (the concentration of salt is fixed for this set of experiments). These results were quite surprising, as it is well known that the impedance of an object is correlated with its morphology, and more especially with the roughness of its surface²⁴⁰.

IV.1 Exploring the impact of low EDOT concentration for high EDOT-to-BQ ratios

In order to assess whether or not the low concentration of EDOT was the reason behind the absence of growth for ratios higher than 1:2, an electrolyte was prepared bypassing the constraint fixing the total amount of electroactive species in the solution. An electrolyte with a 1:10 EDOT-to-BQ ratio was thus used, with a monomer concentration of 4 mM, as growth was still previously observed for a lesser concentration. The amount of NaPSS was kept to 1 mM.

In these conditions, dendritic growth was observed. Once again, the electric properties of the resulting dendrites did not appear to be influenced by the modification of the electrochemical environment, but it has to be reported that growth happened rather unconventionally, in a zigzag manner, as can be observed in Figure 4.6.

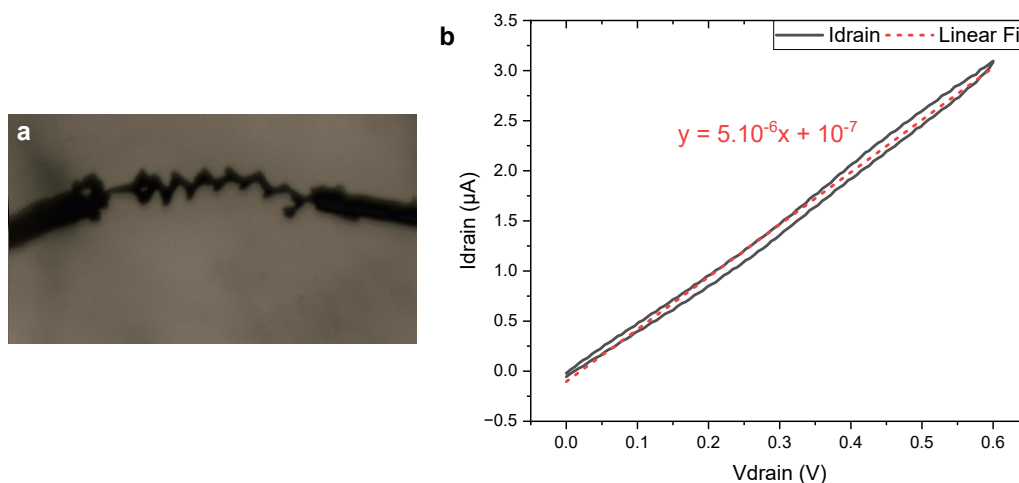


Figure 4.6: Dendritic growth for a 1:10 EDOT-to-BQ ratio, with $[\text{EDOT}] = 4 \text{ mM}$. **a)** Microscopic picture of the dendrite after growth termination. The two gold wires are $240 \mu\text{m}$ apart. **b)** Output characteristic of the dendrite when a potential difference is applied between its terminals. The linear fit (in red) provides an estimation of the conductance of the dendrite, in this case $5 \mu\text{S}$.

The dendrite was formed in 405 seconds, which is well correlated with the results presented in Figure 4.5 and tends to confirm that the increase in completion time is not only due to the lower monomer concentration but also to the increase of BQ. However, its estimated conductance of about $5 \mu\text{S}$ makes it the less conductive dendrite. The large quantity of BQ present in the system could facilitate the overoxidation of the polymer, decreasing its conductivity, but this effect could also be attributed to the morphological variance or a mechanical frailty of the dendrite. As always, this interpretation remains up for discussion.

IV.2 Conclusion

BQ undergoes electrochemical reduction quite easily. This might be why, even at very low quantities of BQ, one can still witness growth, whereas no dendrite could form without the presence of BQ in the electrolyte (respecting the set of imposed conditions). At the other end of the spectrum however, because of the

low solubility of EDOT in water, reaching low EDOT:BQ ratios was synonymous with low concentrations of EDOT given our set of experimental conditions. This mostly explains why it was not possible to grow dendrites for ratios below 1:2 (EDOT:BQ) while respecting the constant amount of electroactive material in the solution, and will motivate us to explore how the concentration of EDOT regulates growth in the next section.

V Influence of the concentration of monomer

The previous experiment highlighted the influence that the concentration of the precursor monomer, EDOT, had on the growth of a dendrite. For this experiment, the scope of exploration was limited by the low solubility of EDOT in deionized water. Five concentrations were tested, ranging from 3.33 mM to 10 mM, while the distance between the wires, growth frequency and applied potential were set to 240 μm , 80 Hz and 4 V_p respectively. Below that range and under the imposed set of experimental conditions, some deposition was observed, but no dendrite could be completed within a reasonable timescale: in 2.5 mM of EDOT and after about 40 minutes of growth, the resulting fibers are shown in Figure 4.7. In the light of the experiments presented in the previous section of this manuscript, the concentrations of BQ and NaPSS were fixed at 10 mM and 1 mM, respectively.



Figure 4.7: Dendritic growth after 40 min of electropolymerization for a monomer concentration of 2.5 mM, under the set of specified experimental conditions, as detailed in the text.

Figure 4.8a-e shows that the concentration of EDOT in the growth electrolyte does not seem to affect much the morphology of the dendrite, as no evident modification could be detected. This contradicts previous observations that suggested that, at least for in-plane growth, monomer concentration would be one of the parameters influencing the morphology of the grown structure¹⁵⁰. Yet, the range of reachable monomer concentrations was much more limited in our study due to the choice of solvent, which could explain why there were no remarkable differences regarding morphology. If the dendrite grown with the lowest concentration of EDOT appears to be more linear than the others, it is difficult to determine whether this is a consequence of the low concentration of monomer or rather of the high variability inherent to electropolymerization. Conversely, the completion time of the objects seems to be strongly related to the EDOT concentration in the solution, as the time required to complete the dendrite exponentially decreases as the concentration of monomer increases (Figure 4.8g), in accordance in what can be found in

the literature²⁴¹. Meanwhile, the trend for the conductance of the dendrite is the opposite of that of completion time, as it increases with [EDOT] (Figure 4.8f). This does not seem to be correlated with morphology, as all dendrites look similar. As the quantity of NaPSS is equal in all cases, an increased conductivity can also be discarded as the cause of this increasing conductance, as an equivalent doping level would be expected among the five dendrites. The amount of BQ being set to 1 mM, this parameter should also not limit the electropolymerization process, as brought to light in the previous section. Therefore, we suggest that the quality of the junction in the middle of the dendrite might depend on the quantity of EDOT available in the surroundings of the dendritic tips when these two meet during growth. When monomer concentration is high, the gap might be easier to bridge, as supported by the lower completion time compared to when monomer concentration is low.

Once again, the electrochemical impedance of the final structure does not show any relationship with the concentration of monomer in the growth electrolyte (Figure 4.8h), suggesting that chemical parameters do not affect the impedance of the final object.

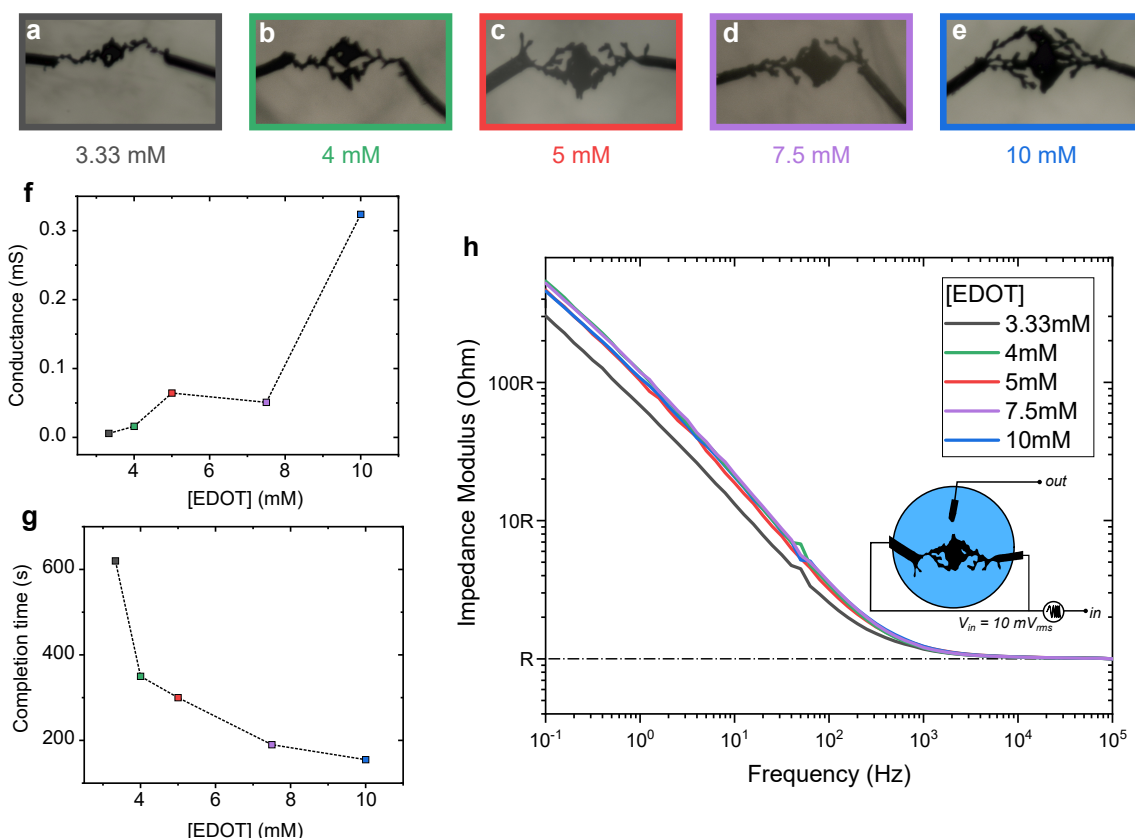


Figure 4.8: Influence of the concentration of EDOT monomer on the morphology and properties of the dendrites. **a-e)** Pictures of the final dendrite at different concentrations of EDOT in the growth solution. **f-g)** Conductance and completion time of the dendrite vs. concentration of EDOT. **h)** Impedance modulus plot, normalized using the high frequency plateau corresponding the impedance of the electrolyte R as the reference. Inset: experimental setup for the AC electrochemical impedance measurement of the dendritic structure, with the two ends of the dendrite shorted on the input and the silver gate electrode as the output.

So far, the study of the composition of the electrolyte and its impact on the morphogenesis of electrogenerated PEDOT fibers has revealed some interesting empiric trends that should be extrapolated to any electropolymerization setup:

1. Salt concentration is an important parameter to take into account during electropolymerization. Even if the results presented earlier do not show any impact of the quantity of NaPSS on the electrical and morphological properties of the dendrites, still there exists an ideal range to allow electropolymerization to happen. This observation could prove critical for growth where the environment cannot be controlled, for example in a biological medium where salt concentration is high.
2. The presence of an oxidizing agent facilitates the electropolymerization process even when only a small quantity is available.
3. Increasing the availability of the monomer allows faster growth as well as an increased conductance of the resulting object.

This enhanced understanding of the system paves the way for a more controlled dendritic growth, as illustrated by the following example.

VI Controlled dendritic growth through the modulation of electrolyte composition

As the previous experiments regarding monomer, oxidizing agent and salt concentration helped to define the framework of polymer fiber electropolymerization, the newly obtained comprehension of the system can now be put into practice. In this section, we will show how dendritic formation can be controlled by adjusting the electrochemical composition of the surrounding electrolyte.

The experimental setup is the same as the one already described previously. Two gold wires ($\varnothing = 25 \mu\text{m}$), $240 \mu\text{m}$ apart, were immersed in an electrolyte containing 10 mM of EDOT and 1 mM of NaPSS, but no BQ, as shown in Figure 4.9. The waveform generator is then turned on, applying a rectangular signal of 80 Hz with an amplitude V_p of 4 V.

During the first two minutes of the experiment, in the absence of oxidizing agent, no growth was to be observed, although the other parameters should allow it. However, when the capillary containing BQ was introduced in the electrolyte and BQ was released, dendritic formation was immediately kickstarted, until full completion of a working dendrite.

Although it was not possible to precisely control the flow of benzoquinone due to a limitation of the setup, this experiment actually suggests that locally manipulating the electrochemical environment might be a way to influence dendritic growth, as it was proposed by Ohira and collaborators who used a micro-space to limit access of the monomer to the sides of the dendrite, thus promoting linear growth¹⁵⁶. One could rely on microfluidic pumps to locally introduce chemical species necessary for

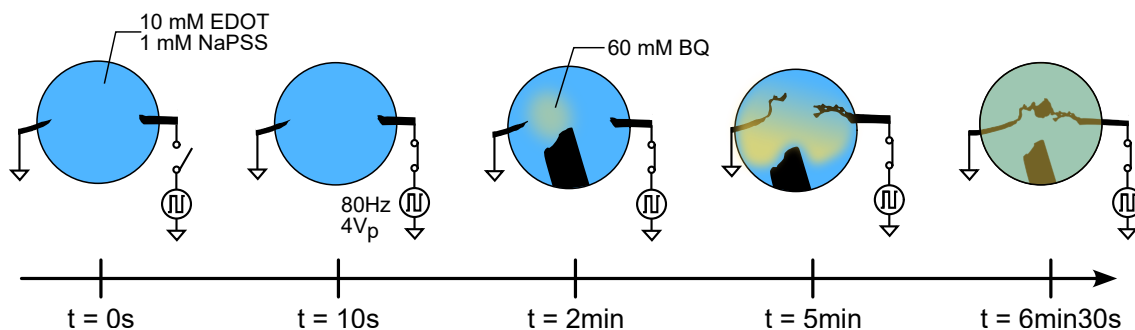


Figure 4.9: Controlled growth of polymer dendrite through the modulation of electrolyte composition, via injection of BQ in the system. First, a signal is applied to the system during two minutes ($t = 10$ s), during which no dendritic growth is observed, before the introduction of BQ through a capillary that kickstarts the electropolymerization process ($t = 2$ min), until the full completion of a dendrite ($t = 6$ min30s).

the electropolymerization process, such as an oxidizing agent (as in the present example) or the monomer itself. This could have tremendous implications, notably in systems where the electrolyte is set and cannot be modified such as biological environments. It would allow the use of non-biocompatible chemicals (e.g., BQ), as the concentration of these could be finely tuned at a precise location, promoting electropolymerization without diffusing in the whole system.

So far, we have explored the influence of the quantity of each electrochemical species present in the electrolyte. The next section of this manuscript will explore the importance of the chemistry of the monomer itself, introducing a new derivative of EDOT to be copolymerized with EDOT.

VII Electro-co-polymerization of EDOT and EDOT-G4

After investigating the influence of the concentration of monomer in the growth solution, the next interrogation that arose was the effect of the monomer itself on the electrochemical properties of the dendrites. Previous studies had already demonstrated that the nature of the monomer was indeed a factor influencing the growth of polymer fibers, sometimes very ostensibly modifying the morphology of the structure^{148–150}. For that purpose, a growing amount of EDOT-G4, a derivative of EDOT modified through side-chain engineering, was introduced in the electrolyte. EDOT-G4 is a derivative of EDOT with a long glycolated side-chain synthesized by Dr David Guérin at IEMN, following the procedure detailed elsewhere¹³⁷. The glycolated side-chain of EDOT-G4 has been shown to increase hydrophilicity via the enhancement of cation affinity and penetration within the bulk of the material. In turn, this material co-deposited with EDOT has demonstrated better performances than unmodified electropolymerized PEDOT for the recording of action potentials in primary cell cultures¹³⁷.

For this experiment, the concentration of BQ and NaPSS and the total amount of monomer in the solution ($[\text{monomer}]_{\text{total}} = [\text{EDOT}] + [\text{EDOT-G4}]$) were set to

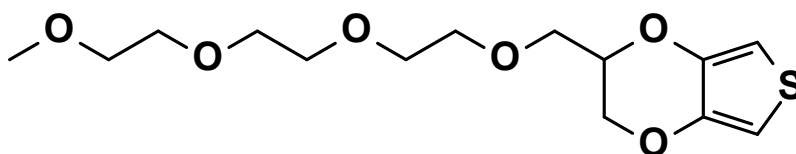


Figure 4.10: Chemical structure of the EDOT-G4 monomer, with its glycolated side-chain.

10 mM, 1 mM and 10 mM respectively. Consequently, the amount of EDOT in the system decreases as the EDOT-G4:EDOT ratio increases. It should be noted here that the EDOT-G4 used for that study was in all respects identical to that used by Ghazal and coworkers¹³⁷, as it came from the same batch. Consequently, no variability is expected to be induced by the synthetization method of the monomer.

Seven dendrites were grown, in the same conditions as described previously. A first object with no EDOT-G4 was used as the reference. Dendrites were formed in an electrolyte containing up to 70% of EDOT-G4, with no or little effect over their morphology (see Figure 11a-g). Above that threshold, it was not possible to observe any dendritic growth. Although we did not check the chemical composition of the dendrites because of their fragility and the difficulty to manipulate them once grown, a previous work led by our team proved that, using the same electrolyte, it was possible to electro-co-deposit EDOT and EDOT-G4 onto a gold electrode. This blend notably showed a lower Young's modulus and improved performances during *in vitro* experiments compared to unmodified PEDOT¹³⁷. The impossibility to deposit any material above the threshold of 70% of EDOT-G4 in the growth solution was also reported in that study, which would tend to corroborate the results obtained here.

Nevertheless, in the absence of material characterization, it remains difficult to assess whether or not EDOT-G4 is indeed present in the structure of the dendrites. In previous studies found in the literature, the choice of monomer was undoubtedly correlated with the morphology of the electropolymerized fibers¹⁴⁸⁻¹⁵⁰, whereas in this present case the structure of the dendrite does not seem to reveal any information about the integration or not of the precursor monomers in the dendritic object. In addition to that, the conductance of the dendrite decreases linearly as the amount of EDOT-G4 increases (Figure 4.11h). Taking into account the morphology of the dendrites and their likeness (Figure 4.11a-g), the decrease in conductance could be associated with a decrease in conductivity. That would point towards the integration of EDOT-G4 within the structure of the dendrite. In the meantime, the completion time of the dendrite increases with as the proportion of EDOT-G4. These two effects are hard to differentiate from those of a dilution, as exposed in Figure 4.8. At this point, although our results seem coherent with those published in the literature¹³⁷, we cannot ascertain the presence of polymerized EDOT-G4 within the dendritic structure based solely on the results showed in Figure 4.11a-i.

In terms of electrochemical impedance, two different regimes can be observed (Figure 4.11j). For the lowest contents of EDOT-G4 (0% and 20%), a plateau appears at low frequencies, suggesting that a permanent regime has been found.

Yet, this plateau disappears as the quantity of EDOT-G4 increases. Strangely, this plateau was never previously observed for any other dendrite, although growth conditions were exactly the same as for the one with no EDOT-G4. Apart from that, here again the impedance of the dendrites remains very similar independently of the concentration of EDOT-G4.

These observations, combined with the dependence of the high frequency plateau with the gate position, suggest that the setup might have a significant influence over the output of the measure. Therefore, one should be careful not to overinterpret the data when reading such plots and assign these variations to changes in the properties of the objects under study, as it is not clear how the setup impacts the measurements. In addition, the choice of the gate is also consequential: as already discussed, material is of great importance²⁴², but geometry is also a decisive factor^{239,243}, as too large of a gate could limit the time constant of the whole system and screen the impedance of the dendrites in the curves.

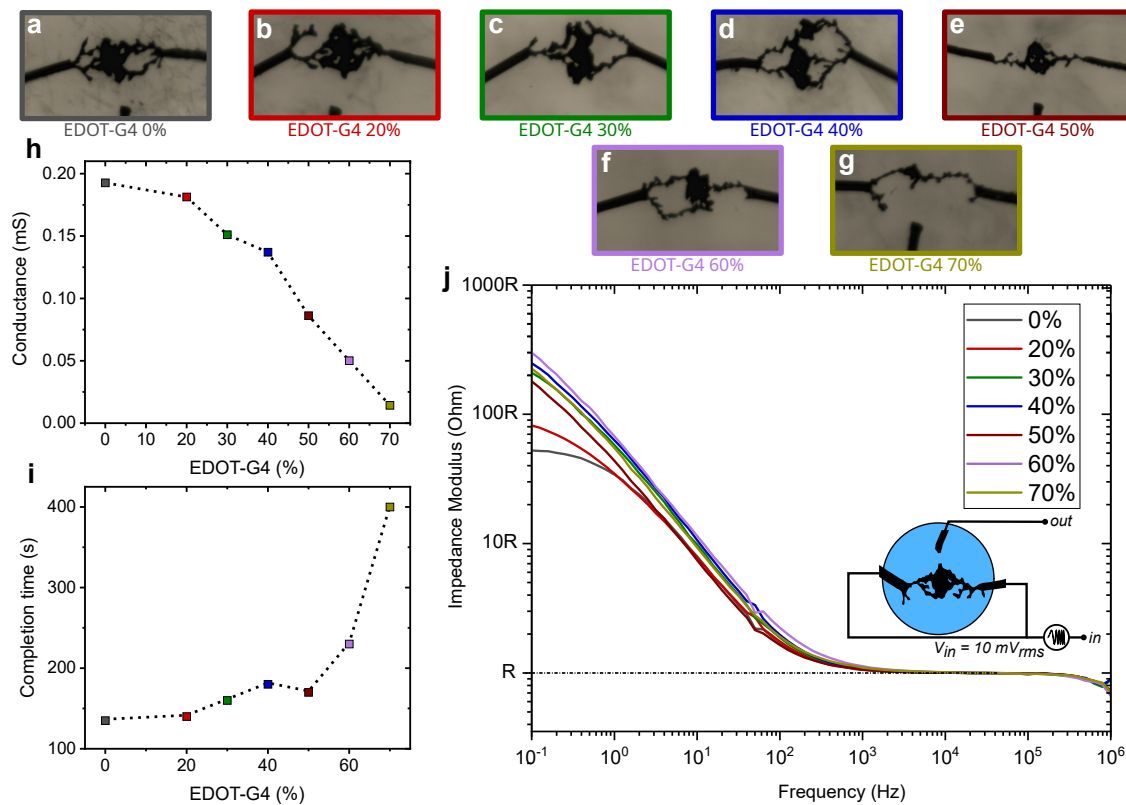


Figure 4.11: Electro-co-polymerization of EDOT and EDOT-G4, a derivative of EDOT with a glycolated side chain, and influence on the morphology and properties of the dendrites. **a-g)** Pictures of the dendrites grown with different ratios of EDOT:EDOT-G4 in the growth solution. **h-i)** Conductance and completion time of the dendrite vs. EDOT-to-EDOT-G4 ratio. **j)** Impedance modulus plot normalized using the high frequency plateau corresponding the impedance of the electrolyte R as the reference. Inset: experimental setup for the AC electrochemical impedance measurement of the dendritic structure, with the two ends of the dendrite shorted on the input and the silver gate electrode as the output.

Due to their unique morphology, characterizing dendrites is not as straightforward as for thin films. If increasing the ratio of EDOT-G4 had been shown to

correlate with a decrease of Young’s modulus¹³⁷, it was not possible to perform the same measurement on freestanding dendrites to obtain indirect insights on their chemical composition. Other techniques had thus to be considered.

VII.1 Estimation of the resistivity of the dendrites

In order to assess the presence of EDOT-G4 in the bulk of the material, the cylindrical model developed by former post-doc Dr Ankush Kumar was used to estimate the conductivity of the dendrites¹⁴⁷. Using this model, the dendrites are considered to be cylindrical which allows to estimate their volume from the pictures taken during growth. Considering that they are homogenous objects, their resistivity can then be computed from empirical as well as theoretical data. The results of this modeling are synthesized in Table 4.1.

EDOT-G4/EDOT Ratio (%)	Resistivity (Ω m)
0	2.05
20	2.00
30	1.54
40	1.62
50	1.83
60	1.63
70	1.11

Table 4.1: Estimated values for the resistivity of the dendrites, using a cylindrical approximation for the determination of the material volume.

From these results, it is hard to either confirm or deny the absence of EDOT-G4 in the grown dendrite. The presence of EDOT-G4 should be correlated with a swelling of the dendrites due to the uptake of water-coordinated cations, thus dedoping the polymer but also increasing the volume of the structure and falsifying the estimations of volume and resistivity. If the computed resistivity is indeed divided by a factor of two, this tendency is not really observed for all the samples, with very similar estimations of resistivity found for 30% to 60% of EDOT-G4. The approximations of the chosen model also leave room for errors. All in all, the variations of the estimated resistivity could come from the model, the stochasticity of the process, the presence of EDOT-G4 in the structure of the dendrites or a combination of all these factors, making it impossible to set on an explanation regarding the set of experiments presented so far. In order to push the investigation a step further and try and verify the presence of EDOT-G4 in the polymer fibers, material characterizations were performed.

VII.2 Time of Flight – Secondary Ion Mass Spectrometry (ToF-SIMS)

ToF-SIMS is a surface analytic technique that uses a focused beam of primary ions sputtering the surface of a sample to produce secondary ions that will be collected and analyzed. The mass/charge ratio of these secondary ions is then used to determine their composition. As this technique is not a central aspect of this thesis, the reader in search of additional information is referred to the work of Vickerman and

Briggs for a more in-depth analysis and understanding of the method²⁴⁴.

Freestanding dendrites (such as the ones discussed all along this chapter) being frail objects, they could not easily be removed from the drop of electrolyte to be analyzed. With the aim of running ToF-SIMS experiments on the dendrites, a solution had to be found to move the fibers without destroying them. To do that, dendrites were grown on a 2D substrate, to which a whole chapter of this manuscript will be dedicated (see DENDRITIC NETWORKS FOR MORPHOLOGY-DEPENDENT COMPUTING HARDWARE). The setup for growth, that is in every aspect similar to the one depicted earlier in Figure 4.1, is detailed in the MATERIALS AND METHODS chapter (see Dendrites on a 2D substrate).

Four dendrites were grown on a 2D substrate, on which the centers of two adjacent electrodes are 100 μm apart, with different concentrations of EDOT-G4 in the growth electrolyte: 0%, 30%, 50% and 70%. The electrical parameters used for the input square waves were the same as for freestanding dendrites, namely 80 Hz, 4 V_p , and a duty cycle of 50%. In addition to the 10 mM of monomer in total, the electrolyte also contains 1 mM of NaPSS and 10 mM of BQ. The resulting dendrites are presented in Figure 4.12.

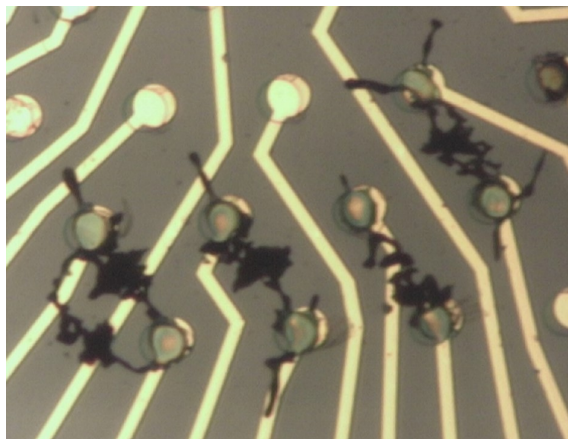


Figure 4.12: Microscopic picture of the dendrites grown on a 2D substrate, in an electrolyte containing (from left to right) 70%, 30%, 0% and 50% of EDOT-G4.

Once again, the addition of EDOT-G4 in the growth electrolyte does not seem to impact the morphology of the dendrites, as it appears that it is visibly impossible to distinguish them in the picture. The conclusion is similar for the conductance and impedance measurements (not shown here), which do not highlight any trend in the properties of the dendrites correlated with the composition of the electrolyte.

This sample was brought to Institut Chevreul* to be analyzed by Dr Nicolas Nuns, using ToF-SIMS. The results of his analysis are presented in the annexes of this manuscript (see Appendix). In a nutshell, no significant difference can be found between the different spectra, and there is no correlation between the changes in the peaks and the amount of EDOT-G4 present in the electrolyte. Here, we note that capturing data by focusing the primary ion beam on the dendrites was not possible,

*<https://chevreul.univ-lille.fr/>

likely due to the cylindrical shape of the dendrites scattering secondary ions away from the detector. The analysis therefore had to be performed on the surface of the electrodes, where material was indeed deposited during growth but to a lesser extent than in the bulk of the dendrite. In conclusion, based on the electrical experiments that were performed in the lab coupled with the ToF-SIMS measurements hereby presented, it would appear that EDOT-G4 is not present in the electropolymerized material.

However, the position of the beam on the dendrites might be an important parameter to take into account. Indeed, Raman spectroscopy performed on PEDOT:PSS dendrites with the help of Dr Michel Daher Mansour suggests that the degree of organization could vary alongside the dendrite.

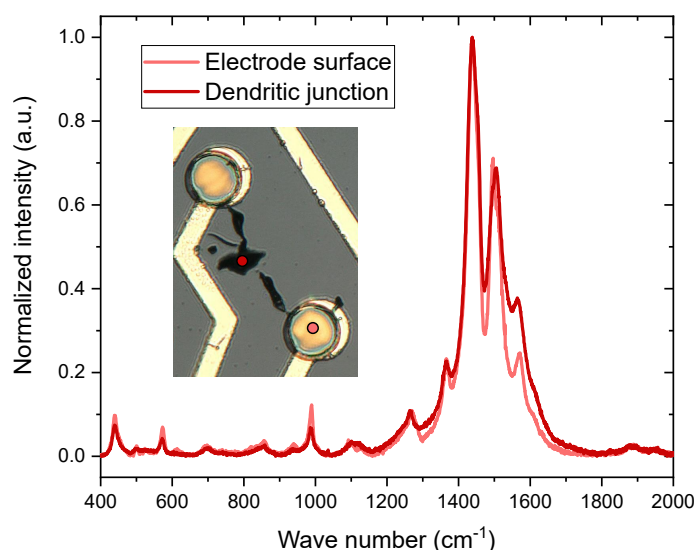


Figure 4.13: Position-dependent degree of organization in PEDOT:PSS dendrites. Raman spectroscopy on a PEDOT:PSS dendrite, both at the surface of a gold electrode and at the junction in the middle of the fiber. The inset picture shows the dendrite on which the measurements were performed, as well as the localization of the beam for both curves.

Qualitatively, Figure 4.13 illustrates that the degree of organization of the dendrite does not seem to be uniform and could be position-dependent, a behavior that is usually not observed in thin films and that demonstrates the difficulties of characterizing such unusual objects. Working with dendritic-like fibers therefore calls for a special attention when designing characterization experiments. In addition, these methods can be destructive, and they require dendrites to be translated to 2D substrates. Yet, it remains unclear how substrate material and the electrodeposited polymer interact during growth, and how that might affect the properties of the dendrites. Nevertheless, growing dendrites on a 2D substrate could enable more characterization techniques to be used, such as AFM. Additionally, since the glycolated chain of EDOT-G4 should present a high affinity with sodium ions¹³⁷, sodium indicators could also help detecting changes in the composition of the electrogenerated dendrites.

VII.3 Other monomers

Two other monomers synthesized in our lab by Dr David Guérin were considered to be electropolymerized, a dithiophene and a thienothiophene molecule.

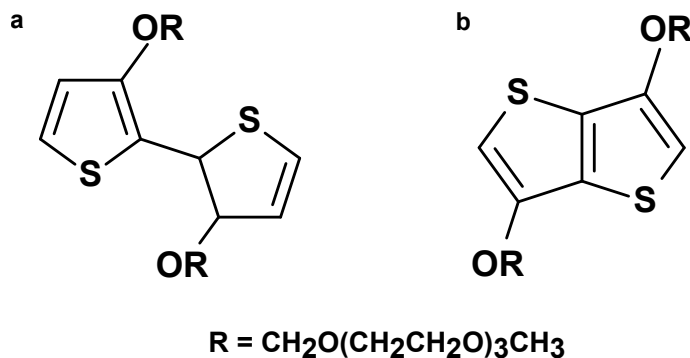


Figure 4.14: EDOT-derivatives. Chemical structure of the a) dithiophene and b) thienothiophene molecule. The glycolated side-chain is the same as for EDOT-G4.

However, due to low water solubility, an aqueous electrolyte cannot support the electropolymerization reaction. As DMSO is a biocompatible solvent and has a large electrochemical window, it was examined as an alternative to deionized water, but the supporting electrolyte NaPSS could not be dissolved in this solvent. Different mixtures of DIW and DMSO were tested without much success.

VIII Conclusion

In this chapter, we explored the influence of the composition of an electrochemically active aqueous electrolyte over the behavior of electropolymerized PEDOT:PSS dendrites, investigating parameters such as the concentration of salt, the amount of oxidizing agent, the concentration of monomer and the chemical nature of this latter. The findings reveal that electropolymerization of polymer fibers can happen in a vast range of conditions, and that the composition of the electrolyte actually presents little to no correlation with the electrical performances of the PEDOT:PSS dendrites thereby formed nor with their morphology, contrary to the parameters of the waveform generator as it had previously been established thoroughly as part of the thesis work of Kamila Janzakova²⁴⁵. Nevertheless, this work remains important for the global comprehension of the electropolymerization approach, as it highlights that some crucial criterions must be met for growth to be triggered. It was shown notably that salt concentration had a non-negligible influence over the voltage drops in the system, and consequently that too little or too much of it could prevent nucleation. We also demonstrated that the presence of BQ as an oxidizing agent would help with the growth, and that lowering the concentration of monomer would constrain the other parameters of the system, particularly the applied voltage or the distance between the electrodes working as anode and cathode. Given the already high stochasticity of the electropolymerization process, these results might help to understand how to reduce variability and to better control growth mechanisms.

This work therefore paves the way for the development of bottom-up strategies based on brain-inspired hardware, positioning conductive polymer dendrites as a potential candidate for the implantation of platforms with the ability to evolve and create new connections on the flight, not unlike neurons and synapses. These results emphasize the importance of the environment when it comes to growing polymer fibers via electropolymerization: the ionic strength of PBS mimicking that of the extracellular fluid, it clearly appears that the electropolymerization of polymer dendrites *in situ* in biologically-relevant medium represents a technological challenge, although the idea might not be as farfetched as it could seem at first glance^{246–248}.

With that framework in mind, the following chapters of this manuscript will focus on the exploration of the properties of electropolymerized PEDOT:PSS dendrites and dendritic networks, demonstrating the potential of such systems in the realm of brain-inspired computing.

5

FUNCTIONAL PLASTICITY IN POLYMER DENDRITES

I Introduction

As we saw in the first chapter of this manuscript, brain-inspired computing offers a new paradigm in which neuromorphic organic devices appear to have a part to play due to the new opportunities provided by organic materials, which grant numerous advantages over traditional silicon-based technology, especially their ability to implement synaptic-like functions easily. In addition, last chapter brought forward the versatility of the electropolymerization process by exploring the electrochemical conditions within which growth can happen. In particular, polymer dendritic network possesses an important property: they can be grown on the spot, using a bottom-up fabrication approach^{148–151,156,249}. In practice, this means that the network could adapt to the computing task by scaling up its number of inputs and outputs *in operando*, yielding a high degree of parallelism in information collection and processing.

Single electrogenerated polymer dendrites have already proven to work as OECT¹⁴⁷, showing synaptic-like functions such as short-term memory that depend on their unusual morphology. The computational interest of dendritic networks has also been demonstrated in the context of reservoir computing, highlighting the ability of such networks to project input information into high-dimensional planes²²³. Provided that the recording of electrophysiological activity with OECTs has already been reported numerous times^{194,197,198}, integrating structural plasticity, computing and sensing ability into a brain-inspired polymer dendrite network is a very interesting prospect.

In this chapter, we will therefore explore the ability of a PEDOT:PSS dendrite to perform the detection and classification of voltage bursts in Phosphate Buffered Saline (PBS), in the perspective of neurosensing.

Parts of the results presented in this chapter have been published in [250].

II Experimental setup for freestanding dendrite growth

Freestanding dendrites were grown in the usual growth electrolyte as described in the MATERIALS AND METHODS chapter of this manuscript (240 μm , 80 Hz, 5 V_p). Dendrites were grown between two gold wires separated by a distance of 240 μm . Following growth, a volume of twice concentrated PBS (2*PBS), equivalent to the one used for growth, was gently added to the drop of growth electrolyte in order to perform electrical characterizations of the dendrite. This way, the final concentration of PBS in the solution would reach that of normal PBS (1*PBS). Because of the presence of electroactive species in the solution the characterizations were performed keeping the maximum potential difference in the system below 200 mV in absolute value. Otherwise, electropolymerization could be triggered again, adding material to the already existing dendrite and ultimately modifying the electrical properties of the system while performing the measurements.

III Steady-state characterization of freestanding dendrites

As it was demonstrated elsewhere by our group¹⁴⁷, dendritic interconnects possess transistor-like behavior. In this previous study, the gate was an Ag/AgCl electrode much bigger than the dendrite itself. Yet, working towards the integration of polymer fibers would require smaller gate electrodes. A 25 μm diameter silver wire was thus introduced as a gate, and the behavior of the system was assessed after the growth of a dendritic object. The transfer characteristics of the dendrite are presented in Figure 5.1c.

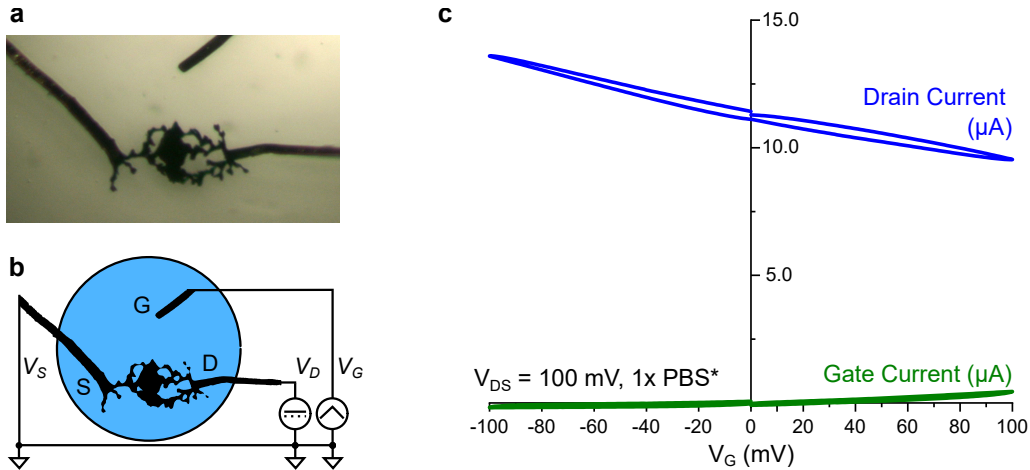


Figure 5.1: DC characterization of a freestanding dendrite. **a)** Microscopic picture of the dendrite after growth completion, showing the two gold wires (left and right) as well as the silver wire (top) used as the gate. **b)** Experimental setup used for electrical characterizations. A small bias is applied between the source and drain electrodes ($V_{DS} = 100$ mV) while the voltage on the gate is swept between -100 mV and 100 mV, as plotted in **c)** Transfer characteristics of the dendrite. Adapted from [250].

The resulting behavior is the one expected for a p-type semiconductor: applying a gate voltage leads to the modulation of the drain current correlated with the modification of cationic concentration in the dendrite, therefore doping/dedoping the material to respect electroneutrality. The conductance of the dendrite during these experiments was rather stable with close to no hysteresis observed for a voltage sweep of ± 20 mV/s. The drain current displays a linear relationship with the gate voltage, with a transconductance of about 20 μ S throughout the whole range of voltage. This observation allows us to infer that the modulation both for $V_G > 0$ V and $V_G < 0$ V is caused by the same physical process, most probably the doping/dedoping of the system due to the modulation of cationic concentration in the vicinity of the dendrite^{191,251}. The effect of anions is not to be discarded, however since the negatively charged PSS⁻ might prevent them from entering the polymer²⁵², their surface effect is likely to be less important than the accumulation of cations within the bulk of the material.

The contribution of the gate current to the modulation of the drain current (around 400 nA at $V_G = 100$ mV, which is about 20% of the total modulation of 2 μ A) points to a small contribution of the faradaic current related to the redox reaction at the silver electrode, as expected from the previous experiments. For a negative gate polarization, this faradaic current is no longer visible, as the silver wire does not undergo oxidation since it is not the highest potential of the system. The observed modulation therefore results from the doping of the material alone, as would be expected for a transistor, thus demonstrating that a dendritic interconnect coupled with a small Ag wire can indeed serve as a transistor.

III.1 Choice of gate electrode

As the choice of gate electrode is important when it comes to the operation of an OECT²⁴², the influence of the gate material in our three electrode system was assessed. Two different materials were considered, gold and silver. Both stands for a different class of electrodes, namely ideally polarizable and ideally non-polarizable (for more details, the reader is redirected to MATERIALS AND METHODS) The assumption was that gold, as a polarizable material, would not be a very efficient gate material, as the resulting double layer capacitance would imply a drop of voltage at the gate-electrolyte interface rather than at the channel-electrolyte interface. Conversely, this effect should be downplayed by the redox behavior of silver.

Figure 5.2c shows the impedimetric measurements performed on a three-wire system such as the one represented in Figure 5.2b. These measures highlight the contribution of the silver wire oxidation under a negative bias. Indeed, unless otherwise specifically mentioned, the experiments were performed in presence of all electroactive species required for electropolymerization, including the oxidizing agent BQ. When the silver electrode has the highest potential in the system, it then starts to oxidize. A corresponding plateau therefore appears at low frequencies, representing the charge-transfer resistance at the gate-electrolyte interface (of about 100 k Ω , see the down-facing filled gray triangles in Figure 5.2c). This behavior is not observed when the same experiment is reproduced in the absence of the electroactive species of the growth solution (in PBS only), nor when the silver wire is not used as the

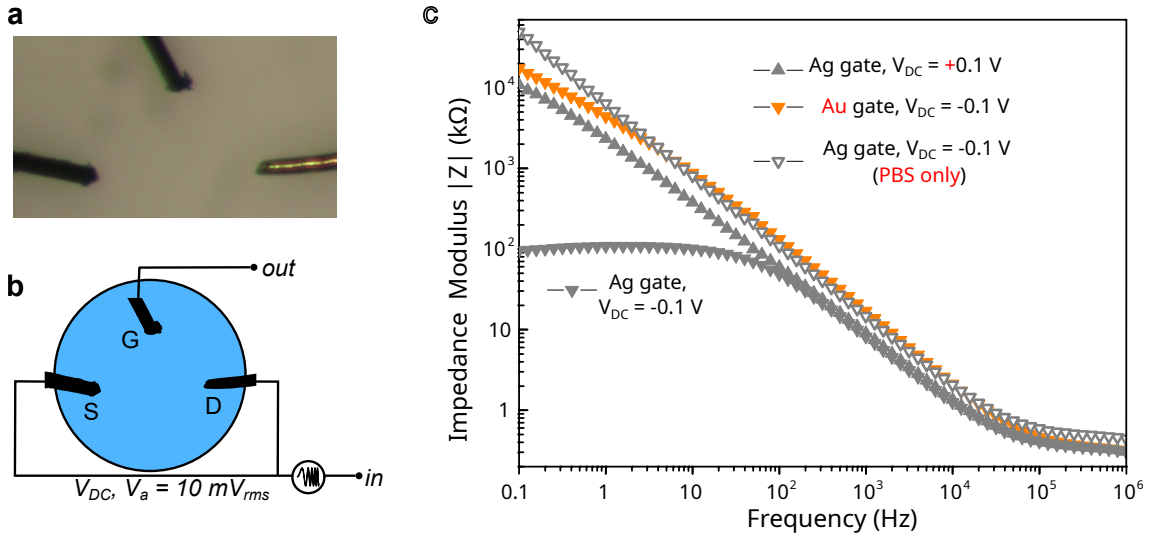


Figure 5.2: Influence of the gate material on the impedance measurement. a) Microscopic picture of the setup. b) Experimental setup. The source and drain electrodes are both $25 \mu\text{m}$ diameter gold wires, and the gate electrode is either a $25 \mu\text{m}$ diameter gold or silver wire. c) Electrochemical impedance measurement in different conditions, with a gold wire as the gate (yellow triangles), or with a silver wire as the gate (gray triangles). The measurements were either performed in an electrolyte containing the growth solution and PBS (filled triangles), or in PBS only (empty triangles). Adapted from [250].

cathode of the system. When the gold wire is used as the gate, no charge-transfer resistance is to be observed whatever the DC bias, as expected.

Consequently, the impedance of the dendrite has to be lower than the plateau discussed previously to be able to analyze the results obtained by EIS, otherwise the faradaic current will take over and hide the ohmic resistance of the object.

III.2 Influence of PBS in the electrolytic solution

Once the gate material was set, the contribution of a high ionic strength electrolyte such as Phosphate Buffered Saline (PBS) to the dynamics of a dendritic transistor with a silver gate electrode was investigated. In that purpose a dendrite was grown, such as the one shown in Figure 5.3a. Impedance measurements were run in the growth electrolyte only, and in the growth electrolyte with additional PBS. Adding PBS to the electrolytic solution served two different purposes. First, mixed with the electrolyte used to grow the dendrite, it allows the solution to become more conductive. By adding PBS, large quantities of ions are introduced into the system and thus it lowers the plateau observed at high frequencies that is related to the impedance of the electrolyte itself. This was necessary to ease the lecture and the analysis of the results obtained by EIS, as otherwise the curves did not span more than one decade. Second, PBS was also chosen to mimic biological medium, as going towards *in vitro* electrophysiology was a motivation from the beginning of this thesis, as it will be developed in a following chapter.

Let us note that the experimental setup is different to the one that was presented in the previous section of this manuscript, and described in Figure 5.3b. Whatever

the DC bias, Figure 5.3c shows that, in the absence of PBS, the electrolyte is less conductive than the dendrite. Therefore, the measurement reads the resistance of the dendrite itself (filled triangles). On the other hand, when PBS is added to the system, the conductivity of the solution is greatly enhanced due to the high ionic strength of PBS.

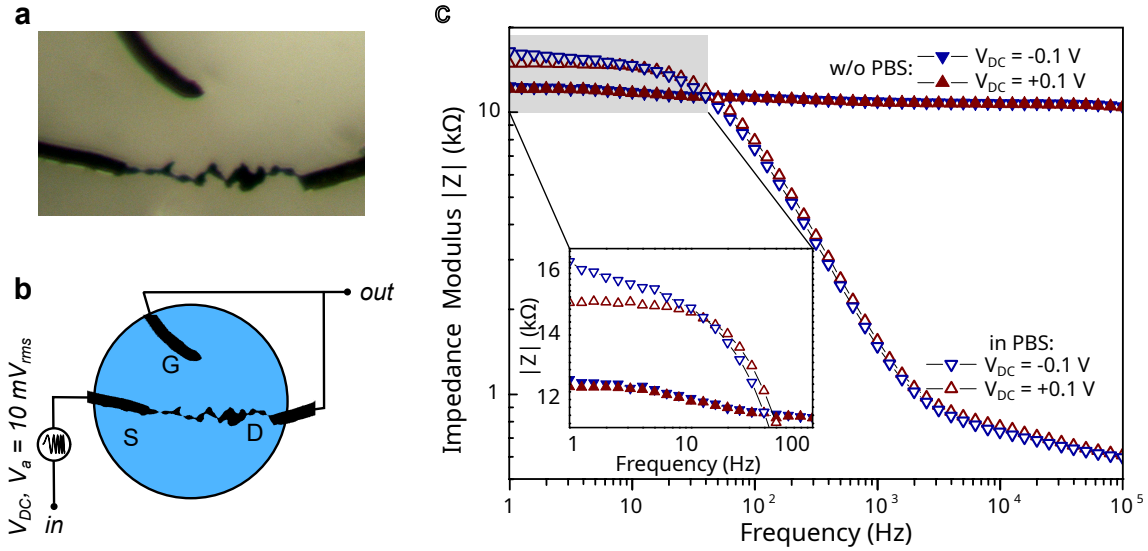


Figure 5.3: Influence of the composition of the electrolyte on electrochemical impedance measurements. a) Microscopic picture of the setup. The dendrite is approximately 240 μm long. b) Experimental setup. The drain and gate electrodes are shorted to the output of the impedimetric instrument and the source is connected to the input. c) Electrochemical impedance measurements in growth electrolyte, with (empty triangles) or without (filled triangles) PBS. In both cases, a small bias voltage ($|V_{DC}| = 100 \text{ mV}$) was applied between the input and the output of the system, either positive (red triangles) or negative (blue triangles). Inset: zoom on the low frequencies of the spectrum. Adapted from [250].

III.3 Electrochemical impedance spectroscopy

As mentioned before, the addition of PBS to the growth solution leads to an increase of the conductivity of the electrolyte. Consequently, the dendrite becomes more resistive than the electrolyte, and two different operational regimes appear when characterizing the system in a diode-connected configuration, with the drain and gate electrodes shorted (see Figure 5.4a). At low frequencies, the measurement reveals the impedance of the dendrite, and thus its doping state, whereas at high frequencies the electrical current is transmitted through the electrolyte (Figure 5.4b)²⁵³. Therefore, the first conclusion that can be drawn from this experiment is that, in physiological conditions, the capacitive response of a polymer dendrite will greatly affect its dynamics.

The impedimetric measurement of the system at low frequencies shows two distinct behaviors, depending on the polarization of the gate. For a positive bias ($V_{DC} = 100 \text{ mV}$), the difference of potential between the source and the gate is negative ($V_{GS} = -100 \text{ mV}$) and so the dendrite behaves as a conductive wire with a resistance

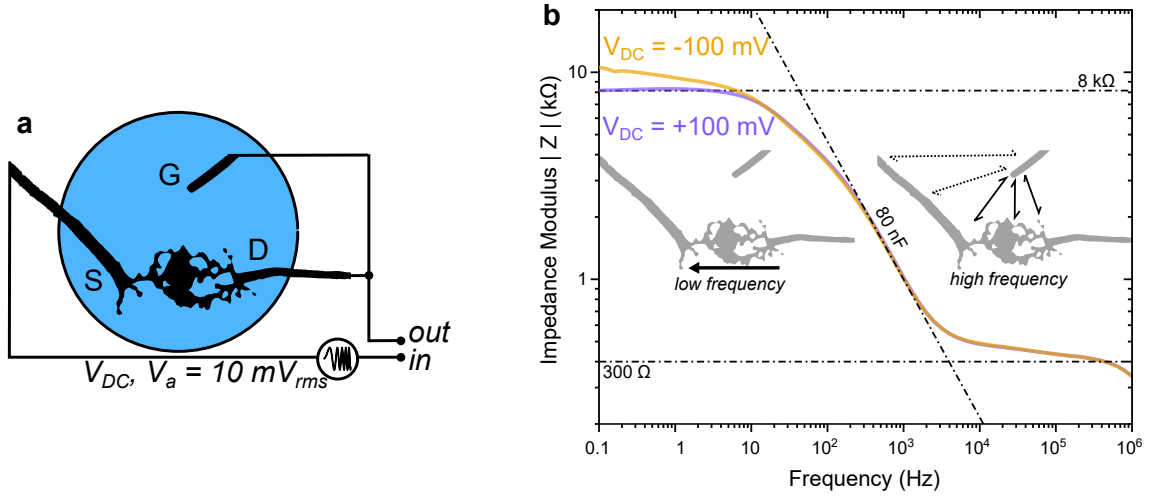


Figure 5.4: Electrochemical impedance measurement of a freestanding dendrite (see Figure 5.1a for the picture of the dendrite). **a)** Experimental setup for the EIS measurement. The drain and gate electrodes are shorted to the output of the measurement instrument, while a constant voltage and a small perturbation is applied to the source of the dendrite. **b)** EIS measurements of the conducting polymer dendrites for two different DC biases, promoting either electrochemical doping ($V_{DC} = 100$ mV) or dedoping ($V_{DC} = -100$ mV). Adapted from [250].

value of about $8\text{ k}\Omega$. However, when the bias is negative (and so $V_{GS} > 0$ V), the dendrite is electrochemically dedoped around the source electrode, and steady state is not reached yet at 100 mHz.

These results seem to indicate that two different temporal dynamics should be expected during the operation of a dendritic OECT, depending on the polarization of the incoming signal as well as the polarization of the device itself. In the event of an incoming signal promoting anionic displacement towards the channel of the transistor, the doping of the material might be much quicker than in the case of an incoming cationic wave promoting dedoping. In other words, the charge of the dendrite with cations appears to be slower than with anions. Two reasons could explain this different behavior:

- The high hydrophobicity of PEDOT domains might be difficult to access for cations, and once inside it could be harder to leave the bulk of the material²⁵⁴.
- The morphology of the dendrites could be an important factor, as ‘dead-end’ branches most definitely participate in the capacitance of the device, but not in conduction. A larger amount of material would necessitate more time to absorb all cations.

In the following section, the behavior of a dendritic transistor will be assessed in the temporal domain.

IV Response to voltage bursts

In the brain, information is encoded and transmitted between neurons through action potentials. However, a single spike does not in itself contain much information.

Bursts, or spike trains, are a much more reliant information carrier, and are one of the ways data is modulated in the brain^{255,256}.

IV.1 Spike train generation

In this section of the manuscript, we will explore the ability of dendritic transistors to perform the detection and classification of voltage bursts. To assess the behavior of polymer dendrites when stimulated with voltage pulses at the gate electrode, different configurations were evaluated using the experimental setup presented in Figure 5.5.

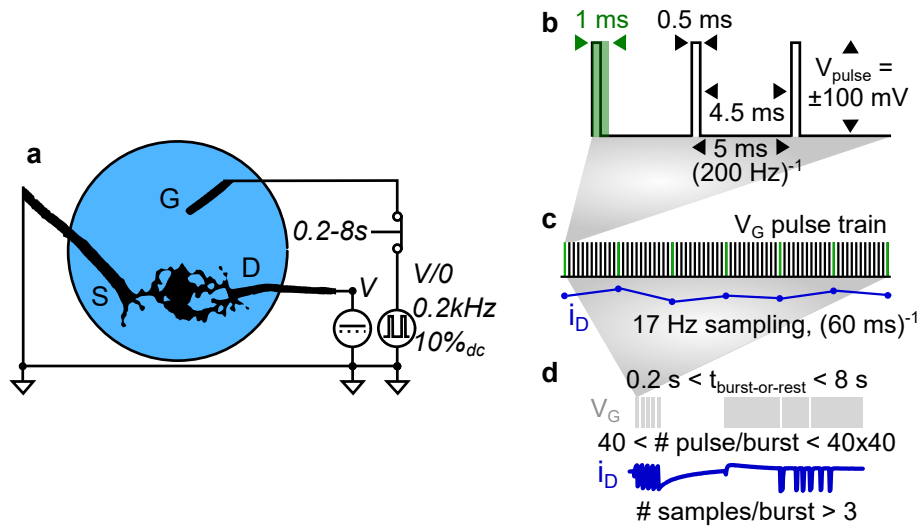


Figure 5.5: Experimental setup for the stimulation of a dendritic OECT with spike trains. **a)** Experimental setup, showing that a continuous bias is applied between the source and drain of the dendrite. Meanwhile, voltage spikes of ± 100 mV are sent to the gate. A mechanical switch (push button) is used to manually connect and disconnect the gate. When the switch is open, the gate is left floating. **b)** The spikes are sent at a frequency of 200 Hz, with an amplitude of ± 100 mV and a duty cycle of 10%. The integration time of the measure lasts 1 ms (green highlight). **c)** The sampling is synchronized with the input sequence at a rate of one sample every 12 spikes, so that the sampling frequency is about 17 Hz. **d)** The information is modulated by the duration of the burst, controlled manually with the push button. The shortest burst (0.2 second) will contain 40 spikes, while the longest one of 8 seconds will contain 1600 spikes. In terms of sampling, it means that at least 3 samples are recorded per burst.

The characterization was performed during the operation of the dendrite as an active device, thus biasing its terminals. A continuous series of spikes was sent to the gate electrode of the system at a frequency of 200 Hz (the details of the waveform are presented in Figure 5.5b). The input signal was constituted by a series of positive or negative unipolar voltage pulses with an amplitude of 100 mV with respect to the resting potential of the gate, so as to promote either cationic or anionic displacement based on the polarization of the dendrite. Bursts were simulated by interrupting signal transmission with a mechanical gate. Although a laboratory model, the parameters of the input waveform were chosen to imitate the electrophysiology of a brain.

IV.2 Short term plasticity

Figure 5.6 presents the recorded source and drain currents for the four different cases that were attempted. The traces immediately show that the input information sent on the gate electrode is projected differently on both terminals of the dendrite, an opportunity which will enable us to retrieve both the polarity as well as the temporality of the bursts (see Feature extraction with polymer dendrite). As it was observed with the impedance spectroscopy measurements, anions and cations appear to obey to different kinetics based on their accumulation profiles. Their temporal dynamics should therefore also be distinguishable, depending on their ability to promote the electrochemical doping/dedoping of the dendrite.

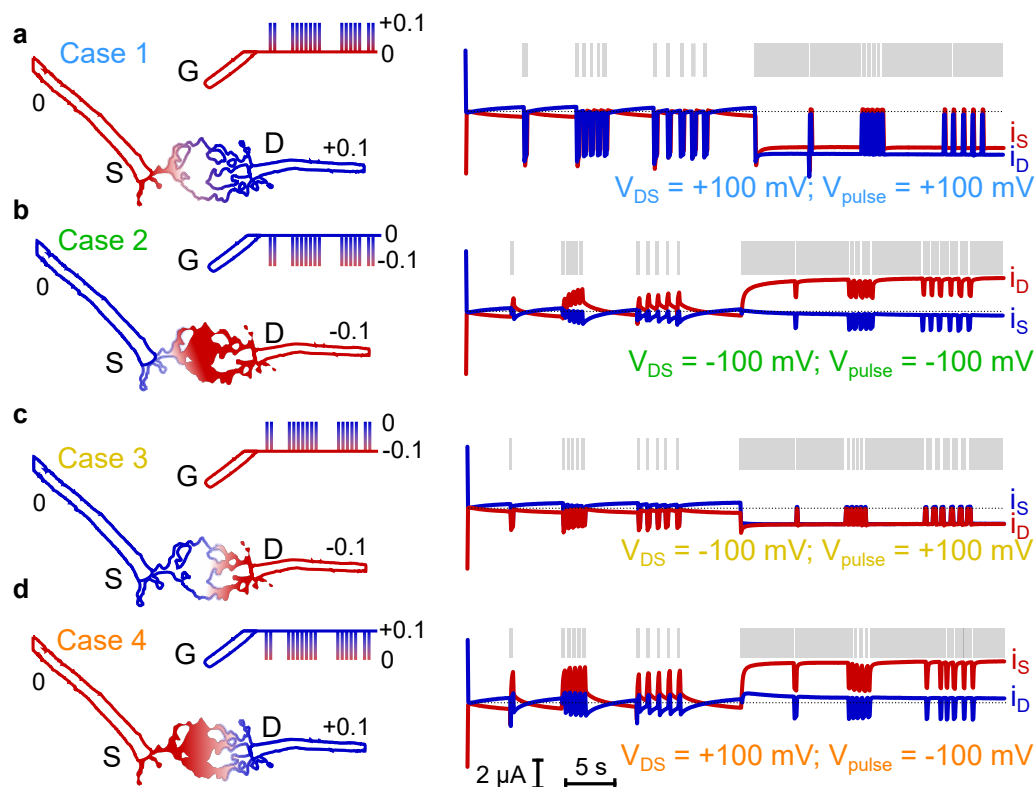


Figure 5.6: Spike train detection with dendrites. The incoming spike trains either increase (a and c) or decrease (b and d) the potential of the gate electrode. Each test was performed while applying a positive (a and d) or a negative (b and c) voltage bias across the dendrite. The drain and source currents are centered on the initial value of the current. The activation of the push button is marked by the grey bars above the traces. For panels a-d, the color represents ionic accumulation, within the bulk of the material for the cations (in red) or on the surface of the dendrite for the anions (in blue). This representation assumes that cations and anions have distinct accumulation profiles within and at the surface of the dendrite. Adapted from [250].

Experimentally, depressing spikes (cases 1 and 3 in Figure 5.6) elicited a faster response of the system than potentiating spikes (cases 2 and 4). Whatever the duration of the burst applied at the gate, the modulation of both the source and drain current reached its maximum after the first train of spikes in the case of depressing voltage pulses. Conversely, in the case of potentiating spikes, a slow build-up was observed in the current modulation. In this case, the maximum modulation was only reached after either a high enough number of bursts, or a certain duration of

the burst. This behavior is best observed in Figure 5.7.

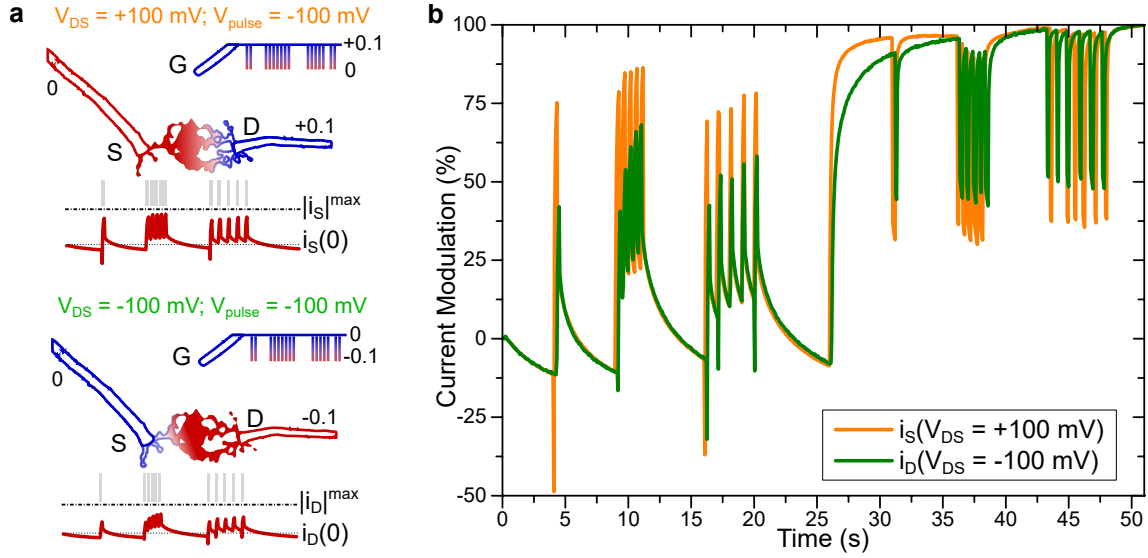


Figure 5.7: Short term plasticity during the operation of the dendrite as an OECT. **a)** Zoom on the first half of the input sequence for the negative spikes (cases 2 and 4 in Figure 5.6), where the slow current build-up can be appreciated. **b)** Current modulation as a function of time, where the sequenced pulse facilitation can be appreciated. Adapted from [250].

The current modulation observed for potentiating spikes is what was expected from the previously discussed results. Indeed, in these last two cases, the polarity of the spikes promotes cation movement toward the gate electrode, and out of the polymer dendrite. As explained earlier, the diffusion of cations from the bulk of the polymer is expected to be slower than the movement of anions, which would tend to accumulate on the surface of the material.

Figure 5.7b shows the slow build-up of current in the case of potentiating spikes (see Figure 5.7a). The distinct accumulation profiles of the different ionic species present in the solution lead to sequenced pulse facilitation, as a train of presynaptic action potentials (the gate signal) results in an increase in post-synaptic activity (source and drain current). Moreover, the influence of the morphology on the response of the dendrite to the stimuli can be clearly appreciated. When V_{DS} is positive, electrochemical dedoping is promoted by the attraction of cations towards the source electrode, as represented in Figure 5.7a. However, from the picture of the dendrite inserted in Figure 5.1a, we can qualitatively assert that the specific capacitance of the system should be more important in the vicinity of the drain electrode than close to the source electrode, due to polymer branches that do not participate in conduction but still increase local capacitance. Although the path followed by electrons does not depend on the polarization of the system (no diode effect in Figure 5.1), the asymmetric morphology of the dendrite might induce different capacitances and time constants locally. As the modulation of current is greater when the dedoped front is located towards the source electrode, we conclude that the specific capacitance in the vicinity of the drain electrode might be higher than that in the vicinity of the source electrode. Consequently, the modulation of current evoked by spike trains is more important when the dedoped front is located in vicin-

ity of the source electrode, compared to when the vicinity of the drain electrode is dedoped ($V_{DS} = -100$ mV).

This asymmetrical behavior is related to the morphology of the dendrites, rather than only the voltage bias, as the four cases depicted in Figure 5.7 correspond to the electrochemical dedoping of one side of the dendrites. Since the morphology of the dendrites can be controlled during growth^{151,230}, these results lead us to believe there exists a relationship between form and function in dendritic systems. This idea will be further stressed in the next chapter of this manuscript, and is illustrated in the following section of this chapter.

IV.3 Feature extraction with polymer dendrite

The morphological asymmetry of the dendrites allows feature extraction during the operation of the object as an active device. Using both terminals individually, the polarity of the spikes as well as their temporality can be retrieved. Due to the distinct sensitivities of both sides of the object, related to the asymmetry in morphology, different pieces of information are recorded from the drain and source electrodes.

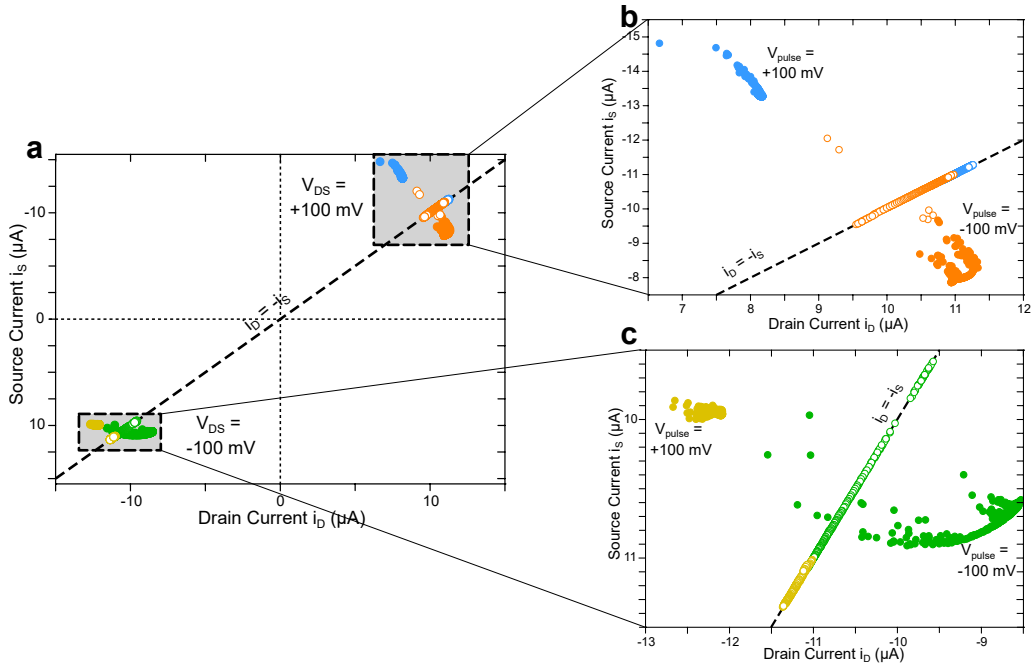


Figure 5.8: Analysis of the output of the system when stimulated with spike trains. a) Source current as a function of drain current. The traces correspond to those presented in Figure 5.6, with the same color coding. b) Zoomed-in view for the cases when $V_{DS} = 100$ mV. c) Zoomed-in view for the cases when $V_{DS} = -100$ mV. Adapted from [250].

Because of the polarization of the device, a constant current is flowing through the dendrite, in such a way that $i_S = -i_D$ (empty circles in Figure 5.8) as the charges are evacuated symmetrically on both electrodes in the absence of perturbation in the system. However, when spike trains are sent through the gate of the electrochemical transistor, the drain and source currents show different patterns, depending on the polarization of both the device and the input spikes.

Upon the application of voltage bursts at the gate, clusters appear when looking at the asymmetry in current, as shown in Figure 5.8. When the drain potential is equal to that of the pulses, the absolute value of the source current is greater than that of the drain current ($V_{DS} = V_{pulses} \implies |i_S| > |i_D|$). This phenomenon is clearly appreciated for $V_{DS} = V_{pulses} = 100$ mV (blue cluster in Figure 5.8). In this situation, the evacuation of the charges does not happen symmetrically on both sides of the dendrites, as the application of a gate voltage creates an imbalance in the system. The largest potential difference is measured between the source and the gate electrodes when the latter is spiking, with $|V_{GS}| = 100$ mV and $V_{GD} = 0$ V. As a consequence, electrochemical charges preferentially flow towards the source terminal.

Conversely, when the pulses are polarized oppositely to the drain-source voltage, the magnitude of the drain current exceeds that of the source current ($V_{DS} = -V_{pulses} \implies |i_D| > |i_S|$). The voltage difference is in this case the most important between the drain and the gate, since $|V_{DG}| = 200$ mV. The flow of ions coming from the gate is therefore directed toward the drain. In Figure 5.8, these situations are represented by the orange and yellow circles.

In addition, no symmetry is to be observed between the top-right (Figure 5.8b) and bottom-left (Figure 5.8c) quadrants in Figure 5.8a. This indicates that the response of the system differs depending on the polarization of the dendrite. This effect is once again due to the morphology of the objects, which capacitance varies locally and implies different charge accumulation profiles. Consequently, the source and drain electrodes collect different pieces of information, which give rise to different projection of the same input pattern. This is what makes it possible for dendritic OECTs to perform temporal discrimination.

IV.4 *In materio* classification of voltage bursts

Figure 5.9 highlights the monotonic relationship that exists between the source/drain current ratio and the gate current ($-i_G = i_D + i_S$). The first observation that can be made looking at this plot is that, for a given V_{DS} , the current asymmetry is not proportional to the gate activity (Figure 5.9a), showing once again that the response of the system to an input is not the opposite of the response for the opposite input.

The second observation is that, for a fixed V_{DS} , the gate current varies monotonically with the current asymmetry. In other words, setting a threshold on the i_S/i_D ratio comes to detecting current flowing from the gate, as data separation would be similar. Setting a threshold on a dimensionless value offers a great advantage for dendritic OECTs over usual sensors. In electrophysiology for example, spike detection necessitates computing the standard deviation of the recorded data to establish a threshold. This computation cost disappears using dendritic transistors, thanks to the device ability to perform *in materio* computing tasks through the different ionic accumulation profiles.

Figure 5.9c illustrates that applying a threshold on current asymmetry allows the detection of most of the bursting activity when the pulses are polarized negatively ($V_{pulses} = -100$ mV). This observation does not hold for positive pulses ($V_{pulses} = 100$ mV).

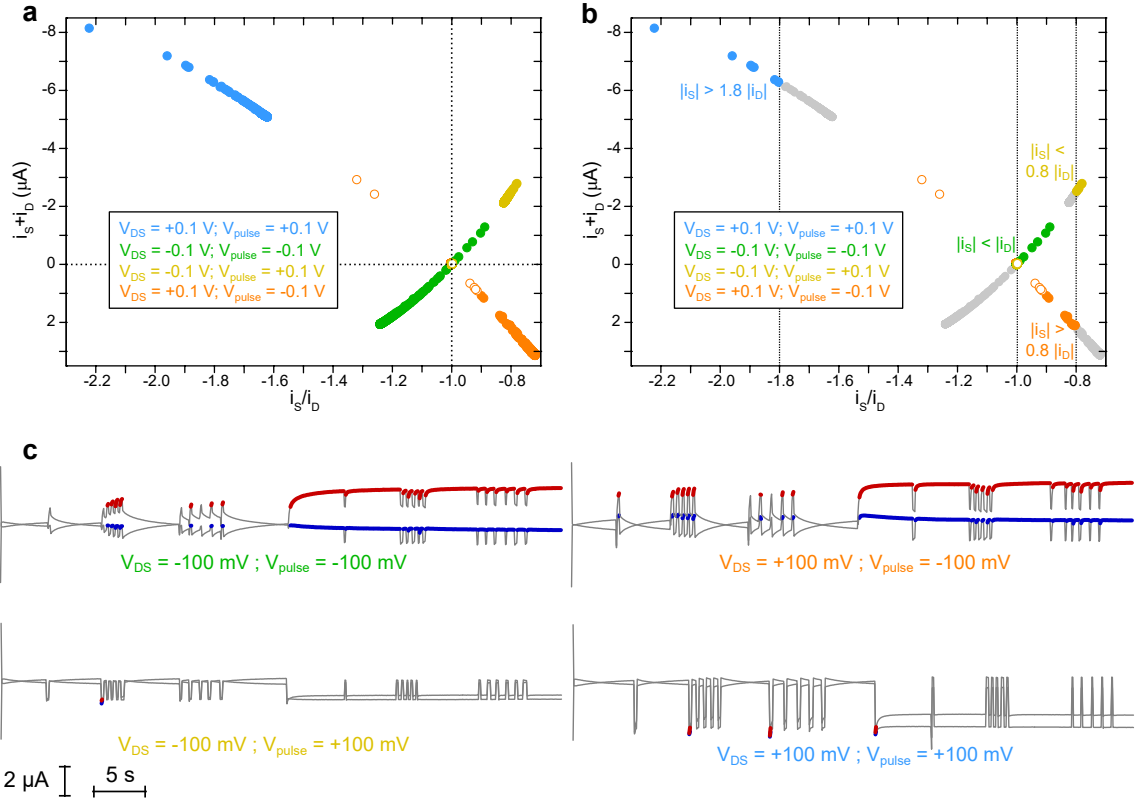


Figure 5.9: Relationship between current asymmetry and gate activity. a) For a fixed V_{DS} , a monotonical relationship between gate activity and source/drain ratio is observed. b) A threshold on current asymmetry is therefore similar to thresholding the gate current, only the data is dimensionless. c) Current-Time traces for the four different cases presented in Figure 5.6. The red and blue dots highlight data crossing the thresholds defined in b). Adapted from [250].

mV), since data separability becomes in this case more difficult. For $V_{DS} = -100$ mV and $V_{pulses} = 100$ mV, the cluster is so tightly packed that it becomes impossible to use a threshold to perform spike detection (yellow circles in Figure 5.9). Nevertheless, when $V_{DS} = V_{pulses} = 100$ mV, thresholding the current asymmetry allows the detection of the first voltage burst of a series after a long enough resting period (approximately 5 seconds in this example). This information could not be obtained by monitoring individually the source and drain currents.

These differences observed in the case of opposite polarizations are attributed to the distinct ionic accumulation profiles in the material. Since cations are expected to be slower to charge and discharge the dendrite, they allow sequenced-pulse facilitation at the second scale, and thus potentiating spikes elicit a richer temporal response of the system, increasing data separability.

V Conclusion

In this chapter, I have established that PEDOT:PSS dendrites behave like electrochemical transistors with a silver wire as the gate in physiologically-plausible environments, where they are able to detect and discriminate voltage events. Thanks to the specific ion dynamics evidenced in this work but also elsewhere in the literature²⁵², dendritic OECTs implement short-term plasticity in the form of sequenced-paired facilitation.

The main idea to take away is that the unusual morphology of these objects provides them with a behavior that could not be found in other devices, such as traditional planar OECTs. Indeed, most of the properties exploited in this chapter of the manuscript arise from the asymmetric morphology of the dendrites, which in turn creates an uneven distribution of the capacitance over the whole length of the fibers. As such, this morphological asymmetry is a key component for the implementation of an electrochemical classifier based on a dendritic OECT, and more generally a unique feature of electrogenerated polymer dendrites, as it will be further developed in the next chapter.

6

DENDRITIC NETWORKS FOR MORPHOLOGY-DEPENDENT COMPUTING HARDWARE

The previous chapter established the potential of polymer dendrites in the context of neural sensing and *in material* signal classification. In particular, it was highlighted that the computing power of electrogenerated polymer dendrites comes from their unique morphology, that could not be obtained via traditional fabrication routes. In the present chapter, we will keep exploring the computational abilities of PEDOT:PSS dendrites in the context of brain-inspired computing. I will demonstrate that, when grown into a network, cross-talking between fibers leads to a complex interplay between structure and function that gives rise to a singular morphology-dependent behavior of the network that opens up the way for unique opportunities. We will first investigate the origin of the nonlinearities in the system, before leveraging them to implement computing functions and perform *in memory* computing.

Perhaps more important than the applications demonstrated in the pages to come, this chapter is built around the idea that variability is a powerful computational tool that one should try to tame in order to enter a new computing paradigm in which opportunities emerge from small differences and irregularities between devices, instead of the usually sought-after uniformity.

I Introduction

One of the main drawbacks of organic electronics has always been its instability and variability, in part due to the phase changes associated with the low transition temperatures of soft matter. Although low-cost, the commercialization of organic electronic materials is still minimal because of operational and long-term ambient stability concerns^{257,258}. Yet, this inherent variability of soft matter could turn beneficial. As we saw in the introductory chapter of this manuscript, electropolymerization has been exploited in organic electrochemical transistors (OECTs) to generate structurally different active materials on lithographically patterned devices in order to enrich the dynamics of a reservoir, thus facilitating the classification of dynamic information¹⁴⁴. This was possible because each OECT of the array was unique, therefore projecting the information on a unique wider dimensional space. This article demonstrated for the first time that variability could enhance the computational potential of a transistor array. More recently, networks of electropolymerized polymer fibers have been shown to implement Hebbian learning^{221,222} and Boolean

logic²²⁴, as well as biosignal classification through the reservoir computing framework²²³. These dendritic systems therefore appear to be interesting candidates for the implementation of an unconventional computing framework, capitalizing on the complex dynamics at work within these networks of artificial dendrites.

However, to this day, organic devices still do not exploit the topological diversity offered by the electropolymerization method. The rich structural variety provided by this bottom-up approach could be taken advantage of to perform *in materio* computing based on the unique behavior of each device. Therefore, electropolymerization is not only a cost-effective alternative to otherwise expensive fabrication techniques, but also a whole new avenue for the conception of a class of a neuro-morphic hardware with morphology-dependent functions.

In this chapter, we will explore the possibilities offered by electropolymerized dendritic networks from a brain-inspired computing perspective by physically implementing hardware-dependent functions with a unique structure-operation relationship. I will characterize and exploit the topological richness of electrogenerated conducting polymer dendrites for *in liquido* computation. In particular, electrochemical self-gating of single dendrites and inter-gating effects cause voltage-to-current nonlinear transformations that can be used to our advantage to process information. I will show that these systems present an inherent form of memory that can be electrochemically programmed by specifically organizing the ion space charges in the polymer network. Finally, I will show that such networks can be used as an electrochemical classifier, where spatiotemporal projections of nonlinearly separable input patterns can be discriminated thanks to the higher dimensional projection inherent to the dendritic network complexity and exploiting *in materio* computational resources.

II Nonlinear behavior of physically connected dendritic networks

Previous studies had brought to light that conducting polymer dendrites behave like OECTs¹⁴⁷: applying a positive voltage to the gate electrode of the system leads cations into the bulk of the material, thus dedoping the PEDOT:PSS channel. So far, these studies were mostly limited to one electrochemical transistor, with a continuous dendritic channel grown between two electrodes. Cucchi and coworkers went a step further and took advantage of the nonlinearities observed in networks of polymer fibers to implement reservoir computing²²³. In their implementation, the authors used feedback connections to increase the nonlinearity of the spatiotemporal signal projection, thus requiring additional circuitry overhead. Here, nonlinearities are induced uniquely by the complexity of network topologies, and the resulting auto-gating and inter-gating effects are mediated through the common electrolyte.

II.1 Self-gating effect

Within a network of physically connected polymer fibers immersed in an aqueous electrolyte, the distribution of potentials plays a key role, influencing the behavior

of the system and the cross-talking between the different dendrites of the network. Figure 6.1 presents the nonlinear electrical behavior of a Y-shaped dendritic system grown between four gold electrodes of a typical MEA (see MATERIALS AND METHODS), operated in PBS. Figure 6.1b shows the asymmetric behavior of the Y-shaped network when pinning the central electrode of the dendritic system to the ground while applying a voltage sweep between -0.9 V and 0.9 V on the left electrode of the network. When the input voltage is negative, cations close the channel of the dendritic transistor. As a single input channel has a residual volume compared to the full dendritic system, the total capacitance of the dendrites that do not participate in the conduction of current is superior to that of the conducting channel. This results in a self-gating effect, with cations accumulating in the conduction path and the current starting plateauing around -0.4 V. On the other end, when the input potential V is positive, thanks to the distribution of capacitance within the system, cations are shared among the three dendrites of the multiterminal OECT. Consequently, the dedoping of the conduction dendrite is far less important than observed when $V < 0$ V, and its conductance remains stable up to 0.9 V. This Y-shaped system thus presents a forward and a reverse direction, similar to a diode.

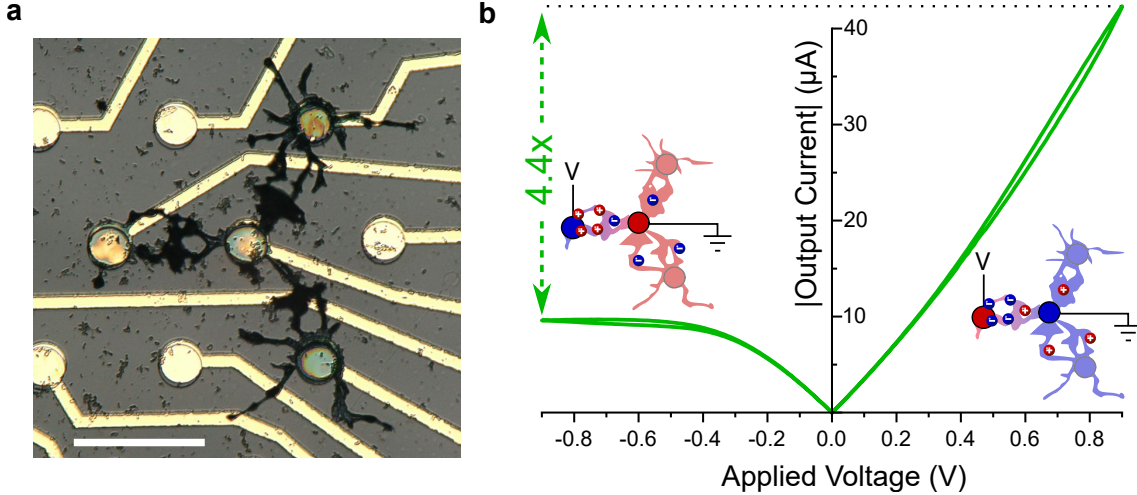


Figure 6.1: Self-gating effect in a dendritic system. a) Microscopic photo of the Y-shaped system. The scale bar represents $100 \mu\text{m}$. b) Output characteristics of the system when the central electrode is grounded and the voltage is swept on the left electrode. The plot shows the asymmetric behavior of the system related to the ionic distribution as shown in the inset figures. Relative potential is represented by the color scale, blue being the lowest potential in the network and red being the highest one.

The rectification coefficient, defined here as $|I_{OUT}|_{0.9V}/|I_{OUT}|_{-0.9V}$, is presented for all the permutations of the applied potential and ground in the system (see Figure 6.2). The higher rectification is obtained for the previously discussed case, with a ratio of 4.4, while in one of the other cases, when the lower right electrode is grounded and V is applied to the left electrode, a symmetrical (although nonlinear) behavior was observed, resulting in a rectification coefficient equal to one. These differences further highlight the intricate relationship between topology and nonlinearity: the specific location of the input conditions the output of the system.

As shown in Figure 6.2, one can influence the way the system behaves by carefully choosing which electrodes to address. Indeed, depending on the polarization of the

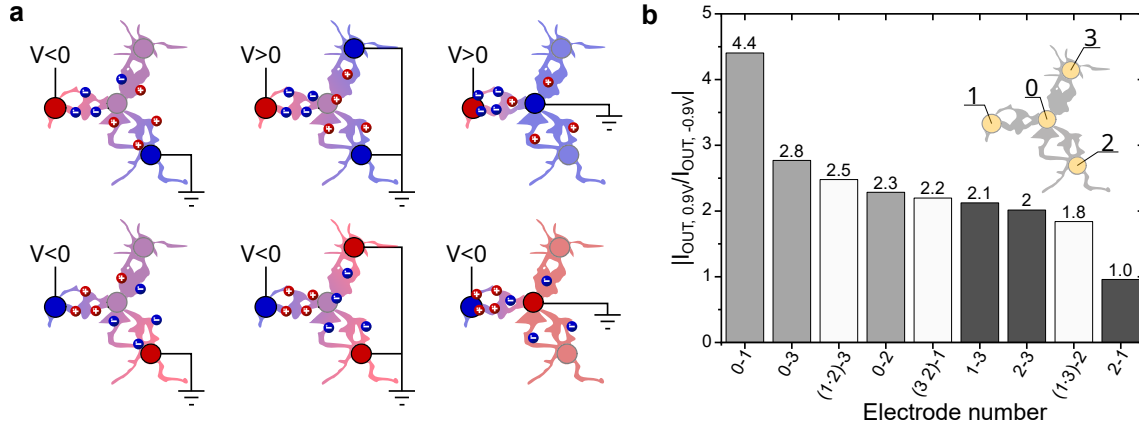


Figure 6.2: Distribution of the rectification coefficients. a) Schematic of ionic distribution within the system when different potentials are applied. Relative potential is represented by the color scale, red being the point of highest potential in the system while blue is the lowest one. b) Breakdown of the rectification coefficients as a function of which terminals are polarized. The x-axis indicates which electrodes are addressed in each configuration (e.g., (1 · 2)-3). The grounded electrode is always written before the hyphen (electrodes 1 and 2 are shorted), and the terminal undergoing voltage sweep is always written after (electrode 3). The numbers correspond to the electrodes indicated in the inset of the figure.

nodes of the circuit, the electrochemical doping profile (supported by the cationic charges in water) within the different PEDOT:PSS branches can be modified. This allows the control of different conductance profiles within a single Y-shaped dendritic network inducing various transport paths in the material, which is made possible only because the object has a singular shape. As such, a polymer dendrite cannot be considered as a discrete element with a well-defined number of inputs and outputs, since a direct structure-properties relationship emerges from these kinds of devices.

Voltage-induced modification of the behavior

In the previous example shown in Figure 6.1, two electrodes of the system were left floating. However, the electrical behavior of the system can also be modified by applying secondary potentials in the network in order to further influence ionic distribution. Figure 6.3a shows two different behaviors for the same system when the dendritic interconnects that are not used for conduction are polarized, compared to when they are left floating (in this case the floating electrodes will actually be connected to the ground with the central electrode). In both cases a diode-like behavior is observed. However, when a secondary potential V' is applied, the rectification coefficient is much more important, more than triple that of the other case (note that the sweeping range had to be reduced in order to avoid exceeding electrolysis voltage, $|V' - V| < 1.23$ V). This is due to the fact that the channel of the conductive dendrite is closed because of the potential difference applied to the system when $V < 0$ V. On the other hand, when $V > 0$ V, the potential difference helps overdoping the conductive path since the lowest potential in the circuit is V' , thus increasing the current that passes through the dendrite.

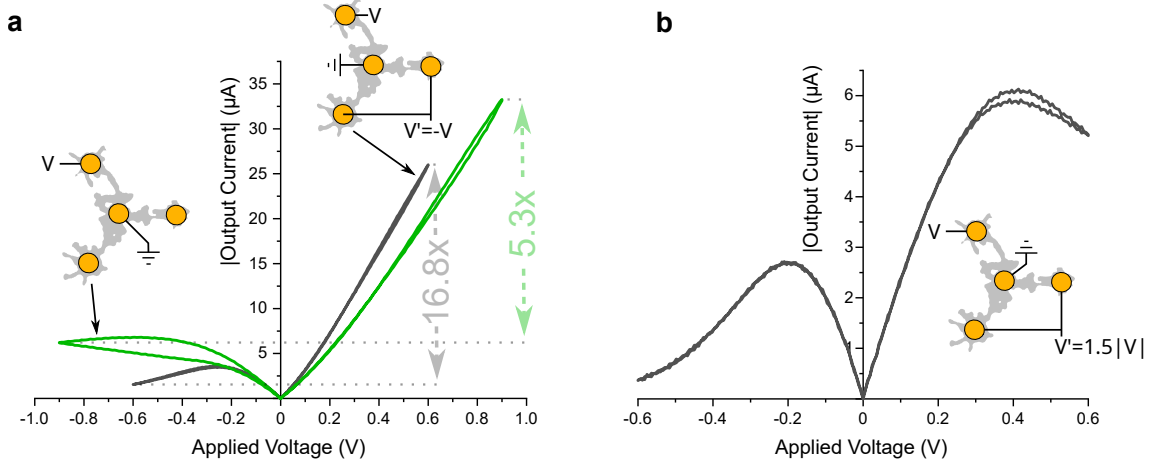


Figure 6.3: Voltage-induced modification of the behavior of a dendritic network. **a)** Output characteristics of a Y-shaped dendritic network, showing that the way the system behaves can be influenced by the application of external potentials on the electrodes of the system. The green curve shows the output of the system when the central electrode is grounded and the voltage V is swept in the top electrode, while the two other outer electrodes of the system are left floating. In contrast, the gray curve presents the output characteristic of the same system when the bottom and right electrodes are polarized such that $V' = -V$. **b)** When applying different potentials on the electrodes of the system, interesting electrical behaviors can be induced, mimicking for instance a negative differential resistance when $V' = 1.5|V|$.

Gaining understanding of the physics of the system allows implementing complex non-linear functions such as a negative differential resistance (NDR). A NDR is a property of electronic circuits that show a decrease in output current when the voltage increases. This behavior is best observed in Figure 6.2b where this voltage-current relationship can clearly be appreciated, in particular for negative potentials. Indeed, when $V < 0$ V, the output current starts increasing before reaching a maximum around -0.2 V. At this point, further increasing the applied potential leads to a decrease of the output current. For a Y-shaped dendritic network such as the one shown in the inset of Figure 6.2b, this behavior was implemented by setting the following condition: $V' = 1.5|V|$. In this way, two opposite effects happen simultaneously: the increase of the output current due to the potential difference between the source and the drain of the dendritic transistor and the dedoping of the conducting path due to the self-gating effect. For $V < 0.2$ V, the first effect dominates, resulting in an increasing output current. However, for higher potentials, dedoping becomes preponderant, leading to a decrease in current although the voltage keeps increasing. As discussed previously, this behavior is related to the ionic distribution in the system, which was taken advantage of in this case to force the closing of the channel of the dendritic transistor and to implement an NDR. Such phenomena are useful for the emulation of neural dynamics, and have for example been used to physically implement artificial neurons²¹⁹.

II.2 Inter-gating effect

In the previous section, I demonstrated that several dendrites physically sharing the same node are intricately connected. However, their electrochemical interconnection through

the common electrolyte is a frequent issue provoking cross-talking. Due to the common electrolyte, one of the most critical technological challenges for OEECTs is the ability to independently address each device of the matrix, which is required to access the information of a single element. This issue can be addressed at the expense of more complex fabrication procedures^{196,259}.

On the other hand, cross-talking between devices can also be apprehended as communication mean between physically disconnected parts of the system^{260,261}. PEDOT:PSS being a good material for efficient OEECT gating^{242,262}, two dendritic transistors operating simultaneously should therefore have a gating effect on each other, as illustrated in Figure 6.4. The influence of a dendritic transistor over another was assessed by growing two parallel dendrites, with the first being noticeably thicker than the second (Figure 6.4a).

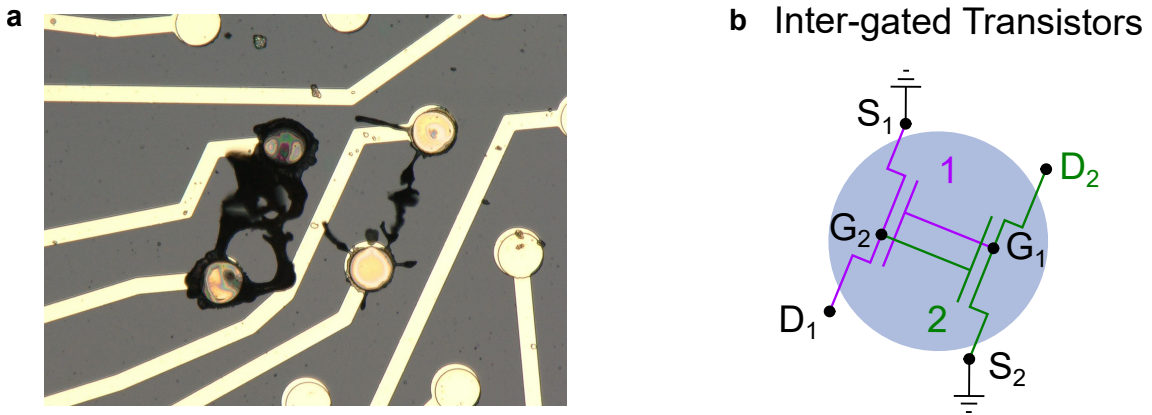


Figure 6.4: Inter-gating effect. a) Microscopic photo of the two parallel dendrites. b) Equivalent circuit of the two inter-gated transistors. The channel of each dendritic transistor behaves as the gate of the other transistor.

The dendrites were operated simultaneously and characterized as OEECTs, with their transfer and output characteristics recorded (see Figure 6.5 and Figure 6.6). Once again, the morphology of the dendrites appears to play a major role in the way the whole system behaves. Indeed, using the thicker dendrite as the gate rather than the thinner one (V_1 fixed while V_2 was swept, see Figure 6.5a) has a greater impact on the output current. Figure 6.5c shows that as V_1 increases, the drain current I_2 gradually decreases.

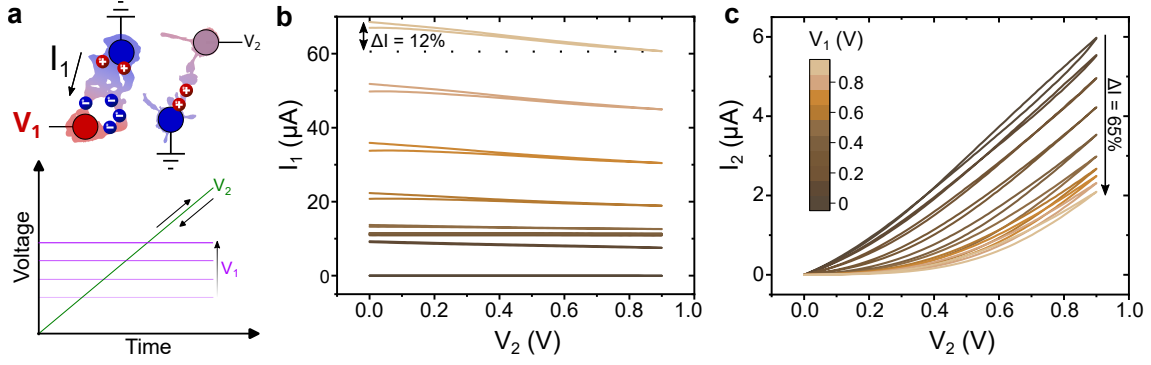


Figure 6.5: I-V curves of the system when the thick dendrite is used as the gate. **a)** Ionic distribution and experimental setup. The color scale indicates the nodes of highest potential (red) and lowest potential (blue) in the circuit. **b)** Modulation of current in the dendrite used as the gate of the system when the drain current (V_2) is swept. **c)** Output characteristics of the system.

The same trend is also observed when the roles are reversed and the thinner dendrite is used as the gate (see Figure 6.6a), although to a lesser degree: in the former case, the drain current decreases by 65% when $V_1 = 0.9$ V versus when $V_1 = 0$ V, compared to around 32% in the latter case (Figure 6.6c).

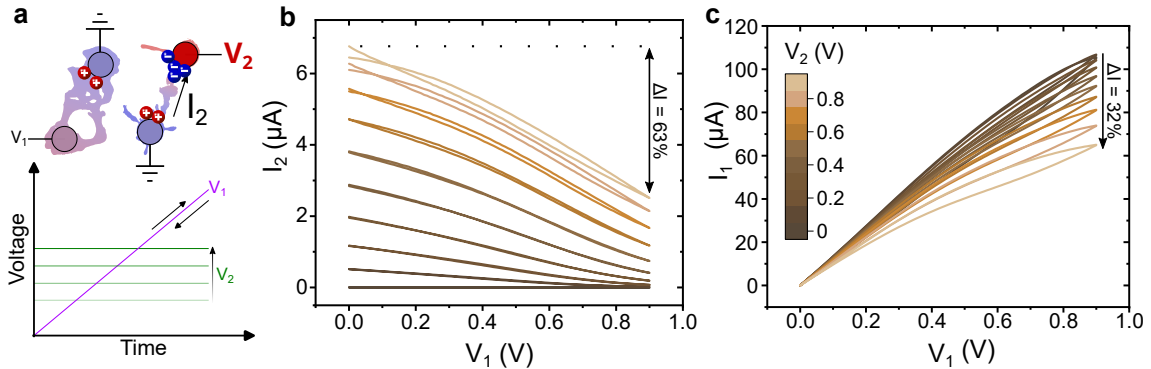


Figure 6.6: I-V curves of the system when the thin dendrite is used as the gate. **a)** Ionic distribution and experimental setup. The color scale indicates the nodes of highest potential (red) and lowest potential (blue) in the circuit. **b)** Modulation of current in the dendrite used as the gate of the system when the drain current (V_1) is swept. **c)** Output characteristics of the system.

The shape of the output characteristics is also distinctly different. When the bulky dendrite is used as the gate, I_2 remains very low up to around 0.4 V, before entering a linear region. Interestingly, this was not observed in the second case (Figure 6.6c), during which the output current follows a linear trend before starting to saturate, although the saturation region is not reached yet at 0.9 V. Once again, the morphology of the dendrites and their ability to retain ions might be at the origin of this phenomenon. With its large volumetric capacitance, PEDOT:PSS is an effective gate material that can efficiently promote doping/dedoping. The voluminous fiber is thereby able to dedope its neighboring dendrite almost entirely at high enough V_1 , as the lowest potential of the system is in that case the grounded electrode of the thin dendrite (Figure 6.5c). V_2 has to overcome a certain threshold to be able to move ions back into the system and to start regenerating the initial

high doping state of the dendrite. When the roles are reversed (Figure 6.5a), the thin dendrite also plays the role of the gate of the system, although less efficiently than its more voluminous counterpart as it is not able to close entirely the channel of the second transistor.

The fatigue effect in the material was investigated to make sure that the decrease of the output current was caused by a modulation of the conductance of the material. The output characteristics of the thick dendrites were thus recorded 10 times in a row, as shown in Figure 6.7. If a slight modulation is to be observed (around 10%), its magnitude is much less important than the effect observed previously, meaning that the modulation of current observed in Figure 6.5 and Figure 6.6 is indeed caused by an inter-gating effect between the two polymer dendrites. This modulation could be due to cations remaining in the bulk of the material²⁵⁴.

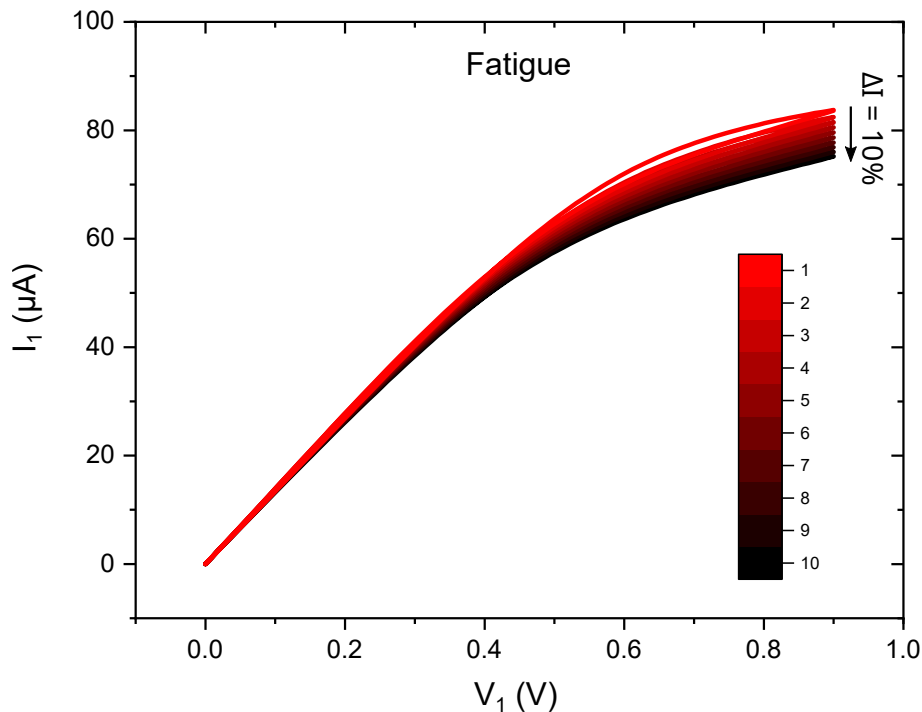


Figure 6.7: Fatigue effect in the more voluminous dendrite. While the thick dendrite was cycled in PBS, the thin dendrite was left floating.

The previous sections have established that a network of conductive polymer dendrites behaves as a multiterminal transistor, within which the dendrites are not a collection of independent elementary nodes, but are rather interconnected and interdependent objects. If a system made of multiple dendrites physically connected experiences a self-gating effect, two independent devices sharing the same electrolyte also undergo a similar inter-gating effect, where each fiber simultaneously acts as the channel and the gate of a transistor. As a consequence of this cross-talking, computing can be implemented on systems of electrically interconnected dendrites (forming a single continuous topology) as well as systems of distinctive dendritic topologies electrochemically connected with an electrolyte.

III Multiply-accumulate operation

Taking advantage of the inherent properties of dendritic networks, a multiterminal system was used to implement the multiply-accumulate (MAC) function (Figure 6.8). Monitoring only one output, it becomes possible to detect what combination of inputs was activated. The circuit was made of a Y-shaped device as the inputs, and the output current was read from the drain electrode of a single dendrite, all immersed in PBS. Pulses of voltage (0.6 V, 200 ms) were sent as inputs to the outermost electrodes of the Y-shaped device, the central terminal being grounded, in the fashion depicted in Figure 6.8c, while a small potential was continuously applied to the output electrode ($V_{READ} = 100$ mV).

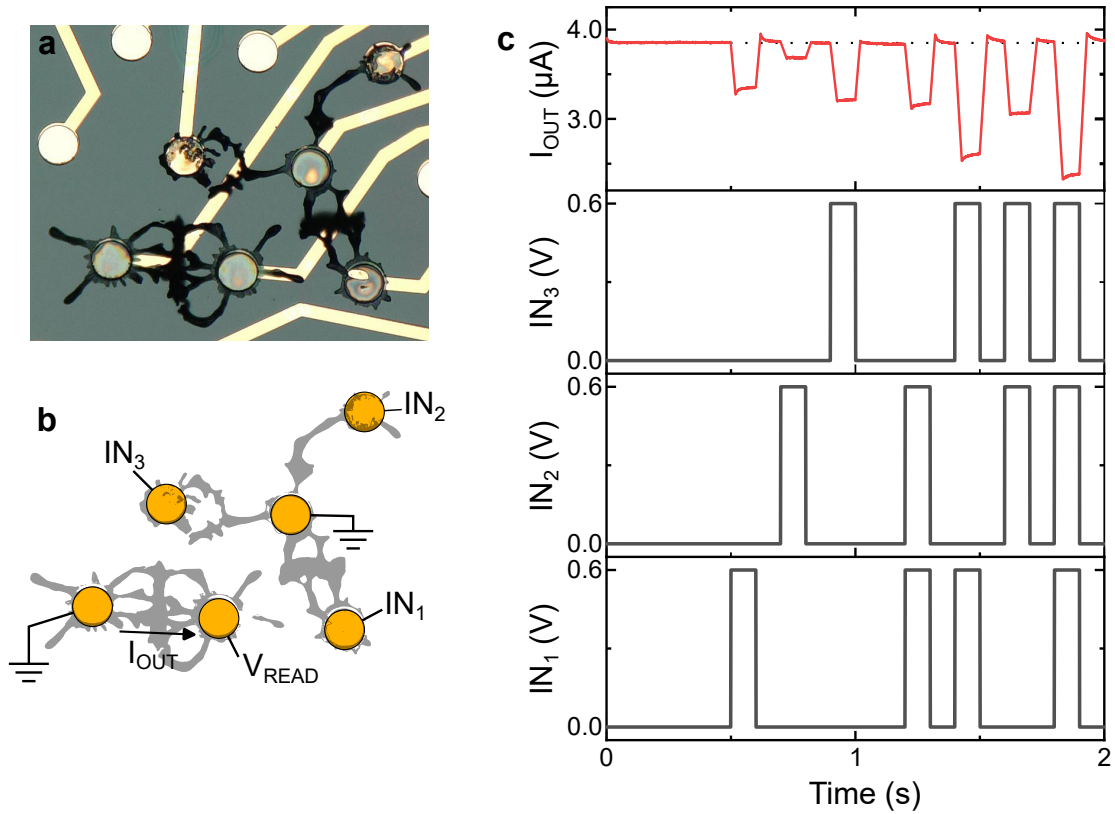


Figure 6.8: Physical implementation of the MAC function with a dendritic system. a) Microscopic photo of the dendritic network used to implement the MAC function. b) Experimental setup. The Y-shaped network is used as the input, while the output current is read at the drain of the single dendrite ($V_{READ} = 100$ mV). c) Output current and input voltages as a function of time. Transients were manually removed to focus on the steady-state response of the circuit.

As a result of the positive voltage pulses, the output dendrite experienced electrochemical dedoping due to the movement of ions through the electrolyte, as schematized in Figure 6.9. The modulation of current is characteristic of the activated input node: the output current undergoes a 13% modulation for IN_1 , 5% for IN_2 and 18% for IN_3 . The magnitude of the modulation appears to be related to the morphology of the dendrites, as seen in Figure 6.5 and Figure 6.6, as well as the distance that separates the input node from the output dendrite, the third input being the closest one to the output dendrite while IN_2 is the farthest away. This

dependency of the gating effect with distance from the drain electrode had already been observed in thin-film OECTs, and proved useful for orientation selectivity²¹⁴.

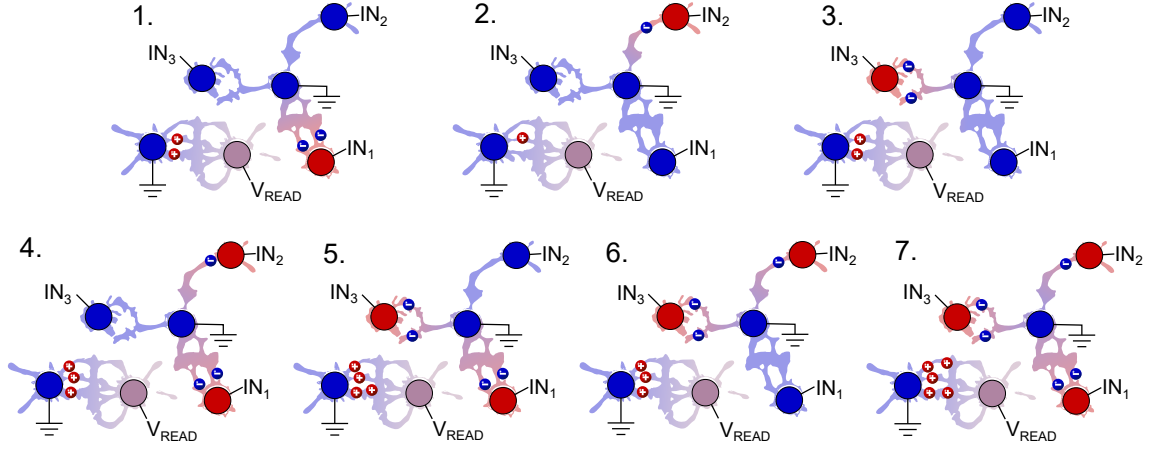


Figure 6.9: Ionic distribution schematizing the dedoping mechanisms in the system when the inputs are polarized. The color scale indicates the nodes of highest potential (red) and lowest potential (blue) in the circuit.

Here, as seen in Figure 6.9c, every single combination of inputs has a unique signature that can be sensed by the output dendrite. The more inputs are activated, the more important is the modulation of the output current. When all three inputs are set to 0.6 V, this modulation reaches a maximum of 38%. Therefore, with a read-out from a single node in the circuit, the system is able to discriminate which inputs have been activated. Such operation reproduces the multiply-accumulate (MAC) function that is a key operation in artificial neural networks. In this particular implementation, the weight is defined by the topology of the dendrites belonging to the Y-shaped device, while the accumulation is implemented with the read-out device. Nevertheless, it is important to note that, as for now, the stochasticity involved in the dendritic growth process implies that controlling with precision the weight of each dendrite is not so straightforward. As it has been demonstrated, the morphology of dendritic objects can be controlled globally. However, at the local scale, the stochastic nature of dendritic growth intrinsically promotes large variabilities, conflicting with the idea of physical reproducibility (and a fortiori of the reproducibility of electrical properties). Moreover, when operating with a higher number of polarized nodes in the system, growing a perfectly predefined network of dendritic OECTs can become delicate. From a general perspective, MAC operation corresponds to a linear function that could be rationalized for a small topological dendritic system (i.e., three branches of the Y-shaped device). In the following of the discussion, I will show that more complex topologies can lead to nonlinear functions that increase the computational power of dendritic networks.

IV *In materio* computing

The influence of the system on itself and on the nearby PEDOT:PSS dendrites, as well as its ability to retain information over time and discriminate past voltage events, are essential ingredients for the implementation of complex information processing tasks. Indeed, performing time series prediction thanks to the nonlinear

dynamics of such an organic electrochemical network²²³, coupled with the structural plasticity offered by this type of hardware^{151,221}, makes dendritic networks an exciting candidate to instill a new direction in the search of brain-inspired materials for the hardware revolution that is eagerly waited for. In this sense, conductive polymer networks could find themselves at the crossroad between different domains of research, borrowing ideas both from the biological realm (neurogenesis and synaptic plasticity) and computing. Yet so far, if dendritic networks appear to be a very useful tool to project information in a reservoir of higher dimension, a subsequent layer of machine learning is still required to perform classification²²³. In the following paragraph, I will propose a new framework based solely on hardware, capable of *in memory* spatiotemporal information processing, with the output condensing in a single value of current.

IV.1 Conductance programming using voltage pulses

As it was suggested that ionic motion and electrochemical dedoping were the causes of the inter-gating effect described previously, the volatility between high- and low-charge states in the dendritic system was studied to evaluate the temporality of the memory window with respect to different voltage patterns. The ability to learn and retain information over different timescales is of major importance in any computing paradigm drawing inspiration from the brain. As we saw in the first chapter of this manuscript, long- and short-term memory support learning and information processing⁷⁴. If structural plasticity allows the representation of information on longer timescales and can be implemented via electropolymerization, as its been demonstrated in this work as well as in references [151, 222, 224], shorter dynamics are also of primary importance. The present section of this manuscript will therefore explore the temporal dynamics of polymer dendrite networks.

Retention test

Drawing inspiration from the works of van de Burgt and coworkers²¹⁵, we therefore investigated whether it was possible to retain ions within the bulk of the dendrites longer than commonly accepted within a thin film of PEDOT:PSS material. As shown in Figure 6.10, a “WRITE” operation was performed with two parallel dendrites, one being grounded and the second one serving as the gate. Using a switch matrix, all the switches were opened after ten seconds of programming, and brief “READ” operations (about 50 ms) were performed every ten seconds or so, opening all the switches of the matrix between every operation. This way, the system was left as undisturbed as possible, and leak currents were limited.

Two distinct conductance states were observed and retained for some time following the “WRITE” operation, with a “WRITE” potential equal to ± 0.6 V (Figure 6.10c). Both “HIGH” and “LOW” conductance states displayed a similar modulation (respectively 9% and 8%) compared to a base current of around 6.6 μ A.

An exponential fit was applied to obtain the characteristic timescales, namely $\tau_{HIGH} = 387$ s and $\tau_{LOW} = 252$ s. The similar order of magnitude of these two values leads us to believe that the same physical process, most probably a leak current caused by ions returning to the electrolyte, is at the origin of the relaxation

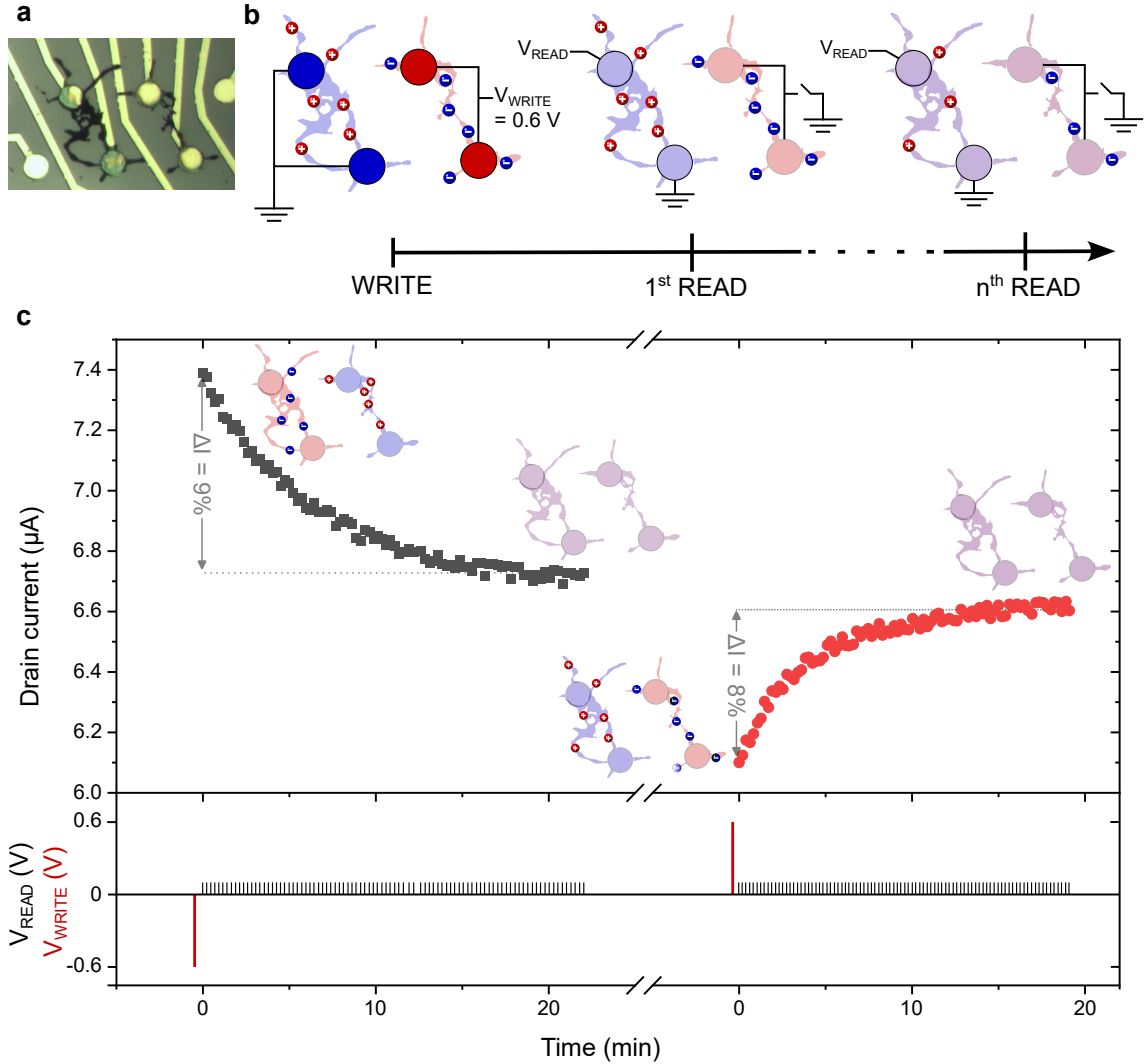


Figure 6.10: Conductance programming using write-and-read operations. a) Microscopic photo of the output dendrite (left), and the dendrite used as the gate (right). b) Working principle and ionic distribution in the system. During the writing phase, the output dendrite is grounded and V_{WRITE} is applied on both terminals of the gate dendrite. Afterward, every 10 s, a short read is performed (around 50 ms) to avoid perturbing the system. During the READ operation, the gate dendrite is left floating. c) Retention test of the system. Two distinct conductance states were observed and retained for some time following the “WRITE” operation, with a “WRITE” potential equal to ± 0.6 V. Both “HIGH” and “LOW” conductance states displayed a similar modulation (respectively 9% and 8%) compared to a base current of around 6.6 μ A. The inset figures show the ionic distribution in the system for each case, red being the highest potential and blue the lowest.

phenomenon observed in both cases. The output dendrite being more voluminous than the one used as the gate, it might be more difficult for cations to reenter the PEDOT:PSS fibers, which could explain why τ_{HIGH} is slightly superior to τ_{LOW} . Another explanation for this slight difference is that the duration of the experiment depends on the time that the equipment needs to save every measurement. Although the number of points is equal in both cases, it can be observed that the retention time for the “HIGH” state was monitored for a little over 20 min, whereas for the “LOW” state the experiment comes shy of the same time mark. In terms of a number of read operations rather than time, what we have is $\tau_{HIGH} = 28$ reads and $\tau_{LOW} = 21$ reads. The retention time of the system could therefore depend on the number of times that the system has been addressed (and thus disturbed), rather than on time itself, in addition to the previous hypothesis.

Cycling test

A cycling test was then performed on the same system, and 500 SET/RESET sequences of programming operations, alternated with READ operations, were run (Figure 6.11). Once again, two distinct conductance states are clearly discernable throughout the duration of the test, that lasted more than four hours. The evaporation of water as the electrolyte solvent, and therefore the increasing ionic concentration, did not seem to affect the behavior of the system over the duration of the experiment, although the lessening of the drop of electrolyte was clearly observable.

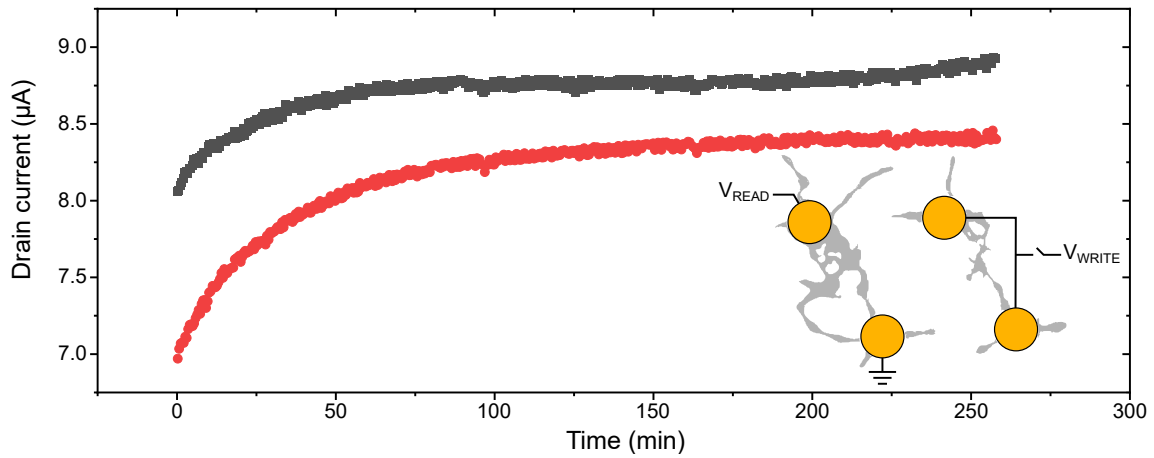


Figure 6.11: Cycling test of the system. The dendrite was consecutively programmed into the HIGH and the LOW conductance states, with $V_{WRITE} = -0.6$ V or $V_{WRITE} = 0.6$ V respectively. A short READ was performed after each WRITE operation. This sequence was repeated 500 times.

In addition, a dendritic network can also be programmed into different conductance states repeatably, as shown in Figure 6.12. Two parallel dendrites were grown, one serving as the gate and the second one used as the readout just as previously described, showing reproducible switching between discrete states for 12 programming pulses ranging from 0.1 V to 0.6 V, then -0.1 V to -0.6 V. The conductance of the device after a programming pulse appears to be relative to its past state, with positive pulses at the gate tending to dedope the readout and decrease its conductance, while conversely negative pulses would overdope it, increasing conductance. Dendritic networks therefore display state-dependent memory abilities.

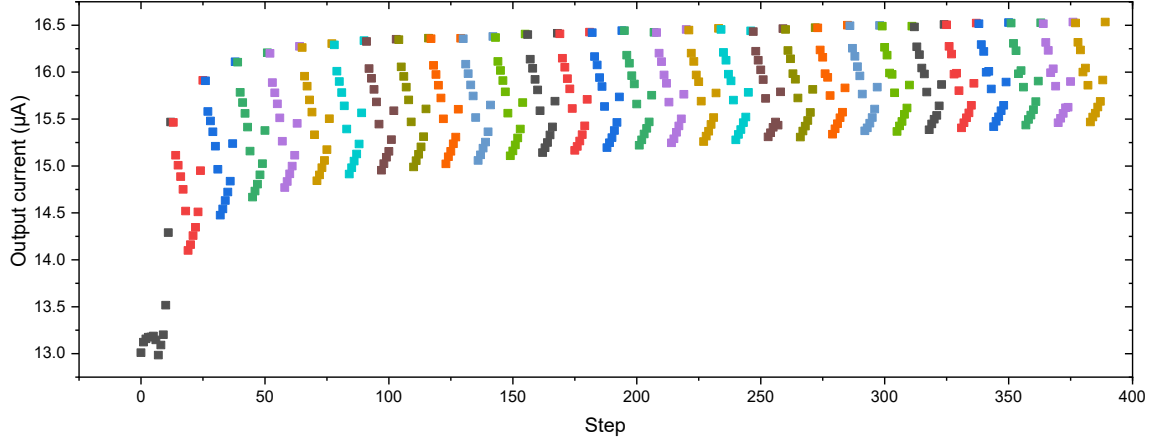


Figure 6.12: Voltage-induced conductance programming. A sequence of 12 programming pulses (0.1 V to 0.6 V then -0.1 V to -0.6 V) was repeated 30 times, showing reproducible switching between different conductance states.

While performing the experiment, a slow positive drift was experienced by the system, before reaching equilibrium some time after the beginning of the experiment, as can be seen in Figure 6.12 and Figure 6.13. This drift has also been detected in other similar tests that were run with different dendrites, although we do not clearly identify the cause of it. The origin of this drift was investigated, without much success. Further experiments showed that the direction of this slow dynamics could be modified depending on V_{WRITE} , and in particular the asymmetry between V_{WRITE}^{LOW} and V_{WRITE}^{HIGH} . In the previous cycling test, the experiment was symmetrical, with $V_{WRITE}^{LOW} = -V_{WRITE}^{HIGH} = 0.6$ V. Figure 6.13b shows that in case of a disequilibrium towards positive voltages, the direction of the drift was reversed. However, this disequilibrium had to be large enough, as in Figure 6.13a that drift was still positive. Interestingly, when a potential difference was continuously applied between the terminals of a dendritic OECT immersed in PBS, no drift could be observed over extended periods of time (2 hours in Figure 6.13c). Consequently, the ageing of the material does not appear to be the cause of the observed behavior.

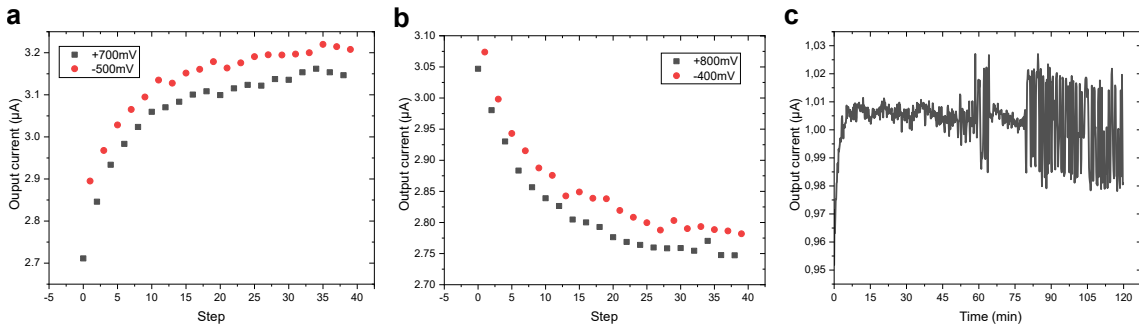


Figure 6.13: Investigation on the origin of the drift. The setup is similar to the one presented in Figure 6.12, but applying asymmetrical potentials. **a)** Cycling test for a dendrite under asymmetrical stress, $V_{WRITE}^{HIGH} = -0.5$ V and $V_{WRITE}^{LOW} = 0.7$ V. **b)** Cycling test for the same dendrite under a different asymmetrical stress, this time, $V_{WRITE}^{HIGH} = -0.4$ V and $V_{WRITE}^{LOW} = 0.8$ V. **c)** Output characteristic of a dendrite under continuous stress ($V_{DS} = 0.1$ V).

As some studies suggest that PEDOT:PSS films retain cations when undergoing

doping/dedoping cycles²⁵⁴, sequentially alternating between HIGH and LOW conductance states might induce a new equilibrium in the system that is only reached after some time. The previous experiments highlight that dendritic networks can be programmed into different conductance states and that they keep track of past voltage events, demonstrating low-volatile memory. These are critical ingredients for the implementation of information processing.

IV.2 Spatial information processing

First, we studied the response of a dendritic network to a single sequence of inputs, projected on different input electrodes. This sequence consists of eight patterns of three bits, always presented in the same chronological order, as introduced in Figure 6.14. The bits are encoded as a potential vs ground, negative to represent ‘0’ and positive to represent ‘1’. Each voltage pattern is applied during ten seconds (“WRITE” operation) and followed by a brief “READ” operation of about 50 ms, during which the output dendrite is biased ($V_{DS} = 100$ mV). Between two ‘WRITE’ operations, the system is at “rest” by setting the three input electrodes and the two terminals of the output dendrite to the ground for ten seconds.

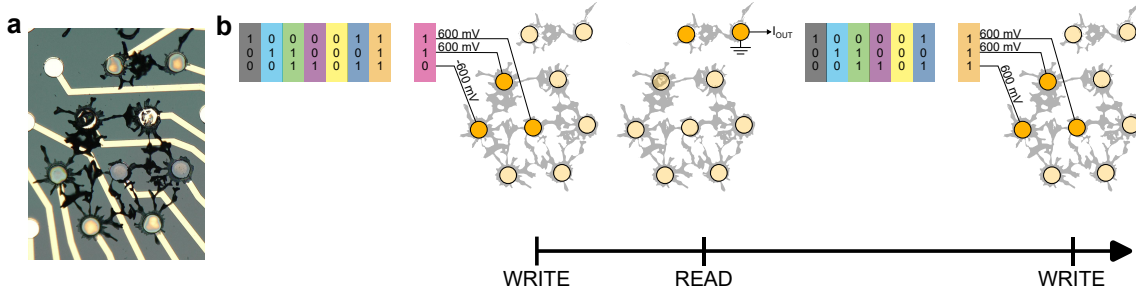


Figure 6.14: Experimental setup. a) Microscopic photo of the dendritic network. b) Principle diagram showing the WRITE and READ operation sequence. Each color represents a 3-bit voltage pattern.

As the input voltage sequence is repeated, a current modulation pattern appears in the output of the system (Figure 6.15a). This cycle is best observed in the normalized currents (Figure 6.15b), where the unique signature can be appreciated, and the drift observed in the output current has been removed. The modulation of current is computed as follows:

$$\frac{\Delta I}{I} = \frac{I_{READ} - I_{REST}}{I_{REST}}$$

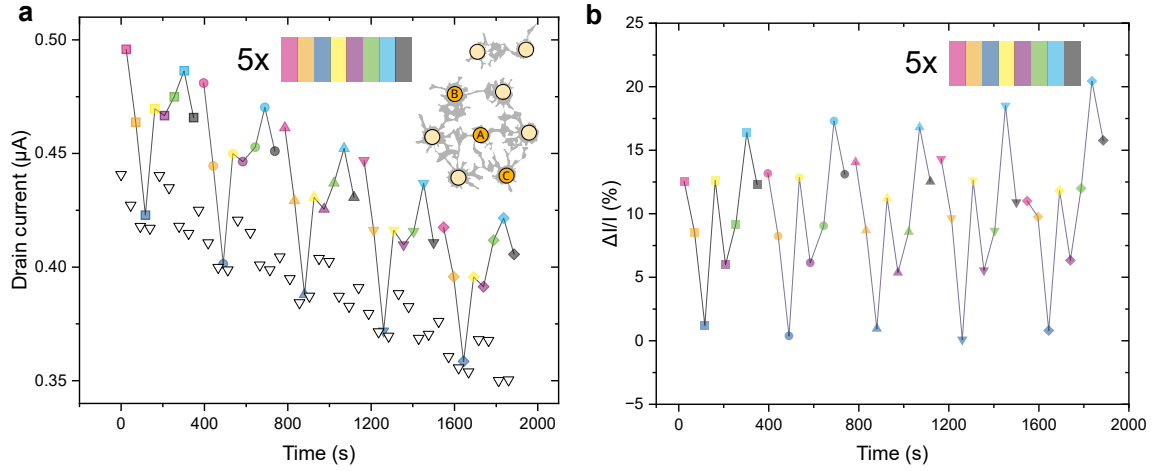


Figure 6.15: Output of the system for five repetitions of the same input sequence. **a)** Output current of the system for the three input electrodes chosen as depicted in the inset figure. The white down-pointing triangles represent the output current after a ‘REST’ operation, which are used to compute the output current variations ($\Delta I/I$). **b)** Output current variations, computed as the difference between the output current measured after a WRITE operation and the previous REST operation (all terminals set to the ground for ten seconds).

When the same temporal sequence is projected on different input electrodes, the output of the network is modified. As this signature is related to the ability of the network to distribute ions between the different dendrites, and thus to the morphology of the network, this suggests that no two electropolymerized systems will give the exact same response to similar inputs, thanks to the stochasticity observed in the electropolymerization process^{154,230}. The possibility to build this type of hardware on demand could have very powerful implications for security purposes, notably in the quest for physically unclonable functions^{263,264}.

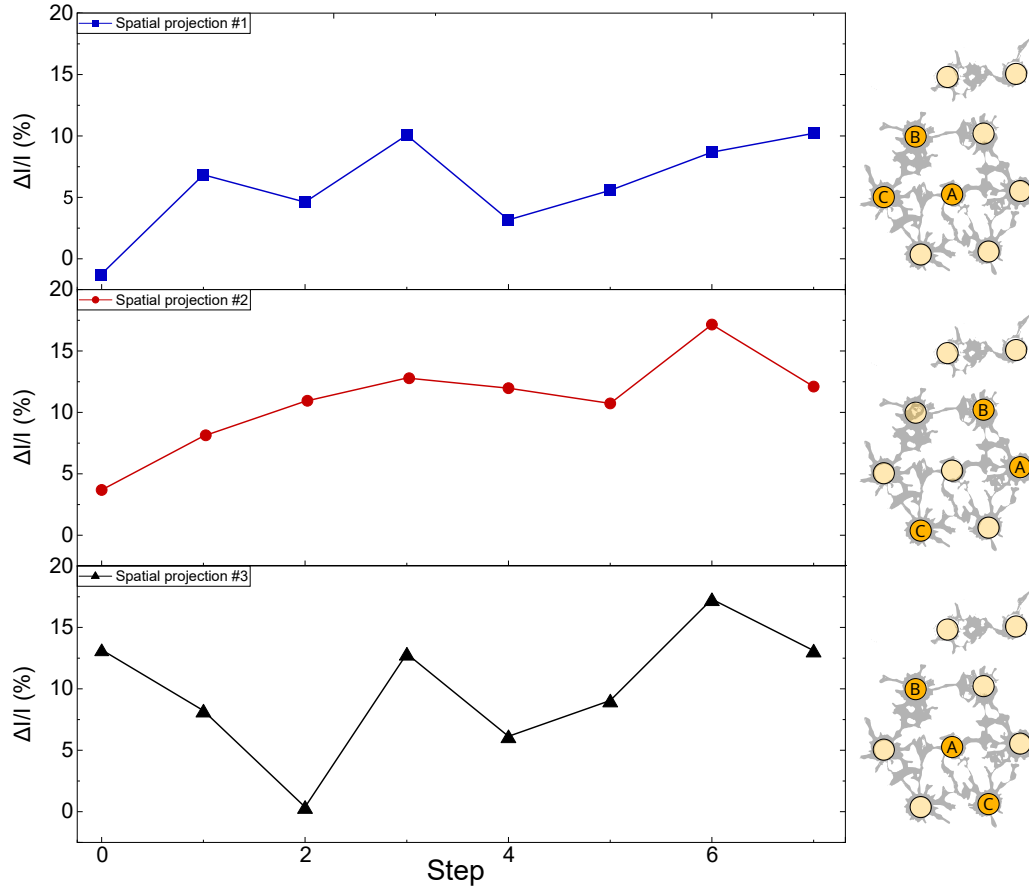


Figure 6.16: Output current variations for different sets of inputs receiving the temporal sequence presented in Figure 6.14. In the schematics of the dendritic system on the right, each letter represents the position of a single bit in the 3-bit pattern (A,B,C).

Interestingly, even when two spatial projections are quite similar, the signature of the system in the output current can be told apart: for spatial projections #1 and #3 (SP1 and SP3), only one of the input electrodes differs (Figure 6.16), yet in both cases the output current variations are clearly distinguishable. Figure 6.17 indeed shows that each spatial projection signature is different and consistently reproducible.

In addition, this variation depends on the position of the input electrodes. Figure 6.17 shows that the maximum increase in current is reached at step 3 (yellow) for SP1 (about 10%), whereas it happens at step 6 for SP2 (light blue) and SP3 and is more consequent (slightly superior to 15%). In other words, the same voltage pattern does not have the same effect on the output of the system depending on which input electrodes are addressed. This means that, given only the output current on one electrode, the system would be able to discriminate between different sources of information. This amounts to a wireless electrochemical communication system, where information can be transmitted though the electrolyte, and each source of information could be identified by its unique signature. Such spatial projection of the input patterns also demonstrates that nonlinear functions emerge when the topology becomes more complex. In comparison to the quasi-linear MAC operation demonstrated with the Y-shaped devices, multi-terminal devices cannot be rationalized as

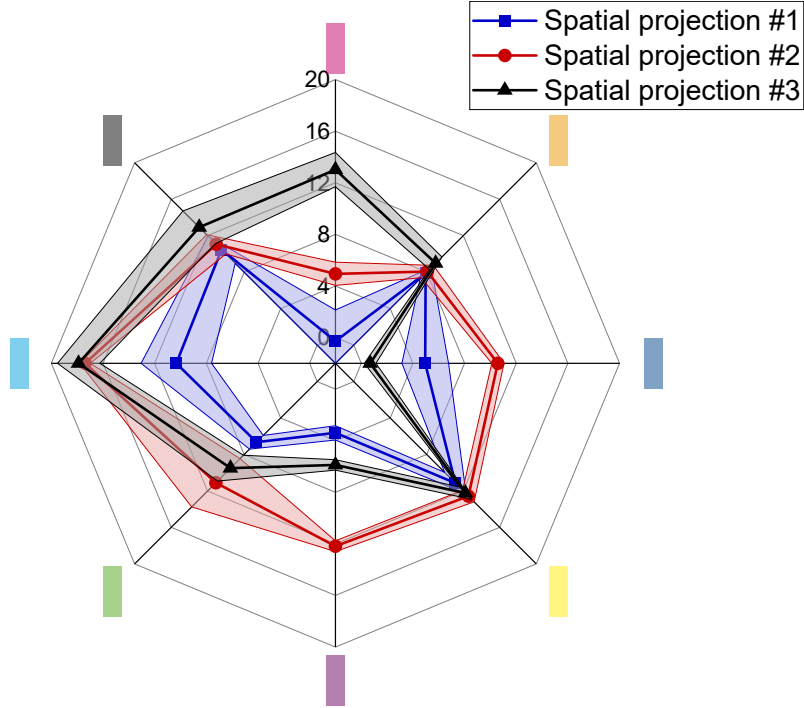


Figure 6.17: Spider diagram showing the average current variation and standard deviation over five cycles for three different sets of inputs. Chronological order is shown clockwise, starting from the top of the diagram.

a linear combination of the inputs. In fact, the (1,1,1) and (0,0,0) patterns do not produce the maximum and minimum modulation of current as would be expected from a linear combination of input patterns.

IV.3 Temporal information processing

If spatial information discrimination allows for the unique identification of the emitting source, the dendritic network is also able to encode temporal information. Indeed, even if two sequences are sent to the same input nodes, the chronological order in which the patterns are presented to the system may influence the output. Using the same dendritic structure as for spatial information processing (see Figure 6.14), we demonstrate here that the output of the system is state dependent.

Figure 6.18 shows that a single voltage pattern can have very different contributions to the output of the system based on the history of the inputs that have previously been presented to the structure. For example, for both temporal projections, the second voltage pattern presented to the dendritic network is identical (the (1,1,1) pattern represented by the orange color). In the first case, the system appears to have been potentiated by the first pattern of the sequence. Therefore, the second voltage pattern has a negative contribution to the output current. However, for the second temporal projection, the output dendrite has already been depressed by the first voltage pattern of the sequence. As a consequence, the modulation of current following the second pattern of the sequence (the orange one) is in that case positive.

Each voltage pattern thereby has a different influence on the output current

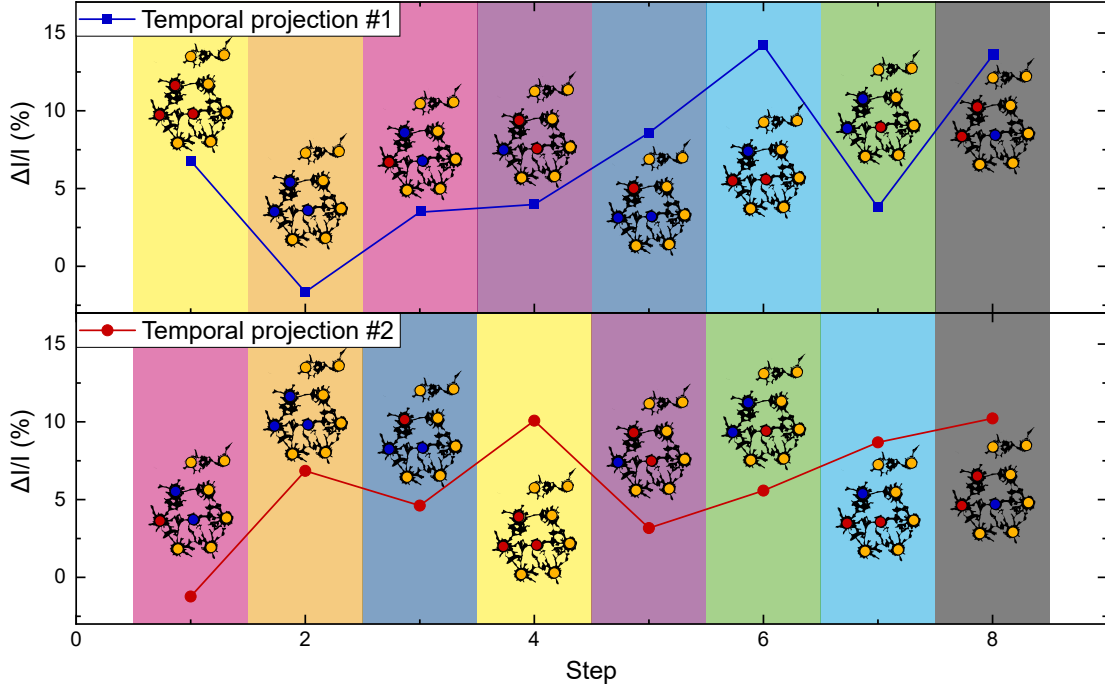


Figure 6.18: Output current variations for two different temporal sequences (the 3-bit patterns are the same, only shuffled) presented to the same input electrodes in the system. Each color represents a 3-bit pattern. Inset : schematic representation of the dendritic system, where blue represents a HIGH logic state (physically encoded as 600 mV), and red represents a LOW logic state (physically encoded as -600 mV).

variation based on the internal state of the system. Figure 6.19 shows that, in order to be interpreted correctly, the whole input sequence needs to be captured. Indeed, reading the output related to only one voltage pattern will not allow to identify that voltage pattern, as its signature may vary depending on the internal electrochemical doping state of the output dendrite, and thus on the previous voltage events that took place within the system. If the network happened to be memoryless, the output current variation would no depend on the chronological order of the input sequence, as the variation of current is always calculated compared to the previous resting state, as explained in the previous section.

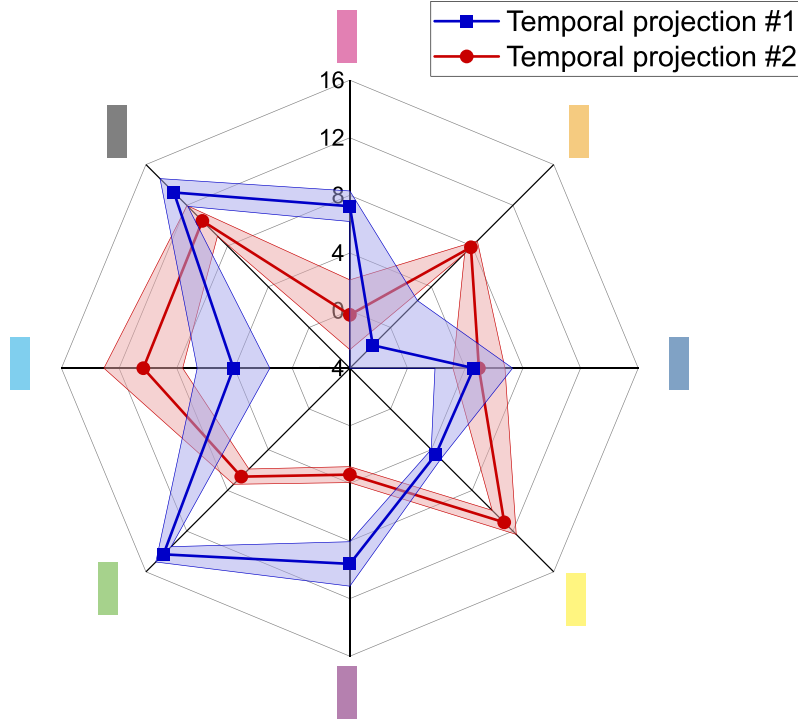


Figure 6.19: Spider diagram showing the average current variations and standard deviation over five cycles for two different temporal sequences. Note that following the diagram clockwise does represent the chronological order for temporal projection #2, but not for #1.

This behavior of dendritic networks could have a tremendous impact on information processing: by only reading the output current of a single dendrite, within a characterized dendritic network, a dendritic device would be able to provide both spatial and temporal information. This is due to the highly complex and interwoven interactions that arise within such networks sharing a common electrolyte, a communication medium that allows the exchange of information between physically disconnected parts of the network.

V Morphology-dependent functions

Exploiting the intrinsic randomness of a manufacturing process to generate a unique output considered to be a signature of a device is an idea that has important repercussions for security purposes, and that has already been explored to implement Physical Unclonable Functions^{263,264}. In that regard, electropolymerized networks have plenty to offer, given the hardware-dependent functions that emerge as a consequence of the particular distribution and morphology of the polymer fibers within the network. The patterns emerging from the spatial and temporal projections in the previous section of this article could indeed be considered as a signature, a unique identification key: in order for the desired output to be generated, the correct input has to be sent to the network, both in terms of the combination of input electrodes but also of the temporal sequence. Moreover, as electropolymerization is an easy-to-perform process, this means that the signature of the device can be generated by the end-user itself, making him the only one in possession of the unique signature

of the device.

This section will demonstrate that dendritic networks with similar topologies still exhibit distinct functional properties. Two different networks were grown, one with thick dendrites (growth frequency of 80 Hz) and the other with thinner objects (growth frequency of 500 Hz). Both networks were grown to have a comparable topology, on each side of a unique dendrite that was used as the readout of both systems (see Figure 6.20).

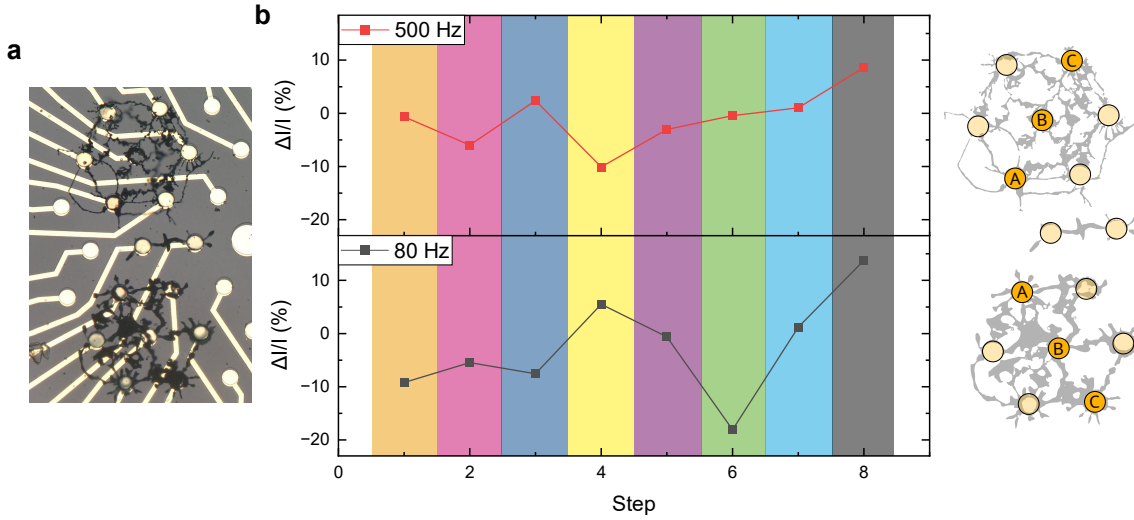


Figure 6.20: Morphology-dependent functions. **a)** Microscopic photo of the system, showing the two dendritic networks grown at 80 Hz (thicker dendrites at the bottom) and 500 Hz (thinner dendrite at the top), with the readout in the middle. **b)** Output current variations of the two devices over one cycle, for the same input sequence. The sets of electrodes were chosen to be symmetrical relative to the readout dendrite.

The same sequence of patterns as presented in Figure 6.14 was applied to both networks sequentially. This resulted in two different outputs, based on which network was addressed, as shown in Figure 6.20 and Figure 6.21. While the signatures of some of the patterns were very similar for both networks (notably the orange, purple, light blue and gray ones), the behavior was divergent for the other inputs. Indeed, the green and pink patterns triggered a potentiating modulation of the output current for the 500 Hz network, while in the 80 Hz network they had a depressing contribution. Consequently, over the whole sequence, each network possesses its own unique signature, as shown in Figure 6.21. The two networks, although they share the same electrolyte and the same readout dendrite, still can be differentiated by their electrical behavior even when presented identical inputs. Thanks to this property, a single readout device could thus be used to read a whole constellation of dendritic networks sharing the same medium, as each network would have a unique input-output relationship with the readout.

Two important conclusions can be drawn. First, although the morphology of the fibers themselves is impossible to control finely, as it has been mentioned elsewhere¹⁵¹, it is still possible to define the topology of the network from a global standpoint. Indeed, the connections between the nodes has essentially been main-

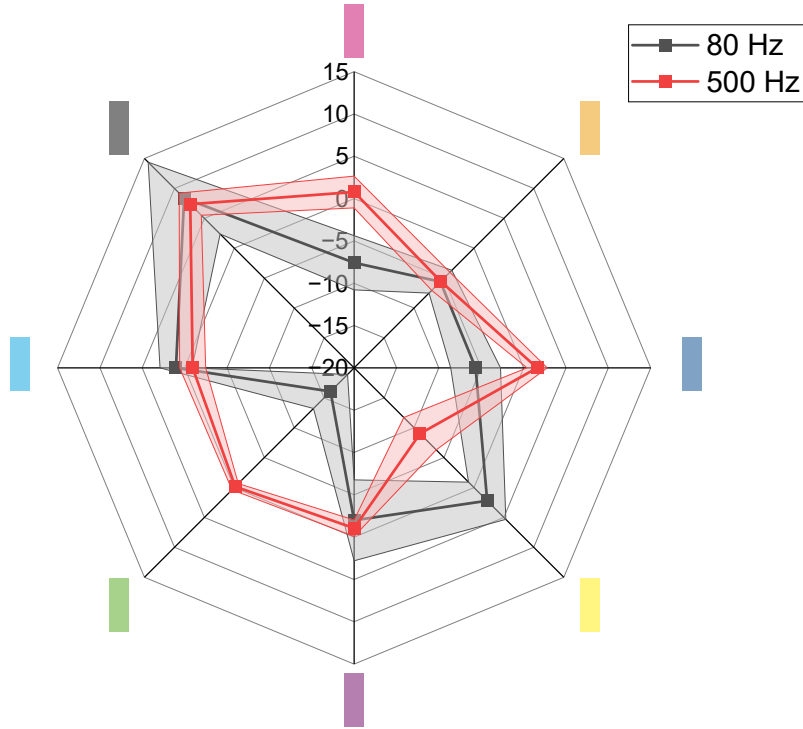


Figure 6.21: Spider diagram showing the average and standard deviation of the output current variations of the two devices computed over 8 cycles. Chronological order is shown clockwise, starting from the top of the diagram.

tained: although the conductance of a specific dendrite might be harder to define, two nodes can still be connected from a binary perspective. Second, two different networks, however similar they might look, are not identical. If this holds from a physical point of view, this is even clearer when comparing their electric behavior. This ensures the uniqueness of every network grown on such a platform as the one presented in this article, which is at the same time a boon (e.g., for cryptographic functionalities or classification problems) and a bane for applications such as deterministic logical functions.

VI Conclusion

Adding to the results detailed in the previous chapter (see FUNCTIONAL PLASTICITY IN POLYMER DENDRITES), this chapter establishes the interest electropolymerized PEDOT:PSS dendritic networks as a new class of morphology-dependent hardware, which properties not only arise from the material but also from the morphology of the fibers and the topology of the network. We highlight the complex interplay between form and function in these structures.

It was brought to light that these objects present a nonlinear and asymmetric electric behavior directly related to their morphology and the ionic distribution across the dendrites, which depends on the activity of all the fibers of the network. PEDOT:PSS dendritic networks also appear to possess an intrinsic memory that, coupled with the self-gating and inter-gating effects of the polymer dendrites, allows the system to perform *in materio* computing tasks, such as spatiotemporal infor-

mation processing. Indeed, conductive polymer networks are able discriminate the source of surrounding voltage events and present state-dependent computing abilities, as in such systems the output of the network seems to be conditioned by past inputs. Finally, we evidence that these properties, coupled with the stochasticity of AC electropolymerization, allow the growth of networks which topologies might be very similar on a large scale, but in which the distinct morphology of each dendrite gives rise to a unique granularity, and in consequence a distinctive electric behavior.

Electropolymerization definitely emerges as a fabrication method of choice in the quest for a new paradigm that could help solve the hardware crisis at hand, as it offers the possibility to create complex structures with minimal resource requirements, implementing an effective bottom-up approach. With close to no need for complex and costly machines and incredible simplicity, it has the potential to foster a new era of organic, cost-effective and bioinspired electronics.

7

DENDRITIC ELECTRODES FOR IN VITRO ELECTROPHYSIOLOGY

In the last two chapters, we delved into the realm of unconventional computing, performing signal processing with dendrites and dendritic networks in a bio-inspired fashion. In particular, in chapter FUNCTIONAL PLASTICITY IN POLYMER DENDRITES, I showed the ability of PEDOT:PSS dendrites to perform the detection and classification of voltage bursts, a function that appears to be very appealing in the context of neurosensing. In addition, PEDOT:PSS is a biocompatible polymer that has been the workhorse of bioelectronic applications for many years. Furthermore, in chapter DENDRITIC NETWORKS FOR MORPHOLOGY-DEPENDENT COMPUTING HARDWARE, the computational possibilities provided by polymer dendrite networks were brought to light. In this chapter, we will therefore explore the usefulness of polymer dendrites in the context of *in vitro* electrophysiology.

I Introduction

A major advantage of organic polymers over common electronic components is their ability to interface seamlessly with biology, which has made organic electronics especially interesting for bioelectronic applications^{164,265}. In particular, PEDOT-based bioelectronics have attracted a lot of attention due to the outstanding electrochemical properties of the polymer when it comes to sensing and stimulation of electroactive cells^{162,266}. In addition, conductive polymers are more mechanically flexible than their inorganic counterpart, whose higher Young's modulus causes mechanical mismatch with soft living tissues, and are not sensitive to surface degradation via corrosion for instance²⁶⁷. If using PEDOT as a coating material on metallic electrodes has been proven to improve the performances of microelectrodes^{137,231,268}, it is also a material of choice for the fabrication of active devices such as OECTs. As already discussed in the first chapter of this thesis, the OECT was demonstrated to be an efficient biosignal transducer^{194,195,199,200}.

One additional reason that made PEDOT coatings so appealing is the ease with which they can be added to already existing devices. If these polymer layers can be spin-coated or drop-casted, electrodeposition might well be the most versatile deposition method, as it was discussed in the previous chapters of this manuscript. The strength of electrodeposition resides in the high degree of control operated over the properties of the electrogenerated material^{127,151}. Yet, electropolymerization has

mostly been employed as a post-fabrication tool for the coating of microelectrodes to increase recording performances^{137,269,270}. But as I have demonstrated in this work, this technique also offers the possibility to grow singular objects with unique properties that could not be obtained with traditional fabrication routes. In this regard, the work of Berggren and coworkers sheds a new light on the possibilities offered by conducting polymers, as they leveraged the metabolism and physiology of plants^{246,271} and animal models^{247,248,272} to grow polymer fibers *in vivo*, thus paving the way for self-organized bioelectronics in living tissues. If the electrodeposition of PEDOT around living cultured neurons had previously been demonstrated to improve the interface performances²⁷³, these recent *in vivo* results hold tremendous opportunities for the implementation of smart interfaces, as it was moreover highlighted that neuromorphic organic devices could implement biohybrid synapses which activity could be modulated by neurotransmitters such as dopamine or serotonin^{217,274}.

Janzakova and coworkers had already shown that PEDOT:PSS fibers operate as OECTs¹⁴⁷, but these objects were never before used for electrophysiological applications. So far in this manuscript, we proved that PEDOT:PSS fibers could be used in a biologically plausible environment as electrochemical classifiers for the detection of voltage bursts (see FUNCTIONAL PLASTICITY IN POLYMER DENDRITES), and that networks of such polymer dendrites could perform spatiotemporal information processing thanks to *in materio* computing abilities (see DENDRITIC NETWORKS FOR MORPHOLOGY-DEPENDENT COMPUTING HARDWARE). The last question that will therefore be explored in this final chapter is the following: could PEDOT:PSS dendritic fibers combine sensing capabilities for the detection of electrophysiological signal and bioinspired information processing in a unique system? Currently, sensing nodes rarely integrate computing units, resulting in a significant amount of energy spent solely for exchanging data with the computing nodes of the system. A smart biohybrid interface could potentially reduce energy consumption by enabling edge computing^{275,276}. In this final chapter, we will first investigate polymer dendrites as a biohybrid interface, and how such objects compare to standard microelectrodes when it comes to neural sensing. Second, we will explore how these polymer fibers perform when used as OECTs in cell cultures.

II Effect of sample preparation on polymer dendrites

II.1 PEDOT:PSS fiber growth

Polymer dendrites were grown on a standard MEA, with a growth frequency varying from 50 Hz to 300 Hz to obtain different morphologies. Following the growth of polymer dendrites on the substrate, as described in MATERIALS AND METHODS, electrical characterizations were realized between each step of the sample preparation. To minimize deviations from the protocol and to avoid introducing unnecessary variability, these characterizations were performed in deionized water instead of PBS for two main reasons. First, since deionized water was used for rinsing after sterilization with ethanol and coating of the adhesion layer, it would minimize the risk of

mechanically stressing the dendrites due to unwanted manipulations. Second, the absence of ions in deionized water would reduce concerns about the potential ageing of the dendrites due to cation penetration within the bulk of the material, which would eliminate a source of variability.

II.2 Sterilization and adhesion coating

During previous electrophysiological experiments led in our group, the sterilization protocol involved boiling the samples in an oven heated to 125°C for 30 min. Due to concerns regarding the mechanical fragility and stability of polymer dendrites, the sterilization protocol had to be modified, and its effects on the electrical properties of PEDOT:PSS fibers was assessed. Although it has been demonstrated that autoclaving does not seem to affect the properties of usual PEDOT:PSS devices²⁷⁷, ethanol sterilization was chosen as an alternative. This step was then followed by the application of a coating layer to improve cellular adhesion to the substrate (see MATERIALS AND METHODS for more details concerning sample preparation).

The effect of these different preparation steps are presented in Figure 7.1, where 35 dendrites were used to compute the average curves, after 3 of them broke during manipulations (two before the first characterization, one after sterilization, all different growth frequencies).

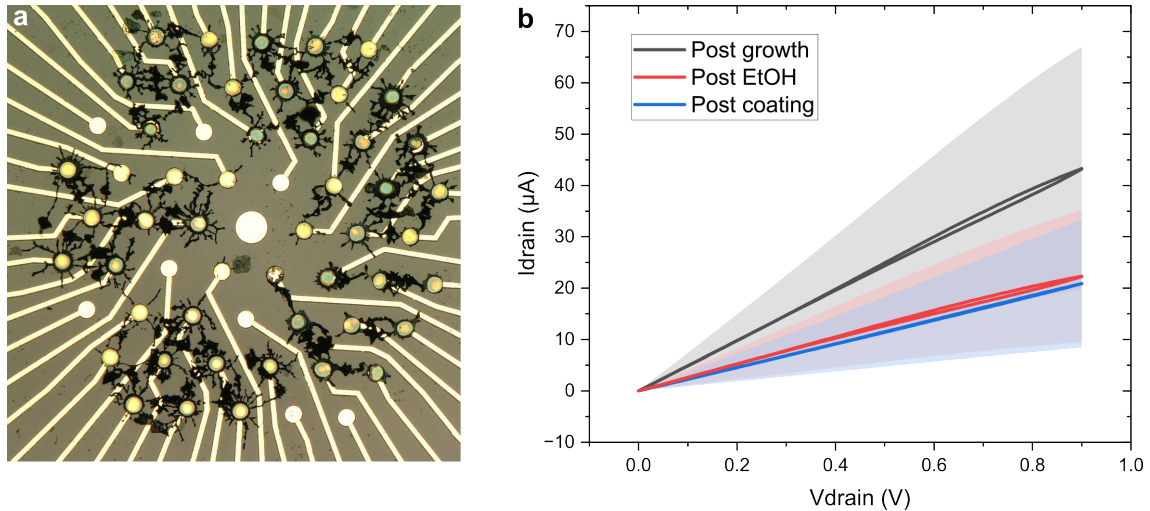


Figure 7.1: Effect of sample preparation on the conductance of polymer dendrites. a) Picture of the MEA used for the test. b) Average output characteristics after each step of the preparation protocol. The average curves were computed over the characteristics of 35 dendrites.

In the absence of ions to modulate their electrochemical doping level, polymer dendrites acted as ohmic conductors in the voltage range used during the experiment, and no saturation was to be observed. In addition, very little to no hysteresis between trace and retrace are to be reported, as expected. Lastly, the electrical behavior of the dendrites was only assessed under a positive bias, as the asymmetric behavior related to the morphology of these fibers should not be observed with no cations to support the auto-gating effect described in DENDRITIC NETWORKS FOR

An important dip of about 50% in conductance was observed after sterilization with ethanol (from 48.1 μS to 24.8 μS), whereas little difference was to be seen after the adhesion coating was applied (from 25.8 μS to 23.2 μS). These results were quite surprising for two reasons. First, post-treatment with alcohol has been demonstrated to increase the conductivity of PEDOT:PSS film through morphological reorganization and a decrease in thickness of the PSS insulating phase^{179,278,279}. Second, the positively charged adhesion proteins of the coating were thought to have an electrochemical dedoping effect of the polymer dendrites (since Poly-D-Lysin is polycationic at neutral pH), which was only observed very marginally experimentally. If this hypothesis could hold in the case of a homogeneous fiber, the experimental observation could however be explained by the two-phase model usually applied to PEDOT:PSS thin films¹⁷⁶: as the outer shell of the dendrite would then be constituted mostly by an insulating PSS-rich shell, a surface effect of the adhesion layer would therefore not have a major effect on the conduction of the polymer.

The main contribution to the drop of conductance might come from manipulation-induced stress and air drying between each step of the preparation protocol. PEDOT:PSS acting like a sponge²⁸⁰, drying up might cause mechanical tensions and ruptures in the polymers dendrites. This effect could even be amplified by the high volatility of ethanol, which would explain why the most important decrease in conductance happened after sterilization. In addition, and from personal experience, the first few rinses always are the most critical ones. However, growing many dendrites from the same electrodes and multiplying the number of dendritic branches seemed to be a good way to minimize damage on the structures.

Even if the protocol led to an important decrease in the conductance of the dendrites, a vast majority of them was still properly functioning as a conductor after coating. This sterilization method was thus implemented for cell cultures.

II.3 Ambient and long term stability

The same MEA was stored under ambient conditions and periodically monitored to assess long-term stability of polymer dendrites. The first and final measurements were conducted 7 months apart, with the sample stored in an uncontrolled environment without any special precautions.

Figure 7.2 presents the evolution in time of the conductance profile of the dendrites of this sample, characterized in DIW. A gradual conductance decrease is observed for all polymer fibers, going from an average of 23.2 μS in December to 5.2 μS the next July. Over a 7-month period, the dendrites underwent a 78% drop in their conductance. Overall, this drop appears to be quite uniform, as the most conductive dendrites tend to remain so over time (see the similar shapes of the curves in Figure 7.2a). The large standard deviation displayed in Figure 7.2c helps to appreciate the variability related to electropolymerization.

Nevertheless, these results highlight that polymer dendrites remain conductive

up to several months after growth, even when stored with no special precautions. It remains unclear whether the loss of conductance observed for all samples is a consequence of the ageing of the material or rather of mechanical damages caused by the addition and drying of the solvent during electrical characterization. A decrease in conductivity in PEDOT:PSS films aged in ambient conditions was already reported in the literature²⁸¹, and concluded that the degradation was primarily driven by a reorganization of the material and a change in morphology. This result is backed up by another study, who demonstrated significant reorganization of crosslinked PEDOT:PSS influenced by the relative humidity and temperature of the environment²⁸². In addition, it appears that upon exposition to water, PEDOT:PSS could suffer partial dissolution, where excess PSS⁻ moieties are removed^{283,284}, although little effect on film conductivity was reported.

In conclusion, although changes in the morphology and the conductivity of PEDOT:PSS dendrites are expected to happen, storage in ambient conditions and the few days that usually separate dendritic growth from cell culture should only marginally affect the electrical properties of the dendrites prior to *in vitro* experiments.

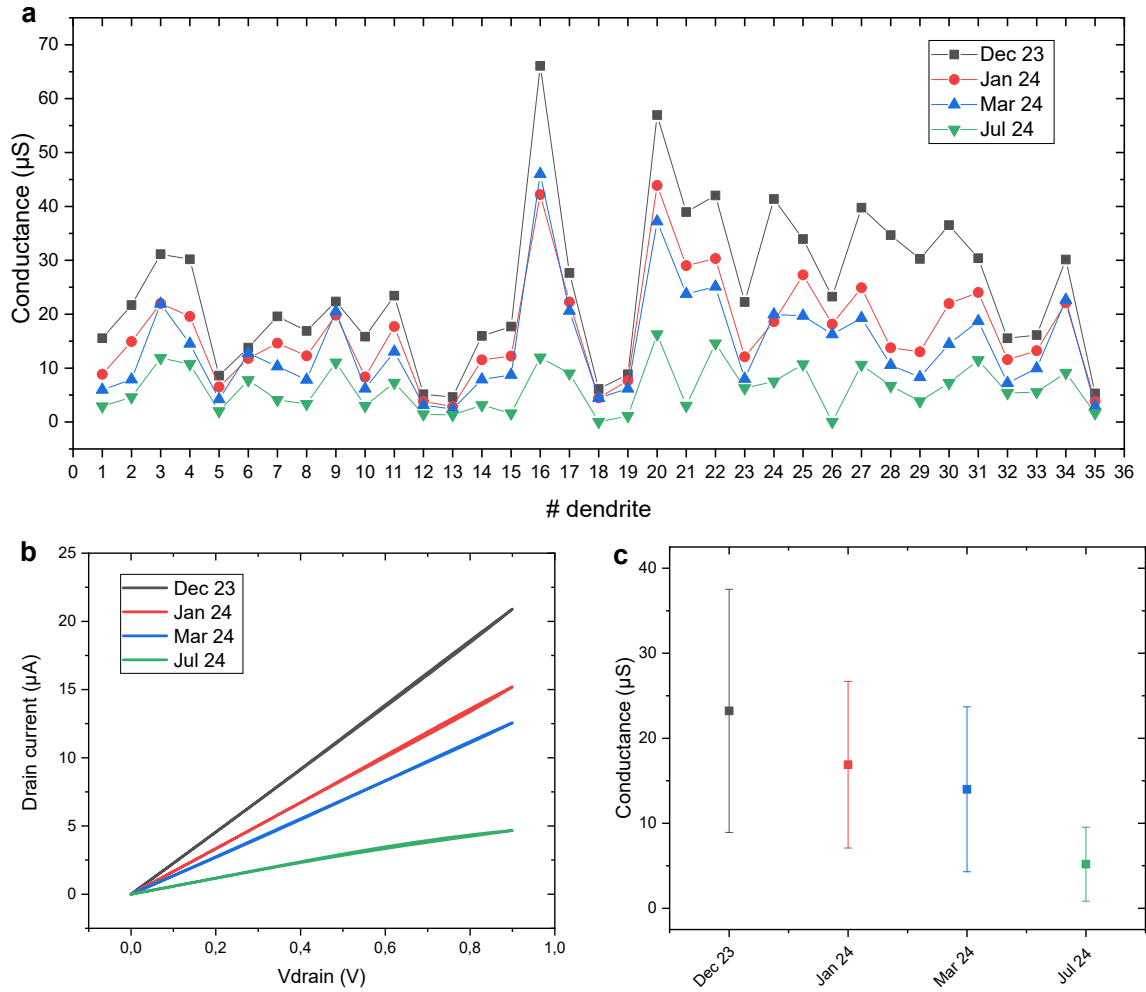


Figure 7.2: Long-term stability of PEDOT:PSS dendrites. **a)** Evolution in time of the conductance of all dendrites. **b)** Average output characteristics ($n = 35$). **c)** Average conductance and standard deviation ($n = 35$). Conductance was simply estimated as $G = I/U$.

III Electrophysiological recordings with dendritic electrodes

III.1 Cortical cell cultures

In order to assess the relevance of polymer dendrites for the realization of bioelectronic interfaces, cells were cultured on top electrogenerated PEDOT:PSS fibers (see MATERIALS AND METHODS) and their development was monitored during 21 days *in vitro* (DIV).

Network development

Figure 7.3 presents pictures of different cell cultures on MEAs with polymer dendrites taken at DIV 3. While Figure 7.3a-b show a high density of cells and a good development of the network, indicative of a healthy cell culture, the lower cell density in Figure 7.3c helps differentiating easily somas from neurites. Overall, cell development did not appear to be hindered by the presence of polymer dendrites. Note that biocompatibility tests had already been performed for the same MEAs within our team at IEMN, and that the biocompatibility of PEDOT:PSS has been assessed numerous times, so that the results were expected.

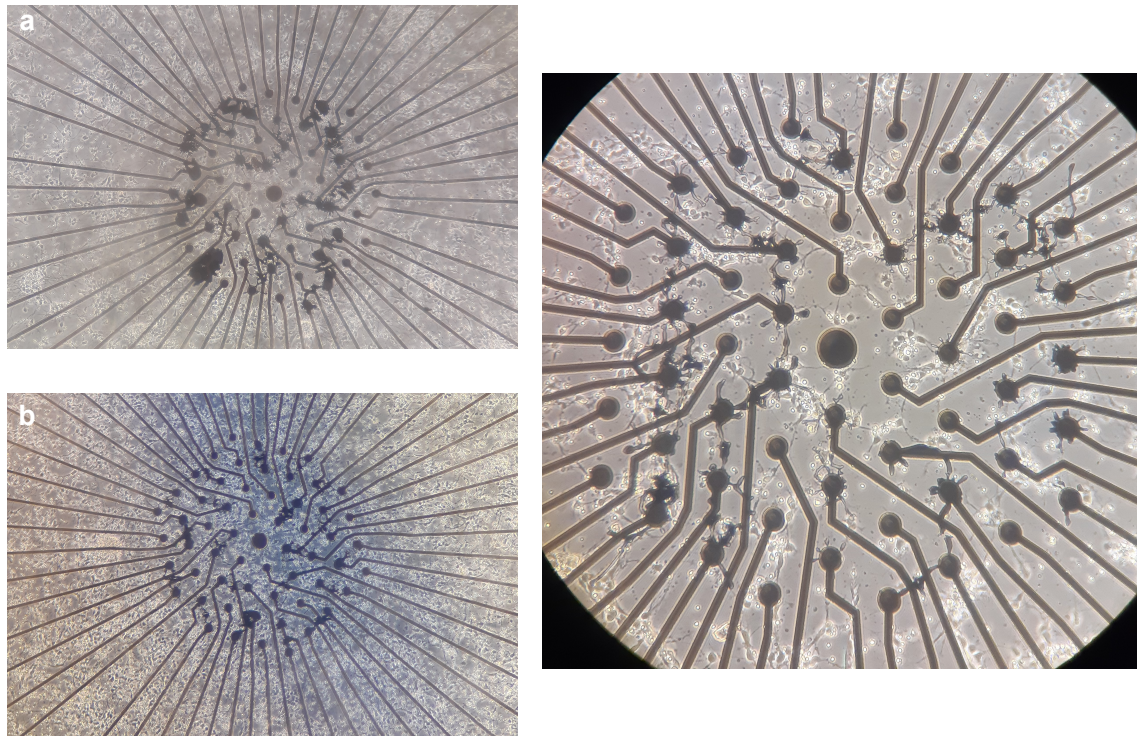


Figure 7.3: Cortical neuronal cells on different MEAs observed with an optical microscope. a-c) Microscopic observation of the development of three different cell cultures at DIV 3. The lower cell density in c) allows to better distinguish somas from neurites. In black between the electrodes, dendritic PEDOT fibers are visible.

At the end of the culture, cells were marked with a fluorescent probe to ensure viability. The samples were then visualized using fluorescent imaging to confirm

cell differentiation. Figure 7.4 shows that after 20 DIV and 21 DIV, cell network is established and neurons are well alive. Clusters nearby PEDOT:PSS fibers (see Figure 7.4a-b) indicate that the presence of artificial dendrites is not detrimental to cell development and network growth, as there is visually no difference between areas with and without polymer dendrites in terms of cell viability. Polymer dendrites therefore do not appear to be an obstacle for network growth.

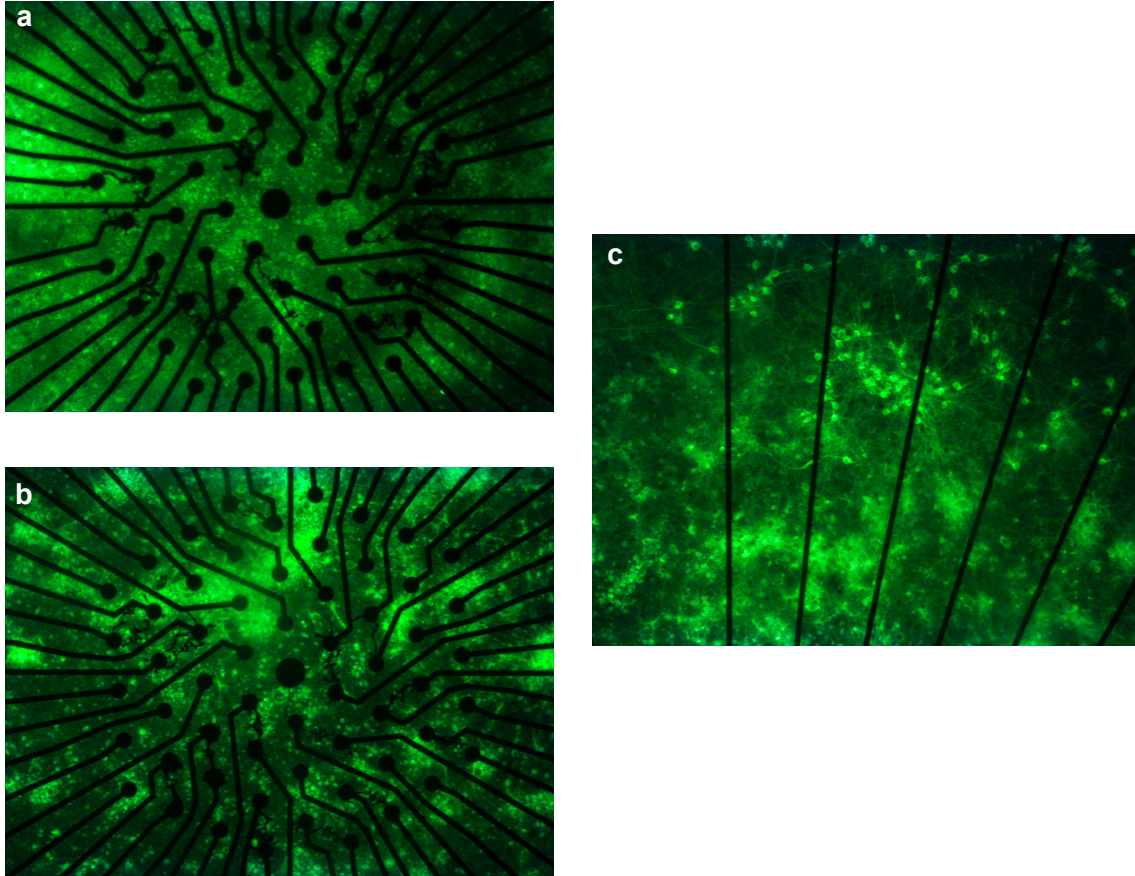


Figure 7.4: Fluorescence imaging of cell cultures with polymer fibers. Central area of the MEA where PEDOT:PSS dendrites are grown after **a)** 20 DIV and **b)** 21 DIV. **c)** The lower cell density in the periphery of the central area allows to distinguish somas and neurites with more clarity (21 DIV).

Finally, cell cultures were then fixed onto the MEAs with glutaraldehyde and prepared to be observed under electron microscopy. Some of the images are presented in Figure 7.5. These images revealed that when cell density was high, as it was the case for Figure 7.5a-c, PEDOT:PSS dendrites were almost completely covered by a layer of cells, except for the biggest protuberances such as the junction in the middle of the dendrites. Once again, this confirms our previous observation that electrogenerated fibers do not appear to prevent network development. With a lesser density of cells, artificial dendrites can better be appreciated (Figure 7.5d-f). Cells even seem to develop within the cracks of PEDOT:PSS fibers (Figure 7.5c), instead than on the parylene insulating film that otherwise covers the sample (see II in MATERIALS AND METHODS).

These images also offer valuable insights into polymer dendrite growth, as the nucleation points of PEDOT:PSS are revealed in the middle of the dendrite where junctions are broken apart (see for example Figure 7.5b for a zoomed-in image, and Figure 7.5d for a wider view). They also identify these junctions as structural weaknesses, as dendrites appear to easily separate at this point rather than anywhere else. While this separation might be influenced by the vacuum conditions required for SEM imaging, it still indicates that the junction between dendrites is not as robust as the rest of the fiber.

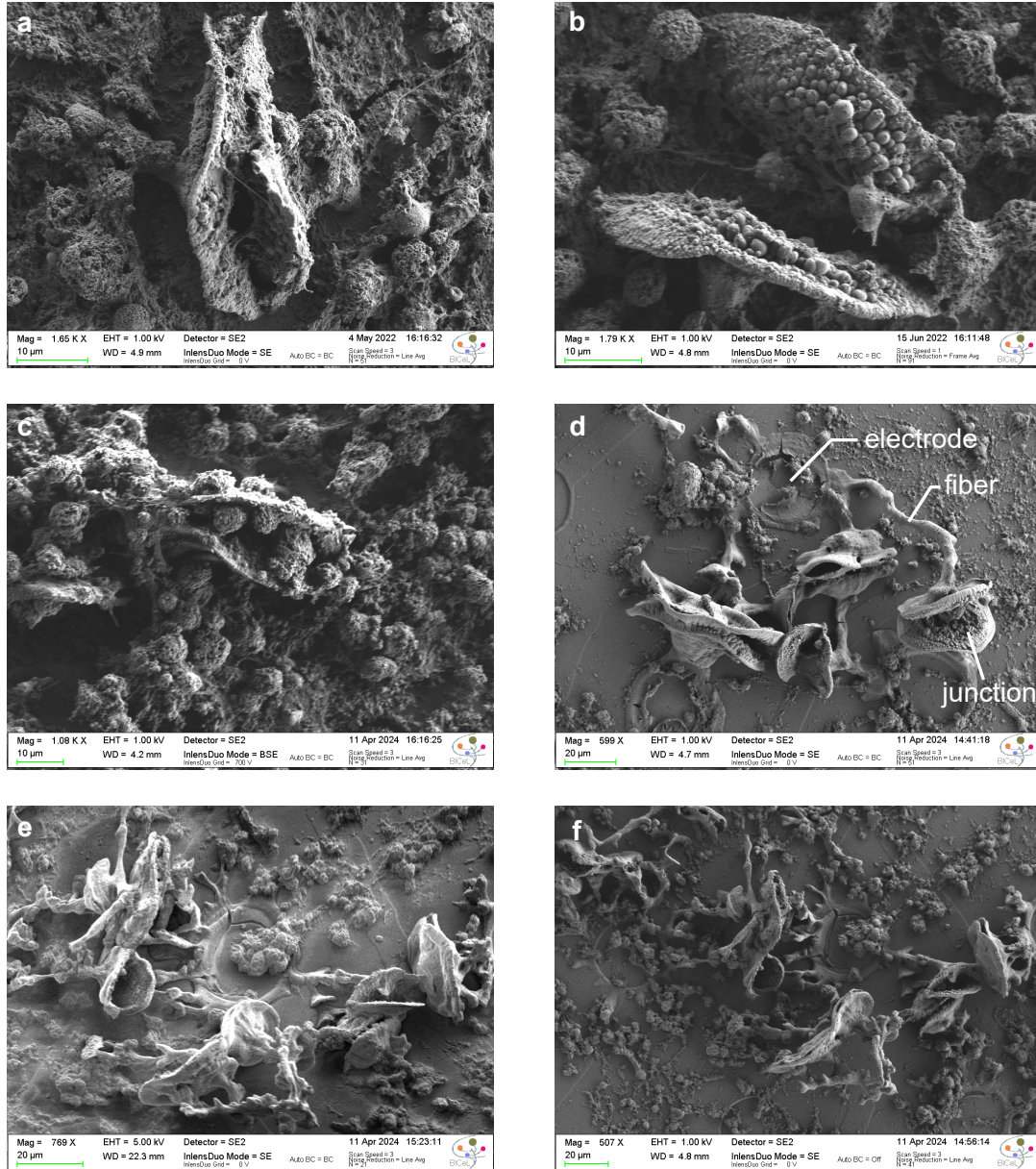


Figure 7.5: SEM images of cortical neuronal cultures on MEAs with PEDOT:PSS dendrites. a-c) A thick layer of cells covers the PEDOT:PSS dendrites, whereas a lesser density allows a better observation of the fibers in d-f). The annotations in d) show an electrode, PEDOT:PSS dendritic fiber and the junction of two of these fibers.

These different imaging modalities therefore made it clear that PEDOT:PSS

fibers do not prohibit neuronal development. In addition, electrical characterizations were performed while these PEDOT:PSS dendritic networks were in contact with cell cultures.

Electrophysiological recordings

PEDOT:PSS fibers were first tested as passive electrodes, and recorded as an usual MEA would be. Figure 7.6 presents some of the recordings obtained on a MEA where PEDOT:PSS dendrites had been grown prior to cell culture.

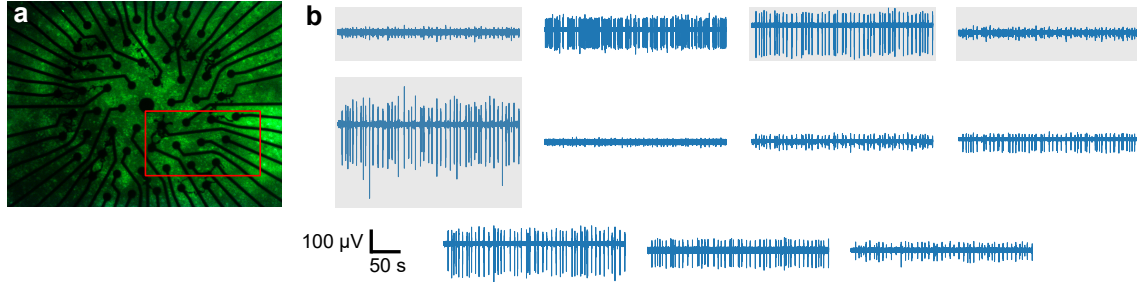


Figure 7.6: High-pass filtered recordings showing spontaneous activity from a primary cell culture on adjacent electrodes. a) The location of the electrodes is highlighted on the fluorescence image, and so are recordings from dendritic electrodes in b). The recordings were performed at DIV 20.

As it can be observed, neurophysiological activity was recorded on both dendritic and gold electrodes at DIV 20. Since action potentials form the basis of neuronal communication and are the most studied electrical signals in electrophysiology, this is where our attention was devoted. The raw signal was high-pass filtered to eliminate power line noise at 50 Hz and low frequency oscillations. Figure 7.7 shows a zoom on a burst and on a few spikes that have been recorded with a dendritic electrode.

Out of the 58 electrodes of the array (60 electrodes including two references), 47 detected at least five spikes over the recording period of 5 minutes. The events detected during the first 100 seconds of the recording are presented in the raster plot in Figure 7.8. The synchrony of the bursts is an indicator of the good development and maturity of the network.

On this sample, out of the 58 electrodes, 47 were considered active (meaning that they had recorded at least 25 spikes during a 5-minutes period), and polymer dendrites were grown out of 20 of these 58 electrodes (are only taken into account electrodes on which polymer dendrites were grown and through which a conduction current could be recorded prior to cell culture). Among these 20 dendritic electrodes, 19 detected at least 25 spikes during the recordings. On average on this sample, dendritic electrodes recorded a reduced background noise ($MAD_{dendrite} = 2.1 \mu V$ vs $MAD_{gold} = 2.6 \mu V$), which was expected as PEDOT:PSS fibers should lower electrode impedance. In addition, SNR was noticeably higher for dendritic electrodes (63.9 or 36.1 dB) than for usual ones (37.2 or 31.4 dB). After performing spike sorting (see MATERIALS AND METHODS) on this recording, 95 different neurons were

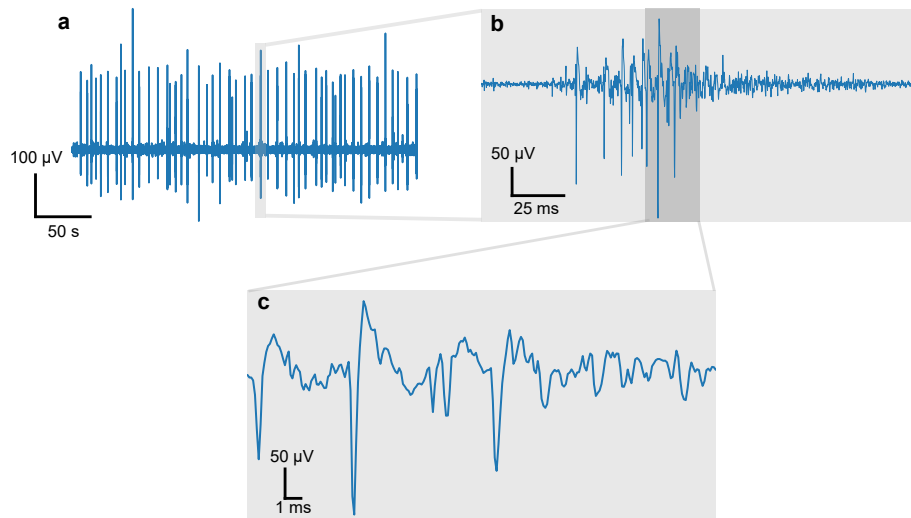


Figure 7.7: Electrophysiology with a dendritic electrode on a primary cell culture. a) 5-minute high-pass filtered recording of a dendritic electrode. b) Zoom on a burst and c) on a few spikes from a cortical neuron cell culture recorded with a dendritic electrode.

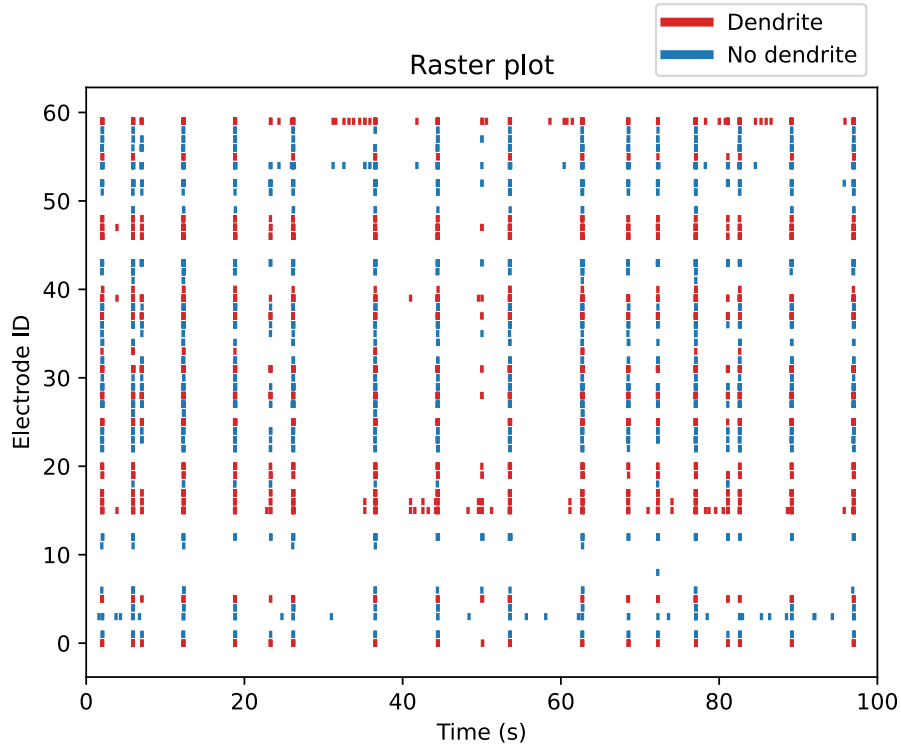


Figure 7.8: Raster plot of the first 100 seconds of the MEA with dendritic electrodes presented in Figure 7.6.

identified. The average number of neurons detected per electrode was also higher for dendritic electrodes (2.1) than for standard ones (1.4).

In order to assess if these differences were statistically significant, a Mann-Whitney U test was performed on both the number of neurons detected and the SNR value for each electrode. A Mann-Whitney U test is a non-parametric statistical test, meaning that it does not assume any particular distribution of the data prior to analysis. This test can be used to determine whether or not the distributions of two independent groups are different.

In the present case, two groups were defined: dendritic electrodes and gold electrodes, which are used as a control. In all cases, the p-value was lower than 0.05, meaning that the medians of the two populations were significantly different, both in terms of SNR and number of detected neurons. In other words, the differences observed between the two kinds of electrodes in this sample are not due to random statistical variations, but rather reflect true differences in the distribution of these parameters, depending on whether or not the electrodes are dendritic.

Although dendritic electrodes appear to present a slight advantage over standard gold electrodes, the previous results do not allow to conclude with certainty about their interest. Notably, since they have been compared with gold electrodes, a question remains unanswered: are the improvements only related to a decreased impedance of the dendritic electrodes compared to classic ones (in which case coating with a polymer thin film would offer the same advantages)?

III.2 Organotypic cultures

Although primary cell cultures are a very useful tool for the study of a large number of homogeneous cells in an isolated environment, they do not reflect well the normal behavior of neurons *in vivo*²⁸⁵. Since they come from a developing brain, organotypic cultures are more physiologically relevant than dissociated cell cultures, as they notably include more types of cells as well as an already developed neuronal network and complex interactions, which should therefore behave more closely to what would be expected from the nervous system. Because they preserve the 3D structure of the brain, organotypic cultures allowed to control the risk of a limited neural development due to the actual size of the polymer fibers that we feared could interfere with neurite growth.

Visualization of the brain slices

Following dissection, the evolution of a brain slice deposited on top of a MEA could be observed by microscope. Figure 7.9 shows a brain slice cultured on top of an array of electrodes with PEDOT:PSS dendrites. The same experimental protocol was followed as the one used for primary cell cultures. For a better comparison, on some of the MEAs used for brain slices, a PEDOT:PSS thin film was deposited on electrodes that had no dendrite (the electrodeposition protocol is described elsewhere^{137,138}).

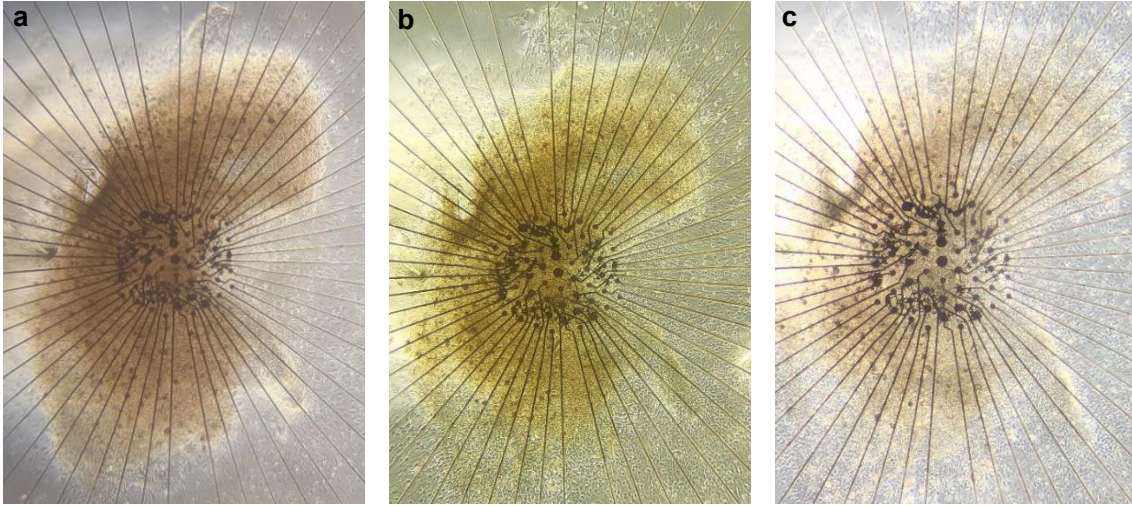


Figure 7.9: Spreading of the brain slice with time, at a) DIV 6, b) DIV 8 and c) DIV 9.

Cellular death was assessed using a fluorescent marker, as shown in Figure 7.10. Contrary to the fluorescence images provided for cortical cell cultures, in the case of brain slices the high density of cells makes it impossible to visualize viable cells correctly. Therefore, in Figure 7.10b, fluorescence indicates cellular death. The small number of cells highlighted by the fluorescence marker demonstrates a low cell death rate. In addition, microscopic visual inspection revealed that the brain slices were in good condition.

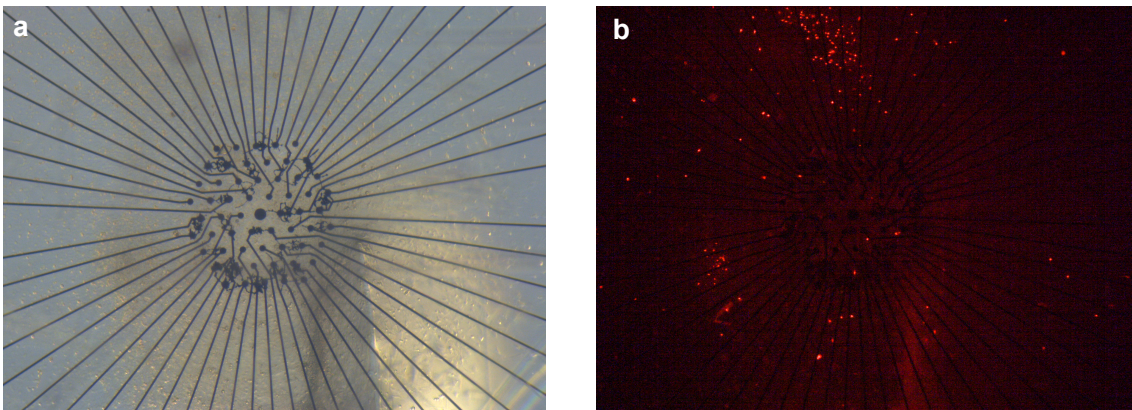


Figure 7.10: Cell death monitoring. a) Optical photo of a sample with a brain slice. b) Corresponding fluorescence image with propidium iodide at DIV 13. Dead cells are marked by fluorescence.

Electrophysiological recordings

Extracellular activity from organotypic cultures was recorded satisfactorily on two different MEAs. Figure 7.11 shows the activity recorded from a brain slice during a 5-minute period on adjacent electrodes.

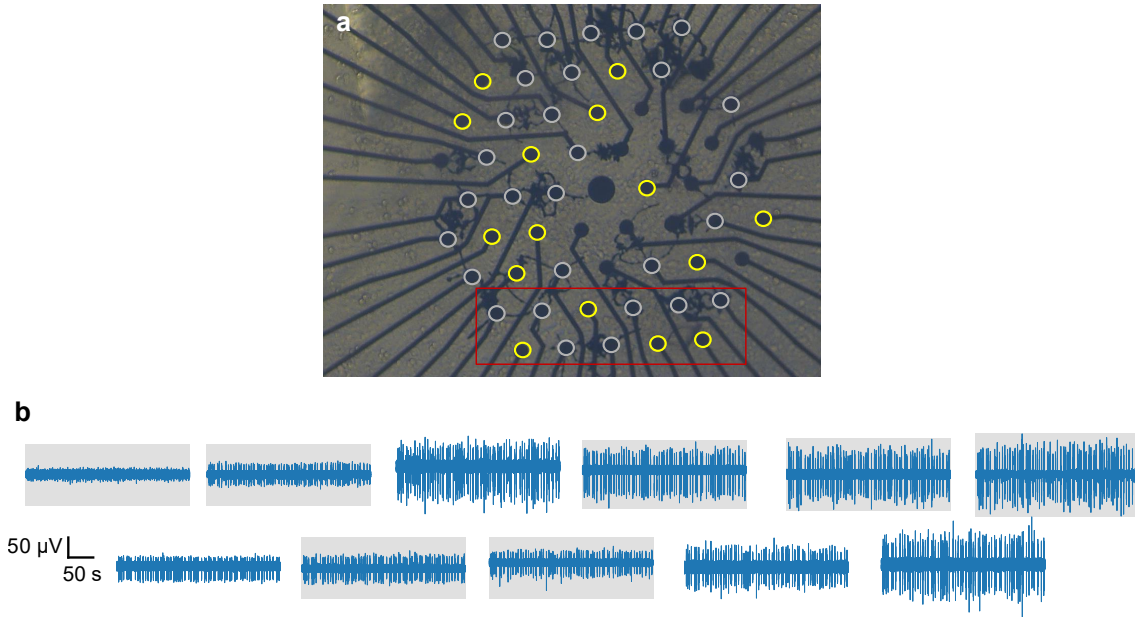


Figure 7.11: High-pass filtered recordings showing spontaneous activity from an organotypic culture (sample 1) on adjacent electrodes. a) The location of the electrodes is highlighted on the microscopic photo, and so are recordings from dendritic electrodes in b). Yellow circles indicate active electrodes without dendrite, and light gray circles indicate active dendritic electrodes. The recordings were performed at DIV 10.

As previously mentioned, raw signal was filtered to eliminate low frequency noise. Zooming in on the recording of a dendritic electrodes allows to visualize individual spikes, as shown in Figure 7.12.

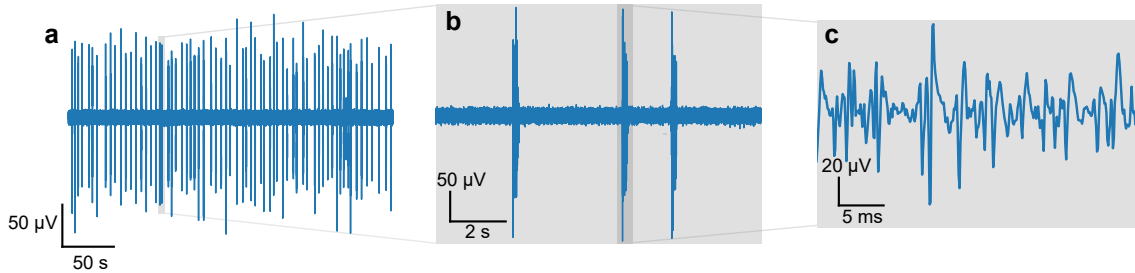


Figure 7.12: Electrophysiology with dendritic electrodes from an organotypic culture. a) 5-minute high-pass filtered recording of a dendritic electrode. b) Zoom on a burst and c) on a few spikes from an organotypic cell culture recorded with a dendritic electrode.

The brain slices on both devices appear to be in good health, as supported by the synchrony of the measured activity displayed in the raster plots in Figure 7.13. These plots also highlight the variability related to electrophysiological experiments, as it clearly appears that cell activity is much sparser for sample 2 than it is for sample 1. Even though the substrates were fabricated simultaneously, different slice cultures show distinct behaviors.

For both samples, dendritic electrodes represent a majority of the active electrodes (71% and 72%), although they only represent 64% and 60% of the electrodes

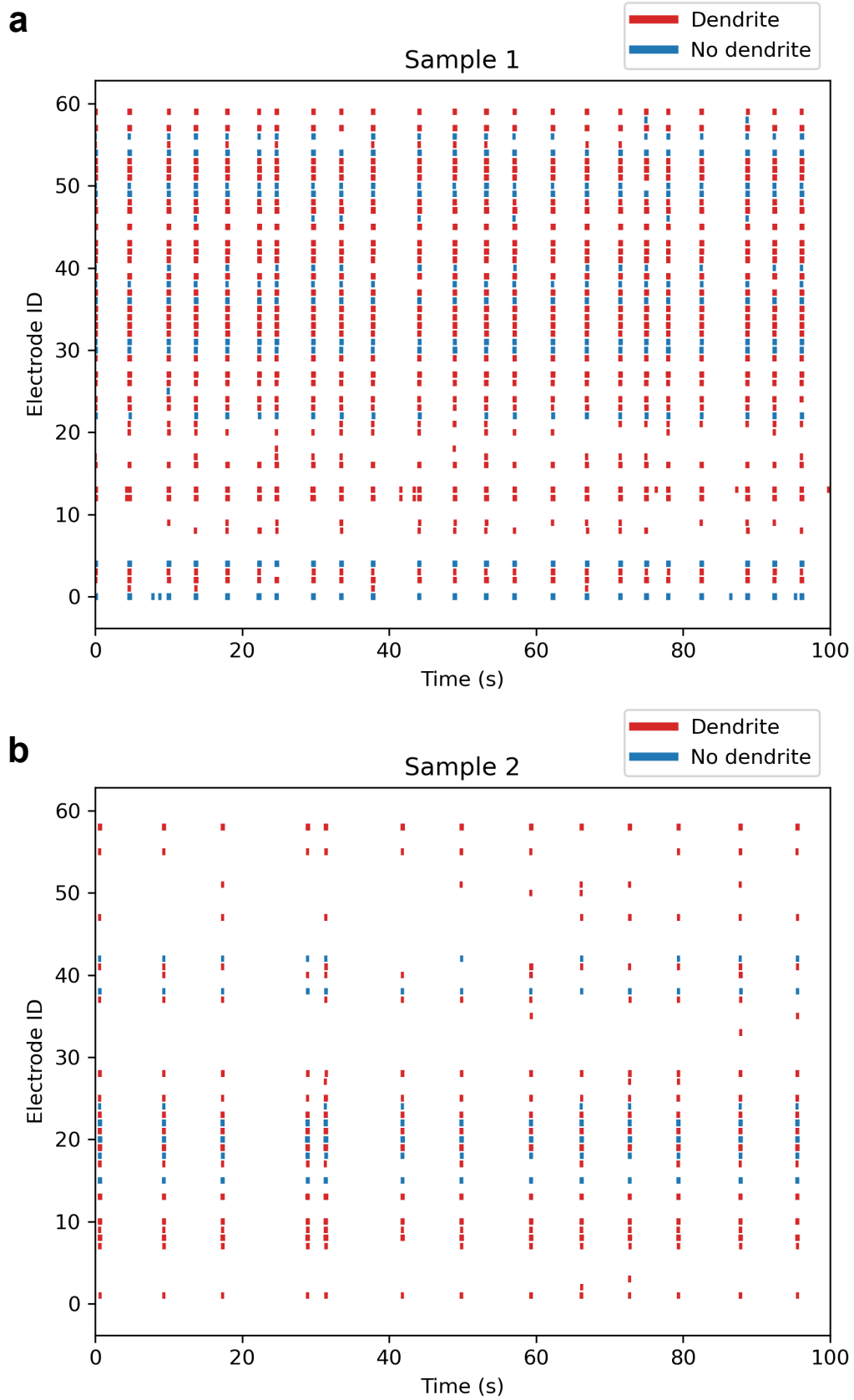


Figure 7.13: Raster plots of first 100 s of recording for a) sample 1 and b) sample 2. These event plots show the synchrony of the activity, which is a sign of a healthy neural network. The activity is much sparser for sample 2.

of the chips. As observed with cortical cell cultures, the SNR and the number of estimated neurons per electrode is always slightly superior for dendritic electrodes (see Table 7.1). In addition, to have a better point of comparison all the electrodes of sample 1 that had no dendrite were covered with a thin film of PEDOT:PSS to ensure that the differences with dendritic electrodes were not only due to the material but also to its morphology.

Statistical analysis, using a Mann-Whitney U test similar to the one performed for primary cell culture revealed that although the differences in SNR for sample 1 are statistically significant, they are not for the number of neurons detected per electrode. Regarding sample 2, the collected data do not allow to rule out the possibility that the observed differences might be caused by random statistical variations.

III.3 Summary of passive electrophysiological recordings

The important characteristics of all the passive recordings presented so far are summarized in Table 7.1.

	Active elec- trodes on the chip	Dendritic electrodes	Active den- dritic elec- trodes	SNR (no dendrite)	SNR (den- drite)	Neurons per elec- trode (no dendrite)	Neurons per den- dritic elec- trode
Organotypic culture 1	45	37	32	19.6 (25.8 dB)	36.3 (31.2 dB)	1.3	1.6
Organotypic culture 2	25	37	18	13.8 (22.8 dB)	17.9 (25.1 dB)	1.1	1.2
Cortical cell culture	47	20	19	37.2 (31.4 dB)	63.9 (36.1 dB)	1.4	2.1

Table 7.1: Electrophysiological recordings with dendrites. Significant statistical differences between dendritic and standard electrodes as determined with a Mann-Whitney U test are highlighted in green.

The results obtained for passive electrophysiological recordings with dendritic electrodes do not allow to assess with precision the influence of the dendrites on the recordings. The reported observations are mainly encouraging as the SNR appears to be significantly higher for dendritic electrodes even when compared to PEDOT-coated gold electrodes. In the case of the cortical cell culture, the number of detected neurons is higher with PEDOT:PSS fibers, which could back up the idea that the spatial extension of polymer dendrites increases the effective area of the electrodes. In addition, PEDOT:PSS fibers are more exposed to the cells of the culture than the electrodes which are approximately 1 μm deep inside a parylene well, potentially improving contact with the neurons²⁸⁶. Yet, organotypic cultures did not allow to validate this trend. In conclusion, these preliminary results point toward enhanced performance of dendritic electrodes, but more data would be needed to confirm this hypothesis and more work still needs to be achieved in order to confirm the benefit of dendritic electrodes and understand how they interact with neurons, as well as the impact of their morphology on the recorded activity.

IV *In vitro* transistor characterizations of PEDOT:PSS dendrites

As discussed in the introduction of this chapter and in the state of the art, the OEET has emerged as a building block of organic bioelectronics, in particular because of its large transconductance that allows local amplification^{173,192,195}. In addition, conducting polymers can easily be functionalized, which is a great asset for the detection of a vast array of biomolecules^{135,140,167,204–206,208}. Given that PEDOT:PSS dendrites do show transistor-like behavior¹⁴⁷ and that brain-inspired functions were demonstrated in the previous chapters of this manuscript, polymer fibers could lay the foundation of a new generation of smart interfaces that could be grown *in operando*, following a bottom-up strategy¹⁴⁶. In the following section, the *in vivo* transistor behavior of these objects will be assessed.

IV.1 Cortical cell cultures

Following sample preparation and passive electrophysiological recordings, PEDOT:PSS dendrites were characterized as OEETs through the recordings of both output and transfer characteristics.

Out of the 18 dendritic growths performed before cell culture, 11 of them were still operational after preparatory characterizations. Here, the 7 dendrites that were broken before cell culture were most probably damaged due to manipulation-induced stress, as the sample had to be rinsed several times and the dendrites were not grown in a network, but rather as a unique fiber between two electrodes, which based on previous experimental results always led to more fragile structures. At 20 DIV, a conduction current was still recorded across 10 polymer dendrites, although their conductance had decreased in every case. Figure 7.14 presents some instances of these *in vitro* characterizations. Even with a lower conductance, the drain current recorded in these dendrites was still clearly discernible above the background noise. On average, the drop of conductance reached about 78% on this sample, ranging from slightly less than 50% to over 88%.

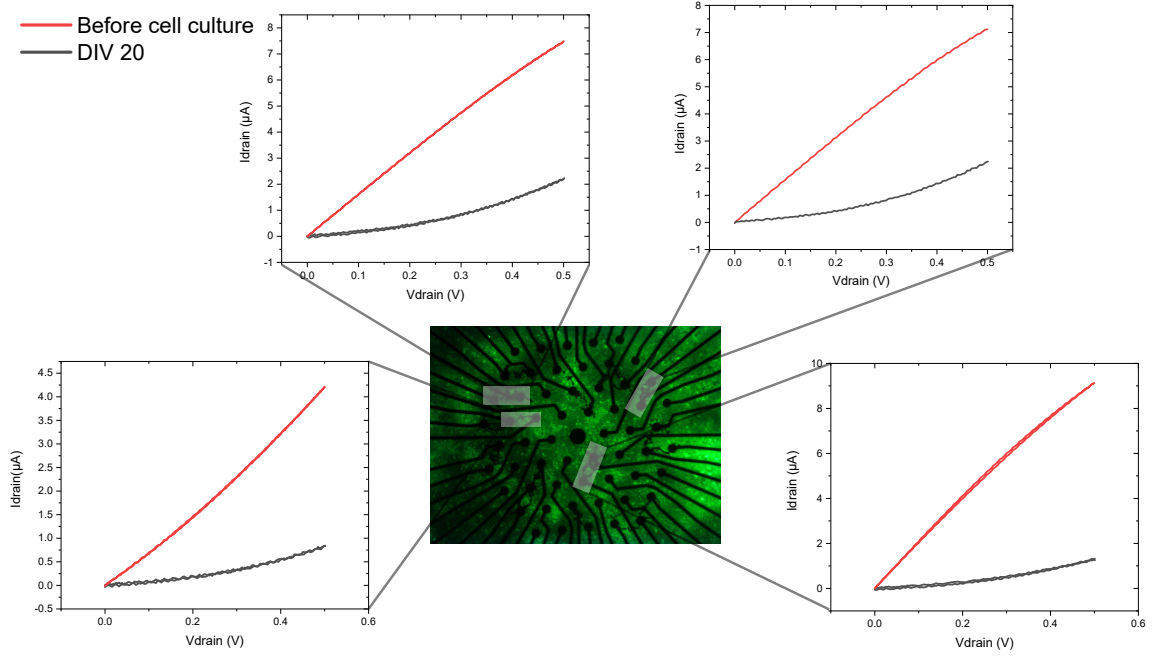


Figure 7.14: *In vitro* output characteristics of dendritic OECTs. In red, output characteristics before cell culture, and in black output characteristics recorded at DIV 20.

Additionally, as a silver wire had been added to the sample to act as an external gate, transfer characteristics could be recorded, as illustrated in Figure 7.15.

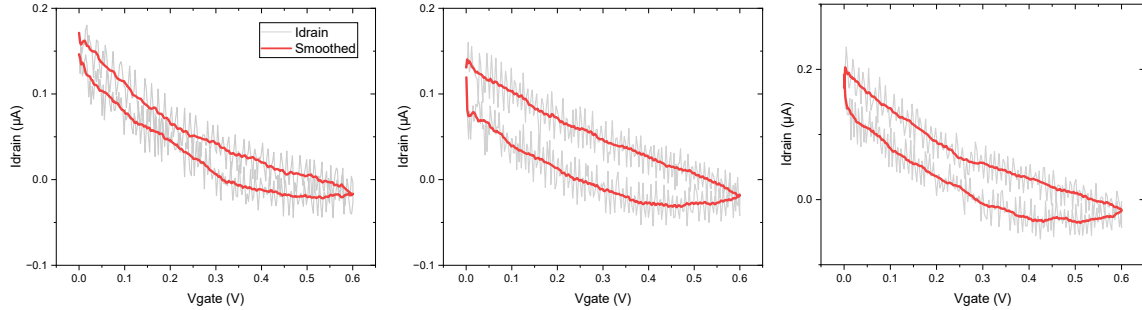


Figure 7.15: *In vitro* transfer characteristics of dendritic OECTs. The recorded curves (in light gray) were smoothed over 20 points (red curves). $V_{DS} = 100$ mV.

A clear decrease of the drain current along with an increase of the gate voltage can be observed. Yet, because of high level of noise in the corresponding gate currents (not shown here), the transistor effect cannot be verified, as the absence of faradaic current between the different terminals of the polymer dendrite cannot be confirmed. In addition, for the three cases given as an example, the drain current appears to become negative for a large enough V_{GATE} . However, cell culture medium is a noisy environment, which could explain both the high level of noise in the recordings and the negative I_{DRAIN} .

All in all, it appears that dendritic interconnects still operate as organic electrochemical transistors when interfaced with living tissue.

IV.2 Organotypic cultures

As we did for primary cultures, the dendrites were also characterized as transistors after organotypic culture. As it was observed for primary cell cultures (see Figure 7.14), conduction current was still recorded through the PEDOT:PSS fibers at DIV 9 and an important drop of conductance was recorded for all dendrites, as illustrated in Figure 7.16. As can be seen in the zoom on the data at DIV 9, these measures also came with much more noise than what was previously observed, most likely due to a change in the software used to perform measurements and a caliber issue.

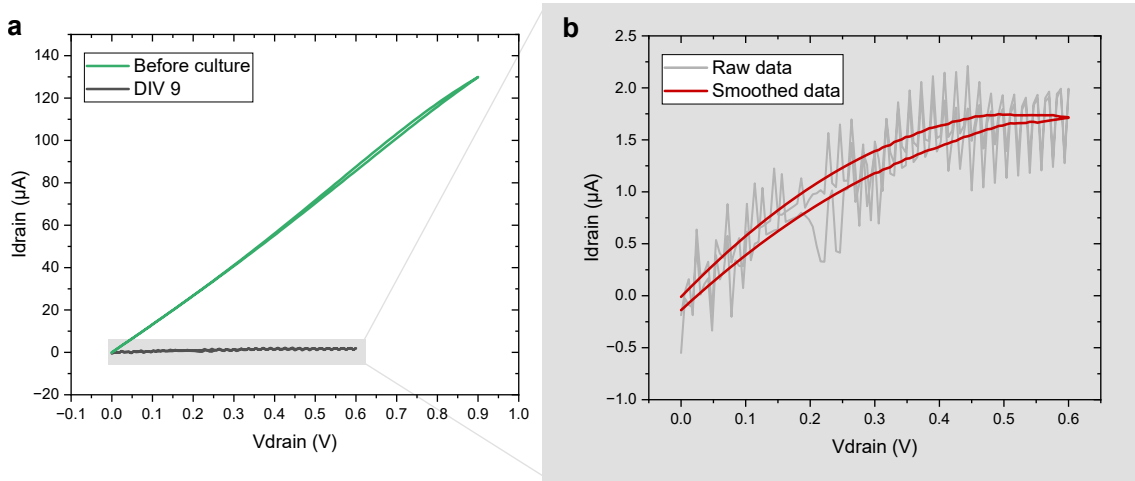


Figure 7.16: Output characteristics of a dendritic OECT from an organotypic culture. a) Comparison between before culture and DIV 9 (sample 2). b) Smoothed output characteristic. The maximum potential difference was kept at 0.6 V during cell culture to disturb as little as possible the brain slice.

If the dendrites were not completely broken, yet it appears that the organotypic culture caused them more damage than the primary culture did. It could be due to mechanical friction related to the deposition of the slice on top of the dendrites. However, because of the low conductance of the fibers and the high level of noise, no transfer characteristic could be recorded. Yet, as Figure 7.16b illustrates, a plateau can be observed in the output current of the dendrite around 0.5 V, which could be attributed to a saturation effect and thus a transistor-like behavior. Although it cannot be once again assessed with absolute certainty that PEDOT:PSS dendritic fibers conserve their transistor properties after cell culture in the light of these experiments, some hints still point in this direction.

IV.3 Conclusion

Transistor characterizations were not entirely conclusive during these experiments, as the dedoping of the PEDOT:PSS fibers with a positive gate voltage was not observed with absolute certainty. Nevertheless, conduction current was still recorded from the polymer fibers after both cortical cell and organotypic cultures, and qualitative analysis of the recordings suggests that the OECT behavior of PEDOT:PSS dendrites was conserved. One thing was clear: cell cultures had an important impact

on the conductance of the artificial dendrites, as this parameter showed an important decrease for all the recorded characteristics. Whether it is due to mechanical constraints and damage, reorganization of the fibers or chemical dedoping related to the culture remains up for discussion.

Once can also note that no spike was detected in the curves presented in this section, although activity was recorded for these samples (see Figure 7.8 and Figure 7.13). This could be due to two reasons. First, because of the low conductance of the objects after cell culture, the integration time of the equipment might be superior to the duration of a spike, effectively screening the action potential in the resulting curves. Second, the expected amplitude of a detected spike could be too small to be observed in the high level of noise reported in the previous experiments.

In conclusion, the data presented in this section suggest that PEDOT:PSS dendritic fibers keep operating as OECTs even after cell culture, but more work is needed to confirm this observation.

V Discussion

The electrophysiological experiments presented above, while not entirely satisfactory, raised two important questions, regarding the choice of the reference electrode and the size of the electrogenerated objects.

V.1 About the importance of the reference electrode

Working with gradually more complex polymer dendrite networks raised concerns about the impedance of such networks taken as a whole, and the repercussions it could have on the electrophysiological recordings. Microstructuration of the electrodes by coating with a thin layer of polymer has proved to decrease the impedance of the electrodes¹³⁷, and PEDOT:PSS dendrites even have the potential to further decrease it by increasing the total effective surface of the electrode due to their spatial extension.

Figure 7.17 compares the modulus of the impedance of dendritic electrodes for different morphologies with gold electrodes coated with a PEDOT:PSS thin film (electrodeposited following the method described in MATERIALS AND METHODS). As it can be observed, the average impedance of a gold electrode coated with a polymer thin film is more important than that of any of the dendritic electrode over the whole frequency range of the experiment. The impedance modulus also seems to be correlated with the quantity of polymer available in each cluster: the highest impedance among dendritic electrodes is obtained for the 1 kHz dendrites, which are also the thinnest. The lowest impedance in this sample was attributed to the 300 Hz cluster, which was one of (if not the) densest cluster.

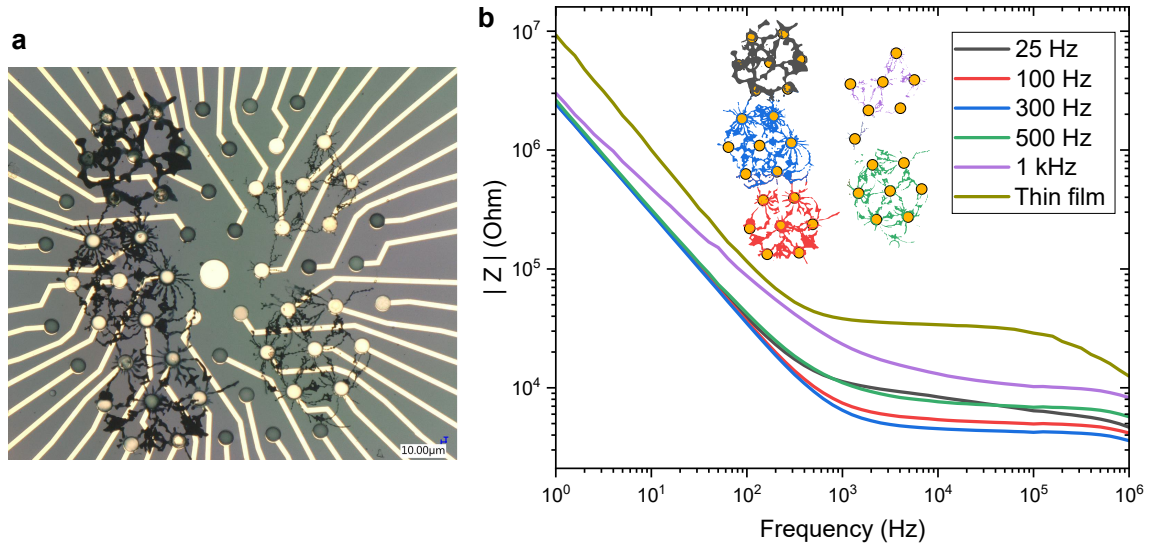


Figure 7.17: Impedance of dendritic electrodes as part of a network. a) Microscopic photo of the polymer dendrite clusters, grown at different frequencies to obtain distinct morphologies (clockwise from top right: 1 kHz, 500 Hz, 100 Hz, 300 Hz, 25 Hz). b) Average impedance of the dendritic electrodes, compared to electrodes coated with a DC electrodeposited PEDOT:PSS thin film. In the inset figure, the color code indicates the growth frequency of the dendrites. The reference used is a 1 mm diameter gold electrode integrated in the MEA.

Interestingly, below 100 Hz, it can be noted that the impedance modulus of the electrodes with dendrites grown between 25 Hz and 500 Hz superpose. Such a perfect coincidence appeared suspicious, as the structure-properties relationship established all along this manuscript would suggest that such distinct morphologies should correspond to so many different curves. In fact, the common denominator to all these curves was the 1 mm diameter gold electrode that had been used as a reference for all the measurements. New experiments were thus run, using this time an external Ag/AgCl electrode as the reference. The results are reported in Figure 7.18.

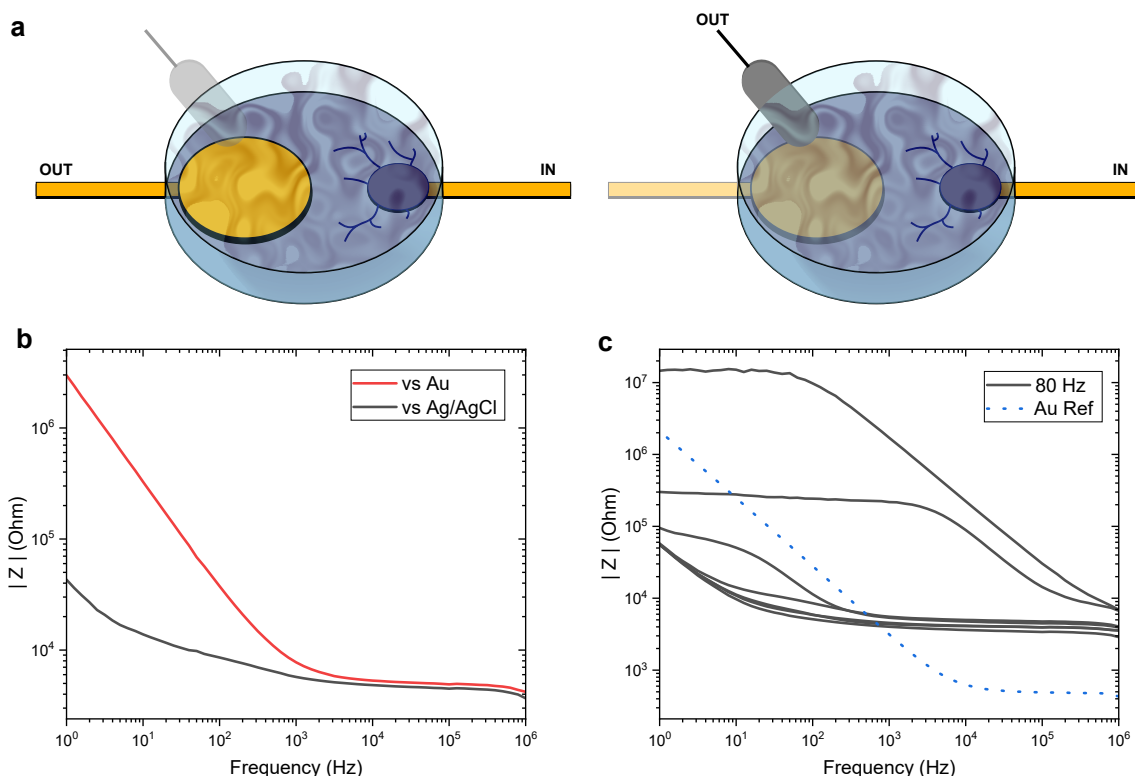


Figure 7.18: Influence of the reference electrode on electrochemical measurement impedance. a) Setup for the electrochemical impedance measurement, using either the in-plane gold electrode of the MEA or an external Ag/AgCl wire as the reference. b) Impedance modulus of a dendritic electrode vs the 1 mm diameter gold reference of the MEA (red), or vs an external Ag/AgCl wire (gray). c) Comparison between the impedance modulus of dendritic electrodes with growth frequency of 80 Hz (grey) and the impedance modulus of the reference electrode of the MEA (blue dotted line), all against an external Ag/AgCl wire.

In Figure 7.18b, the impedance modulus of a dendritic electrode vs Au shows an almost perfect 1-decade-per-decade slope, which corresponds to an ideal capacitance, a surprising behavior for such a system where constant-phase elements would rather be expected^{245,287}. However, when the reference electrode is changed to an external Ag/AgCl wire instead of the in-plane gold reference of the sample, the low frequency part of the curve undergoes major modifications, and its impedance is greatly decreased. Above 1 kHz, the two curves superpose, suggesting that the same physical process is at the origin of the recorded impedance modulus in both cases. Therefore, it appears that at low frequencies, the impedance of the dendritic electrode becomes lower than that of the gold reference, inferring that the reference now screens the low frequency behavior of the system. This phenomenon can be better appreciated in Figure 7.18c, which displays the impedance modulus of different dendritic electrodes grown at 80 Hz, as well as the impedance modulus of the gold reference electrode. These figures show that, when working with electrodes of low impedance such as dendritic electrodes, the choice of the reference electrode turns out to be absolutely crucial. In fact, it might well become a bottleneck for electrophysiological recordings, as Figure 7.18c suggests that around 1 kHz, its impedance might already exceed that of the recording electrodes, potentially resulting in the distortion of the electrophysiological signal and affecting the quality and accuracy

of the recordings.

V.2 Toward miniaturization?

These results put in perspective the high volumetric capacitance of PEDOT:PSS^{177,288}, as we see here that also polymer dendrites cannot be observed by the naked eye (the distance between two adjacent electrodes equals 100 μm), their capacitance still exceeds that of a 1 mm diameter gold electrode, which would explain the lower impedance as $|Z| \propto 1/C$. One solution to rebalance the impedance between dendritic electrodes and the reference could thus be to downscale the dendrites, an idea that will be discussed in the final chapter of this manuscript (see DISCUSSION AND OUTLOOK).

Growing thinner and shorter dendrites will lead to a decrease in the capacitance that will in consequence increase their impedance. In the state of the art, the equivalent model of the neuron/electrode interface was discussed, as recalled in Figure 7.19.

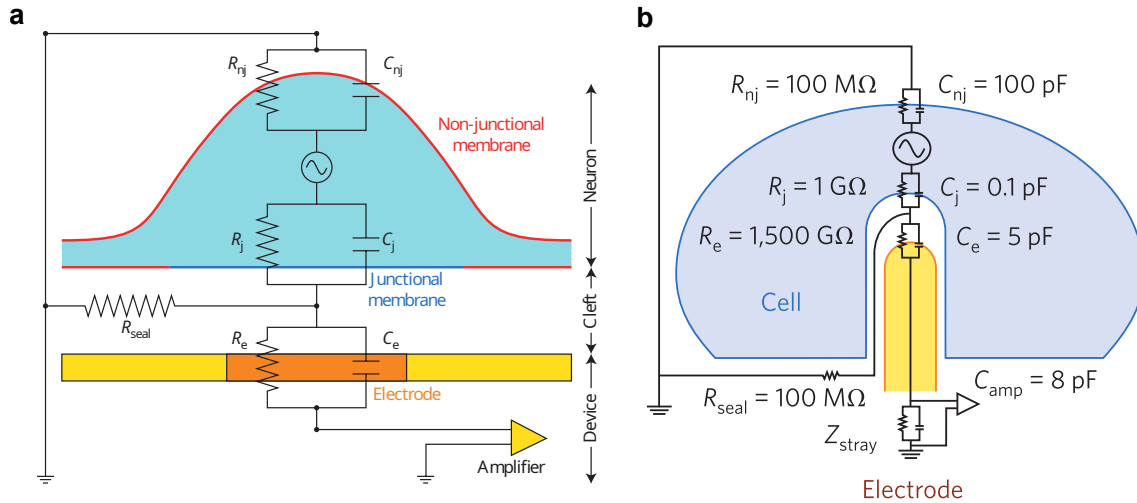


Figure 7.19: Equivalent model of the electrode/neuron interface. a) Point-contact model of the neuron/electrode interface. This representation works both for a passive electrode as well as an active device. b) Order of magnitude of the values of the different parameters of the model. Adapted from [89].

Yet, some considerations differ depending on whether the recording device is a passive electrode or an active transistor. Indeed, in the case of a passive device, the capacitance of the electrode C_E should be as high as possible to limit the drop of voltage at the electrode/electrolyte interface and maximize it between the terminals of the amplifier. However, for an active device such as an OECT, the amplifier is integrated in the device itself. In this case, C_E should then be low to maximize the drop of voltage between the channel of the transistor and the electrolyte.

Mona Ghazal, who spent 6 months in our group as a trainee working on the electrical modelling of the cell/electrode interface, showed that two parameters can help enhancing the coupling between neurons and the transistor channel²⁸⁹, namely the seal resistance and the capacitance of the channel (see Figure 7.20).

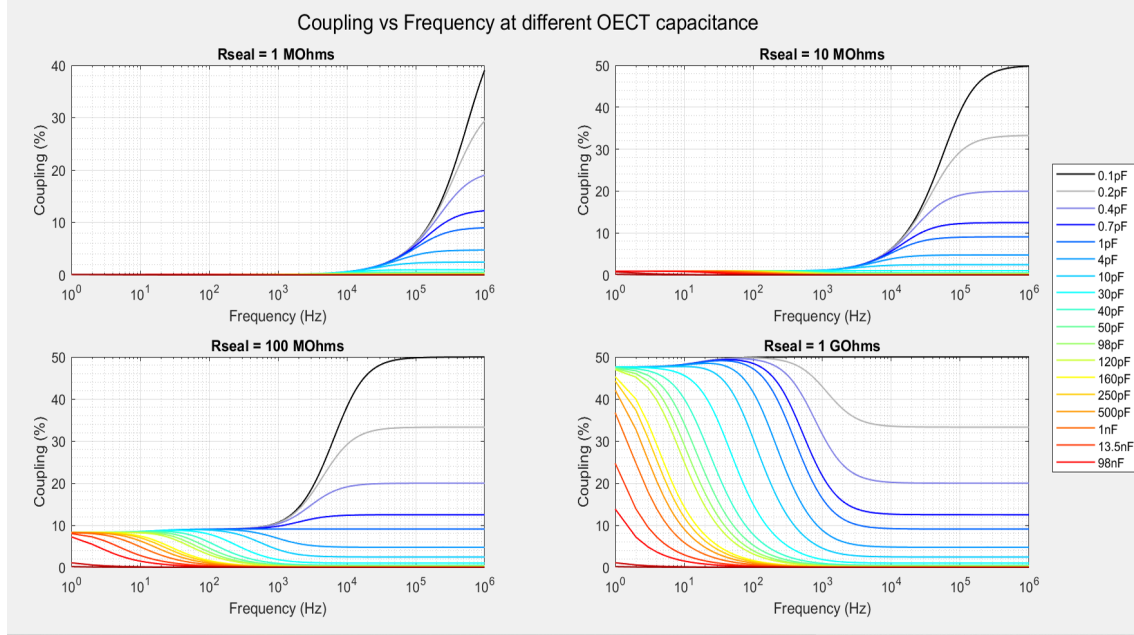


Figure 7.20: Coupling between the neuron and the OECT for different values of the capacitance of the OECT. The equivalent model and the values of the different parameters are based on the model described in Figure 7.19.

Her work demonstrates that first and foremost, the seal resistance R_{seal} is the main parameter affecting the coupling, and that its value should be as high as possible to avoid current leakages returning to the culture medium⁸⁹. Marginally, and for high enough R_{seal} , she demonstrated that decreasing the capacitance of the transistor channel should further increase the coupling.

Decreasing the size of polymer dendrites should therefore comply with both requirements, as in addition to limiting their capacitance it could also allow neurons to engulf them, as it was proposed with mushroom-shaped microelectrodes²⁹⁰, potentially increasing the coupling.

VI Conclusion

The results presented in this chapter provide valuable insights into electrophysiological recordings with dendritic electrodes, but these conclusions are preliminary, and further research is required to confirm these trends. On the one hand, we have demonstrated the detection of spikes in a biological environment with dendritic electrodes, both from cortical cell cultures and brain slices. Our findings suggest that dendritic electrodes could present a higher SNR and detect a greater number of neural units than gold or PEDOT-coated electrodes, although additional data is needed to confirm this hypothesis. We have also shown that after cell cultures, a conduction current was recorded through PEDOT:PSS fibers, and the transistor effect still appears to be observed, although our experiments cannot confirm that point with certainty. On the other hand, the conductance of the polymer dendrites was in all cases highly affected by the culture. The transistor characterization of our devices did not yield much success due to the significant decrease of the conductance of

the dendrites and the inherently high noise level of biological medium. Technical difficulties also limited the outcome of our study. Apart from the hardware itself, biological experiments introduce a high degree of variability that complicates data analysis and makes it difficult to compare the results from different samples (lack of repeatability, limited network development, premature cellular death. . .).

In conclusion, the evidence presented in this chapter supports the idea that polymer dendrites could be an efficient alternative to the other devices currently used in electrophysiology, but more in-depth analysis is required to validate this hypothesis.

8

DISCUSSION AND OUTLOOK

The final chapter of this manuscript will give us the opportunity to take a step back and reflect on the research work that was realized in the context of this thesis. Drawing inspiration from the operation and structure of the brain, the main objective pursued during this work was to explore new ways to perform computing, using the possibilities offered by electropolymerization, an alternative fabrication technique that generates unique objects with singular morphologies that give rise to complex structure-function relationships. In chapter 4, we explored the chemical environment of dendritic growth and studied the influence of the composition of the electrolyte on the morphology and electrical properties of polymer dendrites. We then delved into the computational power offered by these peculiar morphologies to perform tasks that more traditional hardware could not implement. In chapter 5, classification of voltage bursts was demonstrated with polymer dendrites. By monitoring the current on both the source and drain of a dendritic transistor, it was shown that the polarity as well as temporality of these bursts could be retrieved without additional processing. We then showed in chapter 6 that network of polymer dendrites are actually very interesting candidates for the implementation of morphology-dependent hardware with spatiotemporal processing abilities due to the complex interactions that arise between form and functions in these devices. This chapter really settled electropolymerization as a fabrication method on its own, demonstrating the unrivaled computational power offered by the intrinsic stochasticity of the method. Finally, we saw in chapter 7 how such structures could potentially be employed in the context of bioelectronics.

In the following sections, I will propose future directions and perspectives for the work I led during those past three years.

I Towards *in vivo* electrophysiology

In the previous chapter, we entered the realm of bioelectronics through 2D *in vitro* electrophysiology. Although this is a good starting point for working with living cells, *in vitro* experiments are known to be of little bio-realism²⁸⁵. Primary cell cultures for example present a limited number of cells types compared to what is observed in living organisms. If brain slices allow to get closer to *in vivo* conditions, they still fall short in mimicking the full complexity of a living brain's environment.

To get more insights on the potential of polymer dendrites and dendritic networks, *in vivo* experiments definitely appear to be necessary. This however requires a review of the platform that is used, especially to transfer artificial dendrites onto a flexible substrate. By doing so, these structures could be better integrated with living tissue, ensuring that they can withstand the dynamic and complex environment of a living organism, ultimately paving the way for more accurate and relevant bioelectronic applications. Growing polymer dendrites on flexible substrates could enable the development of smart interfaces with *in materio* processing abilities, such as the ones described throughout this manuscript.

I.1 Mechanical frailty

Given the dimensions of a brain (even a rodent one), its curvature should not really cause any issue for micrometric dendrites. However, the mechanical frailty of such objects was still assessed qualitatively. A freestanding dendrite was thus grown following the indications of MATERIALS AND METHODS. Its drain current was then recorded, applying a small potential difference between its terminals, while the gold wires serving as electrodes were gradually being brought closer together, starting 240 μm apart, as depicted in Figure 8.1.

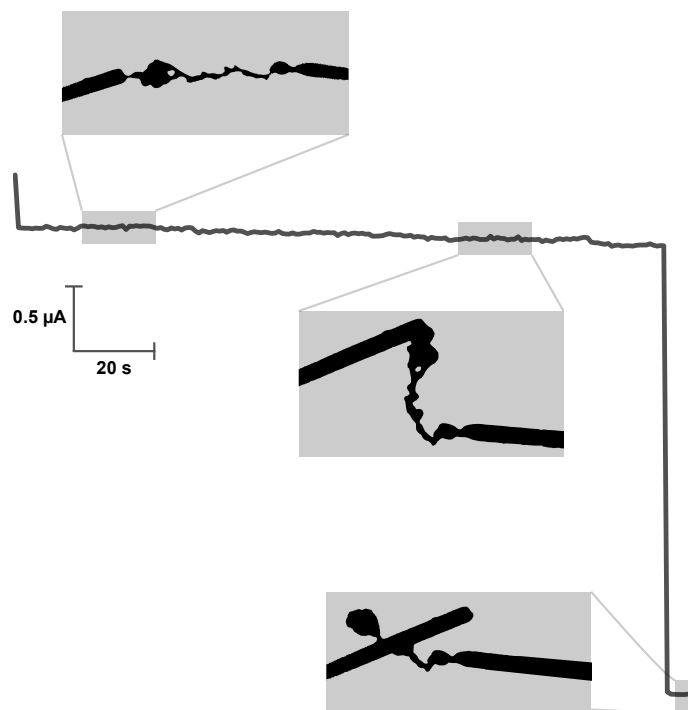


Figure 8.1: Mechanical frailty assessment of a freestanding dendrite. The I-t curve was being recorded while the two electrodes were gradually brought closer together until the dendrite snapped. The inset figures show the dendrite being bent.

The I-t curve shows that the output current of the system does not appear to be affected much by the bending of the dendrite, as it remains almost constant throughout the duration of the experiment (about 2 min), although the bending of the dendrite can clearly be appreciated in the second inset figure. Instead of being gradually damaged, the dendrite broke all of a sudden, which is shown in the video of the experiment as well as in the recording of the output current. This experiment leads us to believe that the dendrite can undergo some mechanical deformation without any significant modification of its electrical behavior, which was a good sign for the development of a flexible device.

I.2 Fabrication of a flexible device

As the technology was available and already mastered at IEMN, polyimide (abbreviated PI, HD4110 Photodefinable polyimide coating) was chosen as the base material for the flexible substrate. No less importantly, PI is an organic polymer that demonstrate desirable mechanical properties in addition to being biocompatible²⁹¹.

Figure 8.2 summarizes the fabrication process of the polyimide-based flexible devices that I developed in anticipation of *in vivo* electrophysiology experiments, which can be divided into four main steps. The design of the device presented in the following pages was freely inspired from the work of Vomero *et al.*²⁹². Due to availability, ease of utilization and cost effectiveness, the connector used was the HiRose FH26W-17S-0.3SHW60, which technical documentation will be available in Appendix. I would also like to show my appreciation to Dr Ahmad Itawi for his precious advice regarding the development of these flexible devices.

Fabrication of the polyimide flexible substrate

After basic cleaning, a 500 nm aluminum sacrificial layer was deposited on a 3" silicon wafer. Following HMDS deposition, PI was then spin-coated on the wafer. Three consecutive spins were performed to ensure maximum uniformity of the PI film. The first spin was realized at 500 rpm to help with the spreading of the film, before increasing the rotation to 1000 rpm and 2500 rpm in order to attain the desired thickness. After a short rest period of about 10 min used to let air bubbles escape, PI was soft baked on a hot plate at 90°C, slowly approaching the wafer (40 s at 2 mm, then 40 s at 1 mm and 180 s in contact with the plate), before a second soft-bake at 110°C for 3 min.

Following PI spin-coating, the devices were patterned via UV-photolithography and developed in cyclopentane for 2 min before rinsing in PA400R.

Finally, PI was cured, ramping up the heat from room temperature to 200°C at a speed of 10°C/min, then stabilized for 30 min before undergoing a second ramp up to 350°C and spending 60 min in the furnace.

Deposition and patterning of the passivation layer

To encapsulate the gold interconnects and for electrical isolation, a 1 μm parylene-C layer was grown on the substrate through chemical vapor deposition. The openings

in the insulating layer were then defined via UV-photolithography (using the AZ 1512 photoresist) and the parylene layer was etched with O_2 Reactive Ion Etching (RIE).

Device separation

Device separation was achieved by immersing the wafer into MF-319 developer for approximately 2 days, renewing the solution after the first day.

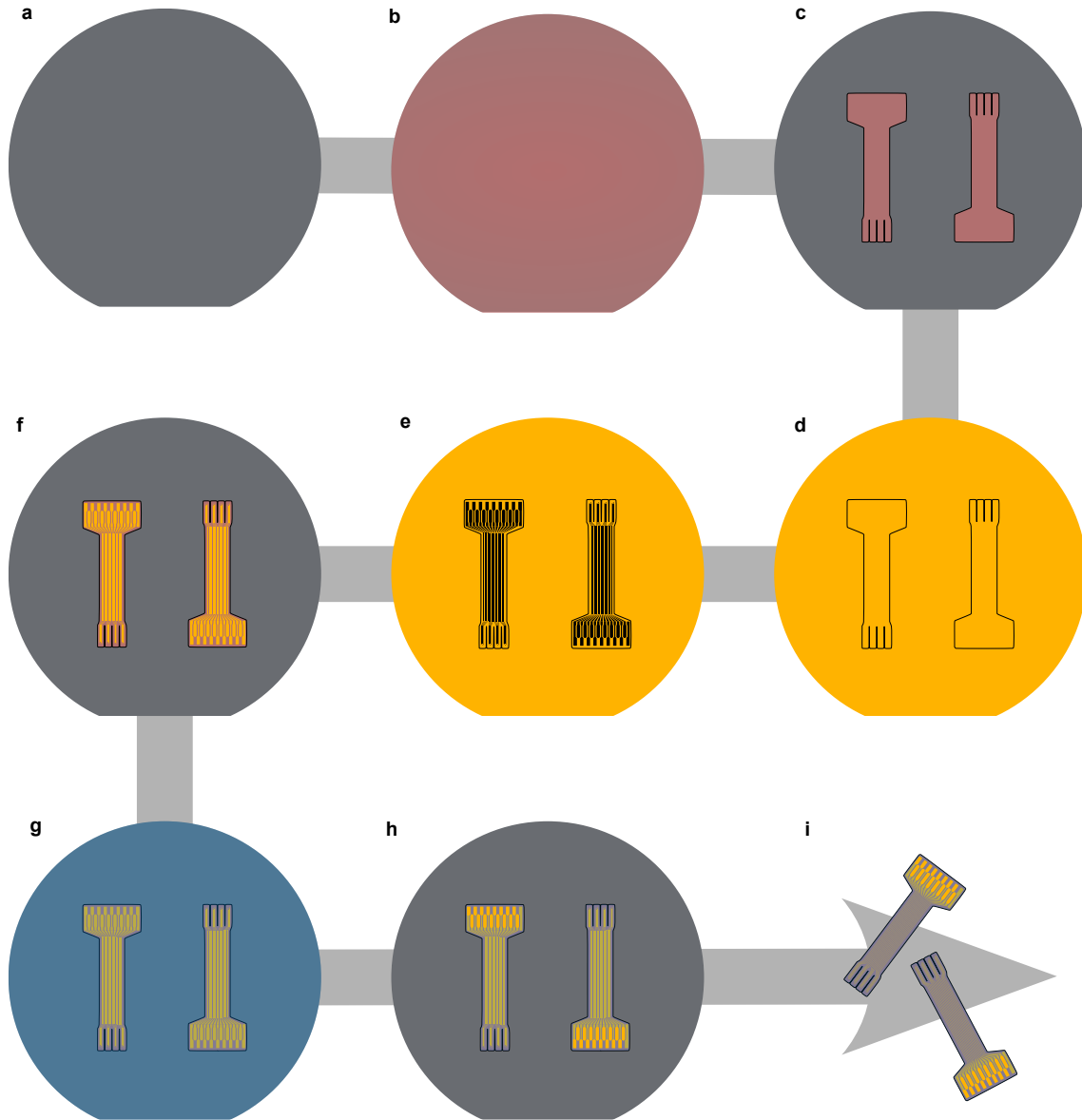


Figure 8.2: Fabrication process of a flexible device. a) Wafer cleaning and deposition of an Al sacrificial layer. b) Spin-coating and softbaking of the PI layer. c) PI patterning via UV-photolithography. d) TiAu deposition. e) Electrode patterning via UV-photolithography. f) Chemical etching. g) Chemical vapor deposition of the insulation layer. h) Patterning of the insulating layer and opening with RIE. i) Device separation. More details about the fabrication process are to be found in the text.

The devices obtained following the fabrication process detailed above are shown in Figure 8.7.

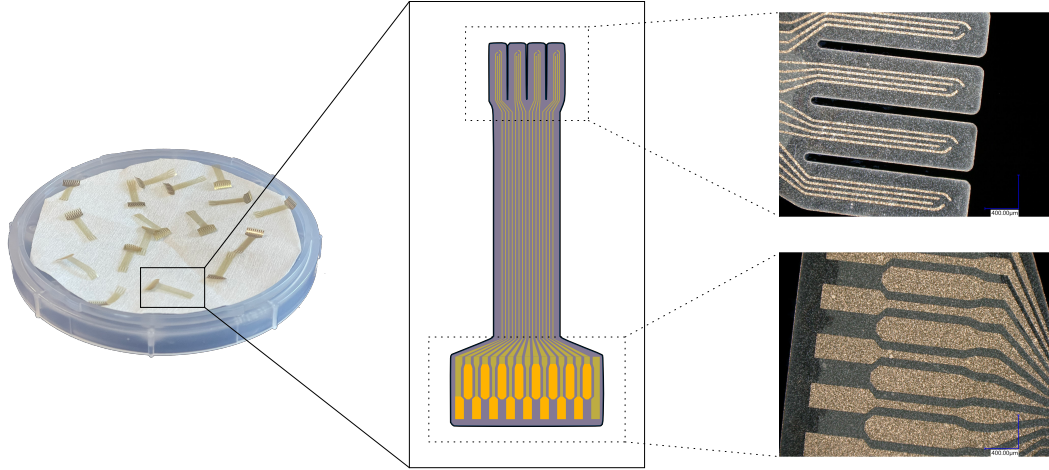


Figure 8.3: Polyimide-based devices for *in vitro* electrophysiology.

After fabrication, the devices were checked for dendritic growth and operation under mechanical stress.

Dendritic growth and characterization under mechanical stress

Using the exact same setup of the one described for dendrites on a 2D substrate presented in MATERIALS AND METHODS, dendritic growth was performed on flexible devices, as illustrated in Figure 8.4.

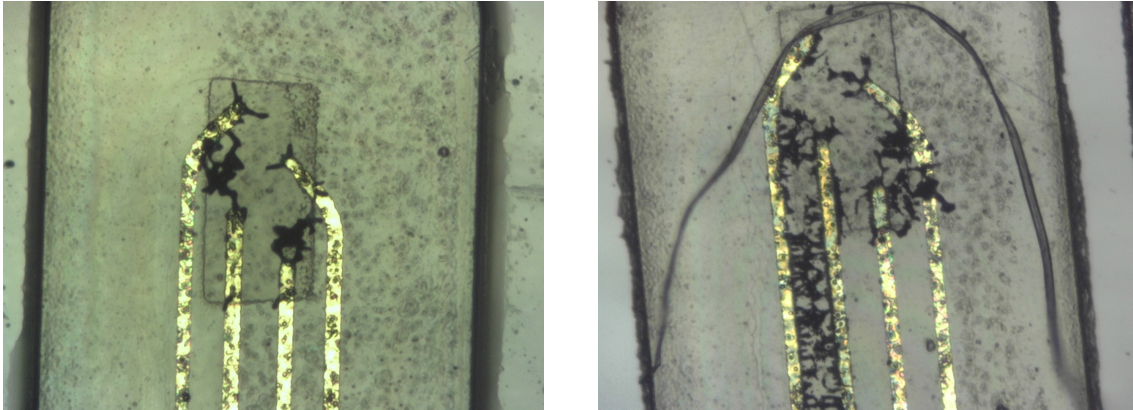


Figure 8.4: Dendritic growth on flexible devices.

In order to get as close as possible to the conditions of an *in vivo* experiment, the device was then placed under mechanical constraints which aimed at mimicking the curvature of a rodent brain, as presented in Figure 8.5. To do so with the equipment of the lab, an Eppendorf was cut in half length-wise to be used as a semi-cylinder. The flexible device was then sandwiched between the two half cylinders to force conformation. A cleanroom wipe was soaked in PBS and placed between the substrate and the top cylinder to retain the electrolyte.

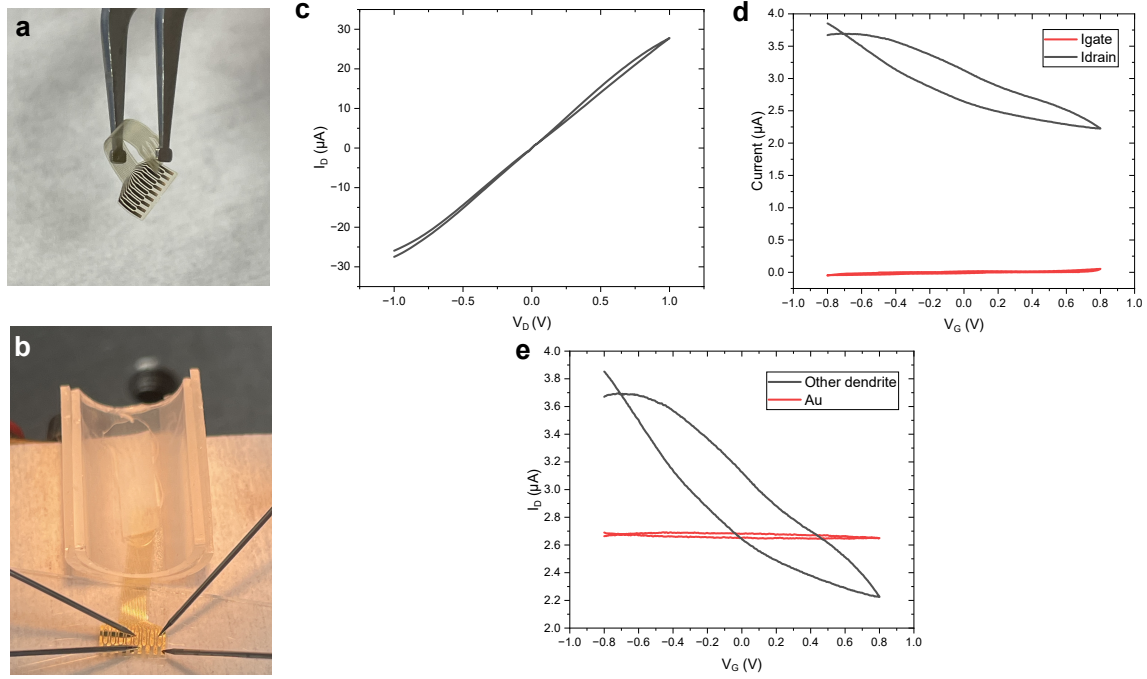


Figure 8.5: Dendritic transistor operation under mechanical constraints. **a)** Thanks to the mechanical properties of PI, the device can be bent without causing any damage. **b)** Photo of the experimental setup used to test the operation under mechanical constraint. An Eppendorf was cut in half length-wise and used to mimic the curvature of a rodent brain. A small piece of cleanroom wipe was soaked in PBS and introduced between the device and the half cylinder. **c)** Output characteristic of a dendritic transistor under mechanical constraint. **d)** Transfer characteristic of a dendritic transistor under mechanical constraint using another dendrite as the gate. $V_{DS} = 100$ mV. **e)** Transfer characteristics of the same dendritic transistor, using either a gold electrode or another dendrite as the gate. $V_{DS} = 100$ mV.

As demonstrated by the output characteristic shown in Figure 8.5c, the dendrite was not destroyed under the physical constraints, as a current was still flowing through the dendritic OECT. The transistor effect was also preserved, as illustrated in Figure 8.5d where the modulation of the drain current without any preponderant faradaic effect (the gate current remains close to none in this voltage window) can be appreciated. Note that in this experiment, another dendrite was used as the gate. Furthermore, the dendrites used as the channel and the gate were located on two different fingers of the device, so that despite the increased distance (approximately $750 \mu m$) separating them, the electrolyte still served as a communication medium, as it was detailed in chapter 6.

To check that the modulation was indeed due to electrochemical doping/dedoping and no other effect, a small gold electrode of the device was tested as the gate of the system (Figure 8.5e). Not being a good choice of material for such a small surface as the electrode used in the present situation²⁴², no modulation of the drain current was observed, as expected. The dendritic transistor therefore appears to operate perfectly under mechanical constraints, offering good hope for *in vivo* experiments with this kind of device in the future. Although *in vivo* conditions are not expected to be more mechanically constraining, a biodegradable scaffold such as polylactic-glycolic acid could be used to protect the dendrites during the insertion

of the devices²⁹³.

Improving the devices

If the previous results turned out to be encouraging, some improvements could be made on the flexible devices. In particular, the adhesion of the insulating layer could be corrected. As highlighted in Figure 8.6, the parylene layer does not seem to be well attached to the devices. Indeed, dendritic growth outside of the openings indicates that growth electrolyte infiltrated under the Pa-C layer, and folding over can clearly be appreciated in the provided pictures.

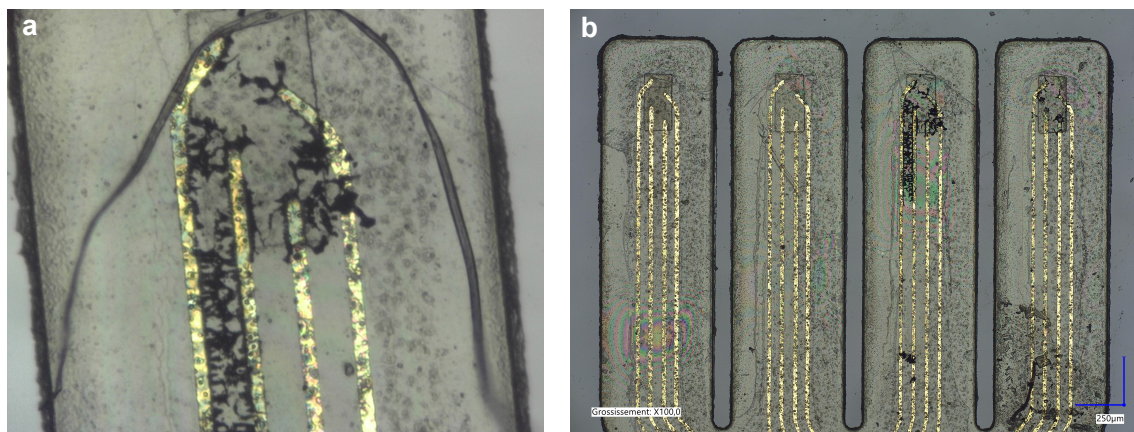


Figure 8.6: Adhesion issue of the insulating layer. a) Close-up photo showing that dendritic growth underneath the parylene layer. b) Zoom-out photo of the same device, highlighting the detachment and folding over of parylene, in particular on the third finger from the left. Iridescence is also visible on the first and third finger, indicating inhomogeneities of the material.

Several solutions can be envisioned to solve this issue. First, the adhesion between polyimide and parylene could be enhanced, using for example surface treatment. Oxygen plasma before Pa-C deposition could help make polyimide hydrophilic and enhance adhesion between the layers. Second, another material could be employed for the insulating layer. SU-8 is for example a negative photoresist frequently used as an encapsulation layer for applications requiring biocompatibility^{294,295}. Using such a resist would also make the fabrication process a little shorter, as the patterning of the passivation layer would only necessitate a classic UV-photolithographic process instead of physical etching. A second PI layer could also be an alternative.

II Miniaturization and 3D network

Making polymer dendrite networks appealing outside of the academic world calls for integration efforts. To do so, artificial dendrites would need to be miniaturized. Not only would it allow footprint to be reduced, but it should also turn out to be beneficial for neural sensing, as it was discussed in the previous chapter.

I therefore fabricated a new substrate with gaps of the order of a few hundreds of nm between two gold electrodes to try nanometric growth (see Figure 8.7).

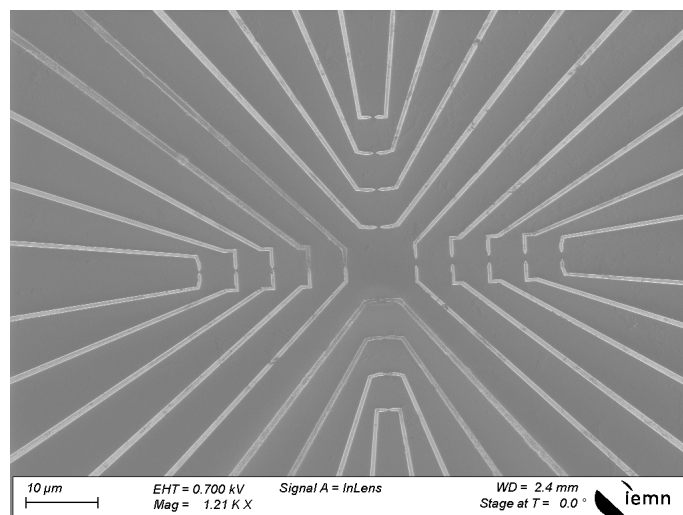


Figure 8.7: SEM image of a substrate for nanometric growth of polymer dendrites.

After a few tests, it appears that dendrites can indeed be grown into such a small gap as the ones shown in the previous image. Figure 8.8 presents two of these electrogenerated objects, observed with a scanning electron microscope.

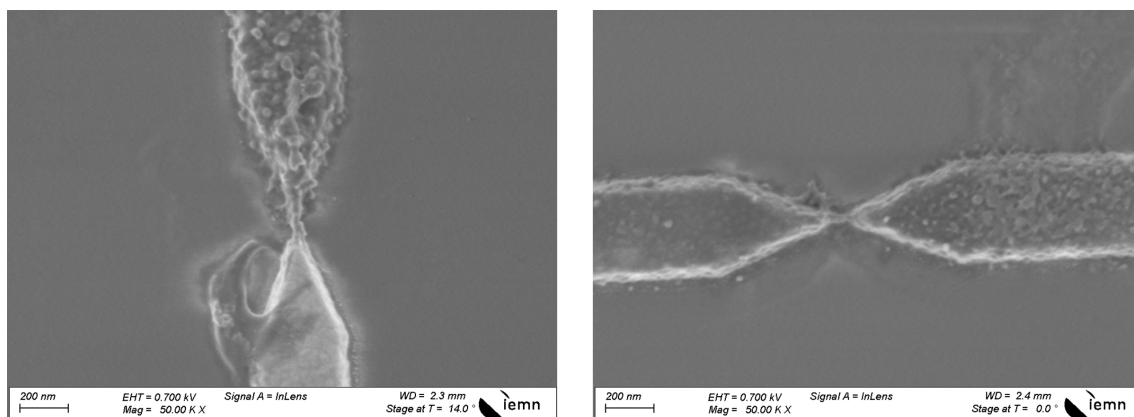


Figure 8.8: SEM images of nanodendrites. The gap between the electrode was filled by a dendrite.

Nanodendrites could also be grown in 3D, as it was for instance proposed by Hagiwara *et al.*²²² to allow structural plasticity on many dimensions. At IEMN, the work led by Luc Brulin during his internship with the help of Dr Paul Moustiez demonstrated the possibility for dendrites to be grown between nanocones, as illustrated in Figure 8.9.

Nevertheless, growing nanometric dendrites presents a few technological challenges. Contrary to micrometric dendrites, which growth can be monitored using a simple microscope, it is not possible to visually check for growth termination with such small objects. In addition to that, due to their small dimensions these objects do not let much current through, which makes it hard to detect in noisy environments such as an electrolytic solution. However, as it was highlighted in chapter 6, not only is the steady-state output current of dendrites interesting, but also their capacitive response. Therefore, if their steady-state response is hardly discernible

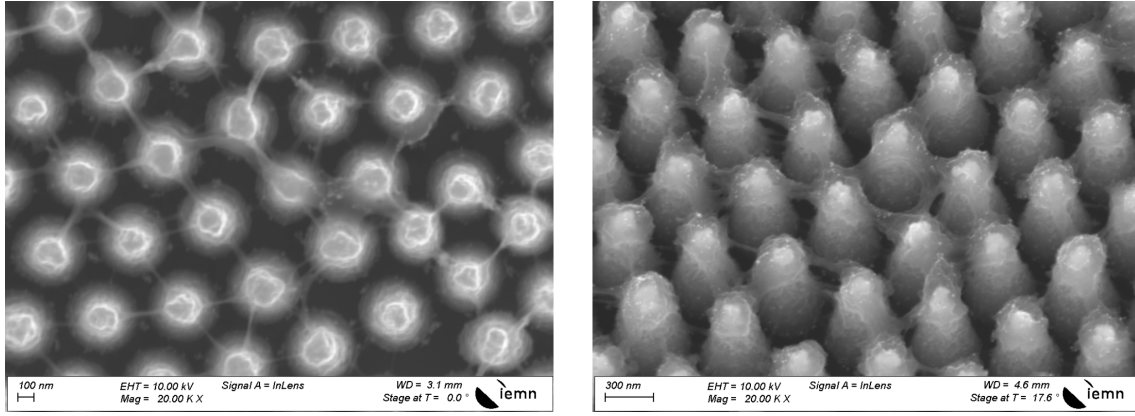


Figure 8.9: Dendritic networks grown between nanoneedles. These SEM images let us appreciate the spreading of the dendritic network between the nanocones. In particular, the tilted image shows that the dendrites are suspended between the tips of the cones.

from background noise, maybe the high capacitance of these objects could turn out to be more helpful than their conductance.

Another challenge might be the mechanical frailty of dendrites. One of the means to circumvent that issue could be to grow dendrites into solid matrices such as hydrogels or porous templates to protect them.

III Final discussion

Biological neural networks are an important source of inspiration when it comes to efficient computing, mainly due to their innate ability to massively parallelize information processing. If large artificial neural network models have proven to be very efficient computationally, they are still far from the energy efficiency of biological brains. In these models, performance comes with high accuracy in the synaptic weights and dense interconnections between layers of neurons that translate into important energy consumption. In biological networks, only useful connections are grown and result in sparse and specialized topologies that optimize resources utilization. Additionally, these topologies possibly bear more profound memory representation that connectionist models based solely on synaptic weight evolution during learning cannot capture. Topologies set the dimensional space in which synaptic plasticity can be expressed, and could possibly be the foundation for semantic representations (i.e., higher level of organization in the representation of concepts)²⁹⁶. The work presented in this manuscript highlights the role of topologies in the spatiotemporal projection of information and could be a framework to reconcile connectionist and semantic approaches. The control operated over the topology of dendritic networks governs the complexity of the operation of the system, due to the interplay between form and function. In other words, if small networks implement nearly linear functions, such as the MAC operation discussed in a previous chapter, more developed systems exhibit nonlinear behaviors that are more complex to implement but offer greater computational power. The bottom-up approach of dendritic growth therefore allows the function of a multiterminal device

to be defined by its physical structure, as complexity emerges from the physical structure of the network itself.

Another concept seldom discussed in the neuromorphic community is the wide diversity of synapses and information carriers in the brain. Although a great number of neuromodulators has already been identified²⁹⁷, our understanding of the underlying mechanisms by which they participate in information processing remains unclear. Especially striking is the varied nature of these neuromodulators, from peptides to gaseous molecules. Apart from neurotransmitters, information also appears to be conveyed by marker molecules, as it has been proposed that their gradient of concentration could guide axonal growth in the early stage of brain development⁷⁸. This sharply contrasts with the current view in electronics, where electrons are the only information carrier. Using an electrochemical platform such as the one presented in this article, the number of information carriers in the system could be increased drastically, as many biomolecules have already been demonstrated to modulate the behavior of organic electrochemical transistors^{192,208,217,298,299}. With recent advances regarding degradability and disintegration of organic polymers^{300–302}, full structural plasticity could also become a way to store information directly in the structure of the network, making the most out of the limited resources of the system by destroying unnecessary synapses to reemploy the monomer molecules more efficiently. For instance, using a laser to destroy a dendrite, we showed that new polymer branches could be formed to recreate the connection, as illustrated in Figure 8.10.

Although this strategy might not be the best way to implement LTD (the substrate was damaged by the laser, and its integration might cause some trouble), it nevertheless illustrates that polymer dendrites are good candidates for the implementation of a full form of structural plasticity. The ability to learn and evolve is indeed an incredibly potent property of the brain, one that the current top-down approach of solid-state electronics cannot easily emulate. In addition, it illustrates the possibility for electrogenerated objects to be repaired in case of damage or destruction. Evolving systems have the great advantage to adapt to the tasks at hand, so that resources are employed as efficiently as possible. In such systems, complex behavior arises from simple local rules, as it was proposed with cellular automata³⁰³. Using the computational power of adaptive bottom-up physical systems has been explored with NanoCells^{19,20,304} or slime mold^{17,305–308}, but the physical structures in these examples were obtained without any physical relationship with the computation to be performed. Here, the versatility of electropolymerization¹⁵¹ offers to adjust the operation of the device as needed *in operando*, enhancing its capabilities and bringing organic electronics one step closer to biology. Implementing this technology in a solid-state electrolyte could make solid-state electronics more plastic and create new possibilities with the integration of a bottom-up additive approach to otherwise deterministic electronics. Indeed, the stochastic growth and the variability of organic electronics, which are inconveniences in the realm of traditional electronics and digital computing, turn into opportunities when it comes to unconventional computing. Having systems which growth is not fully controllable, and which behavior cannot be anticipated is highly valued for cryptographic and security applications such as Physical Unclonable Functions (PUFs), which take advantage of the variability in production to generate devices that present a unique signature²⁶³.

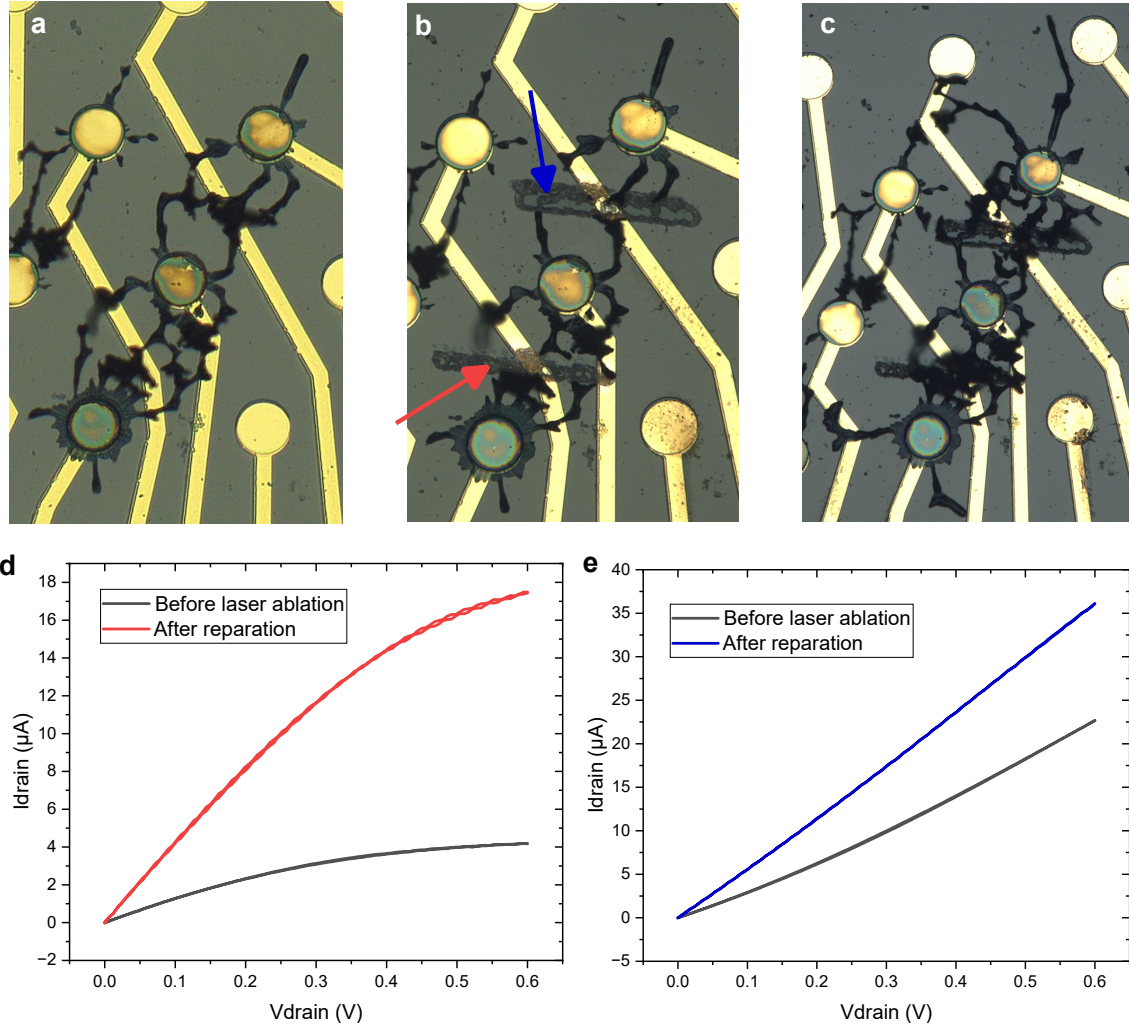


Figure 8.10: Structural plasticity emulated via the destruction of a dendritic interconnect with a laser. **a)** A dendrite was grown on a MEA using the usual setup. **b)** The dendrite was then destroyed with the help of a laser. The blue and red arrows indicate the location of the laser cut. **c)** New interconnects were grown between the two electrodes. **d)** $I(V)$ curve of the dendrite indicated by the red arrow, before laser ablation and after reparation. **e)** $I(V)$ curve of the dendrite indicated by the blue arrow, before laser ablation and after reparation.

Although still a recent and untapped concept, conductive polymer dendrites show great potential for the physical implementation of unconventional computing and the integration with solid-state electronics.

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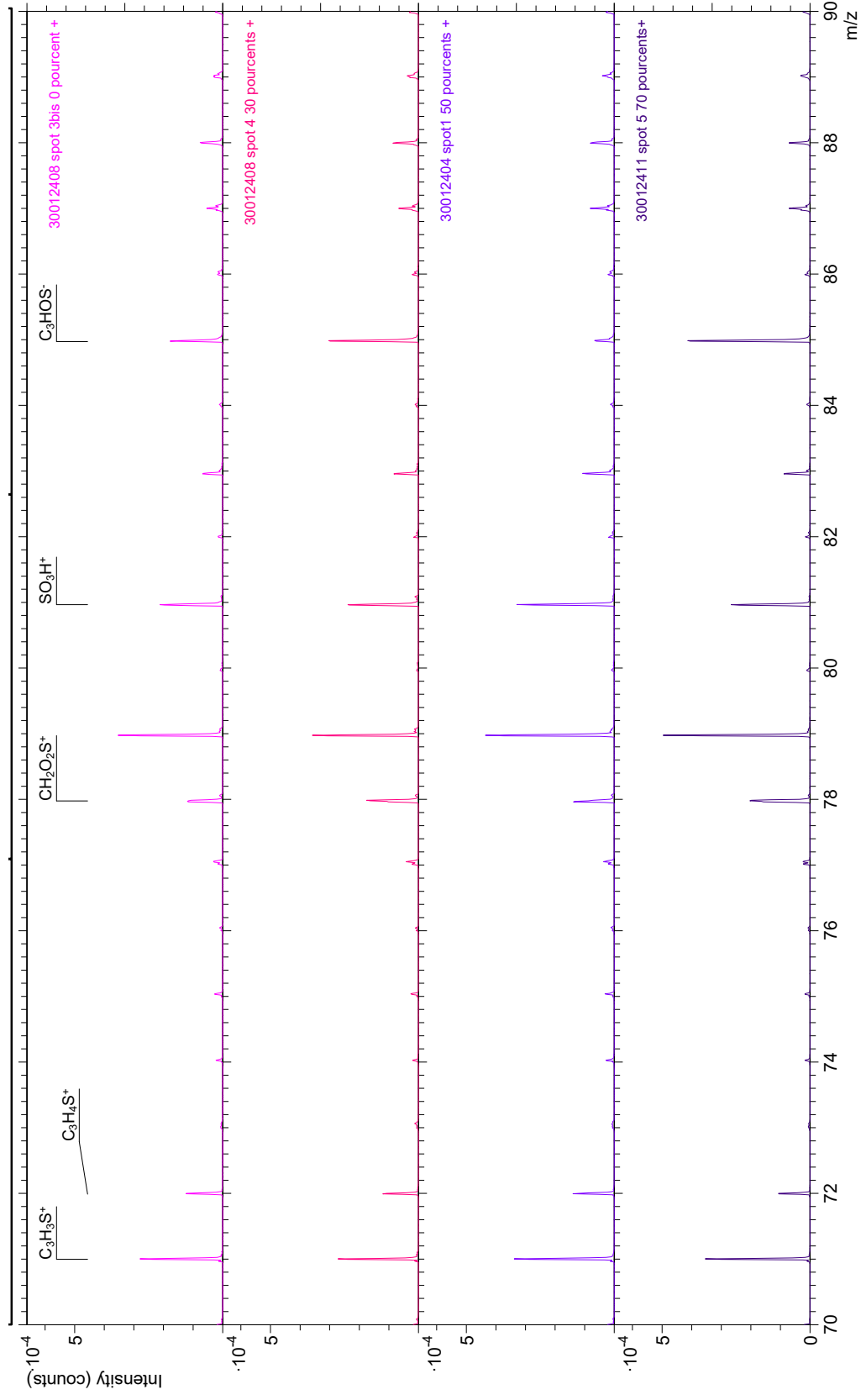
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Part II

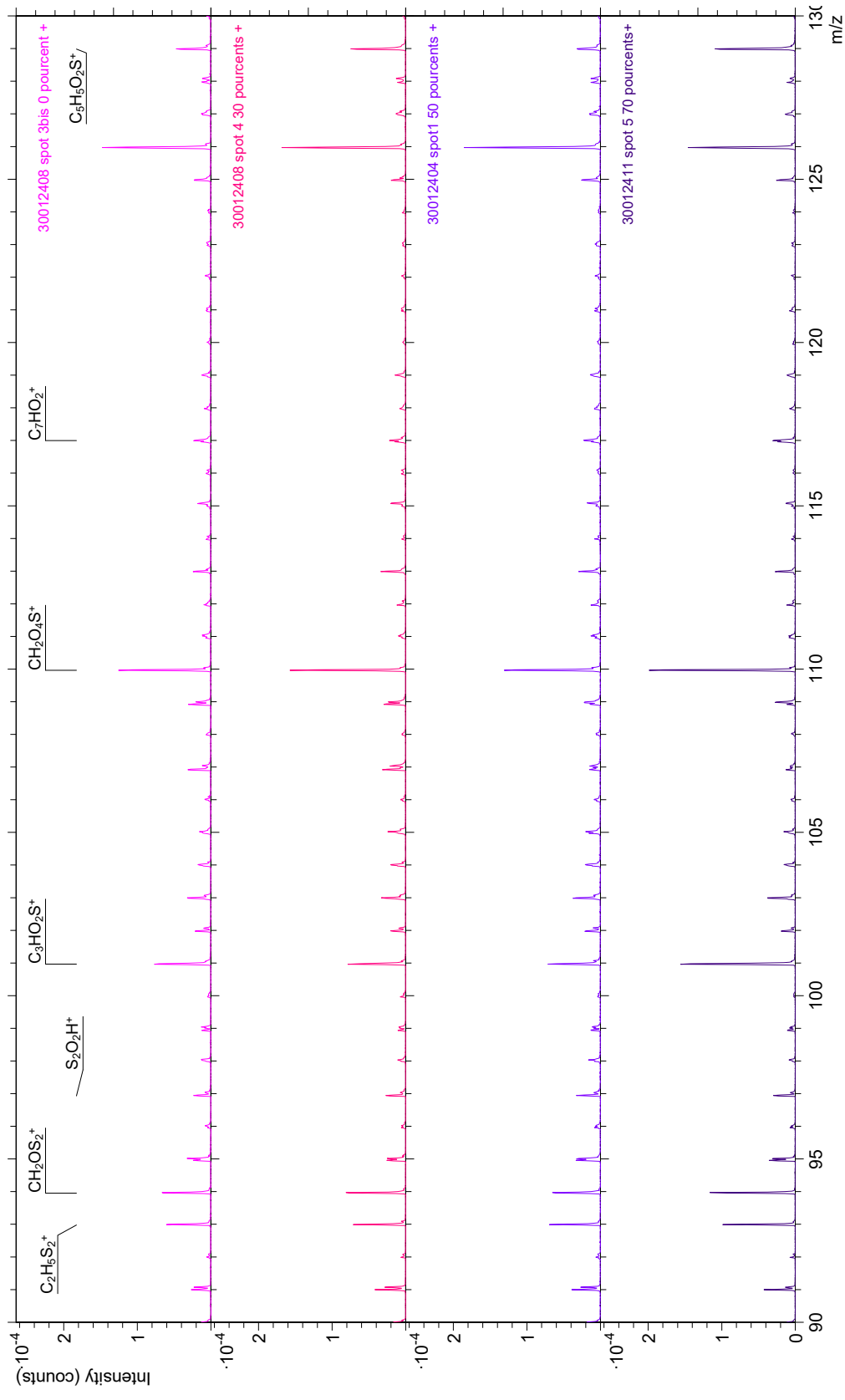
Appendix

ToF-SIMS analysis

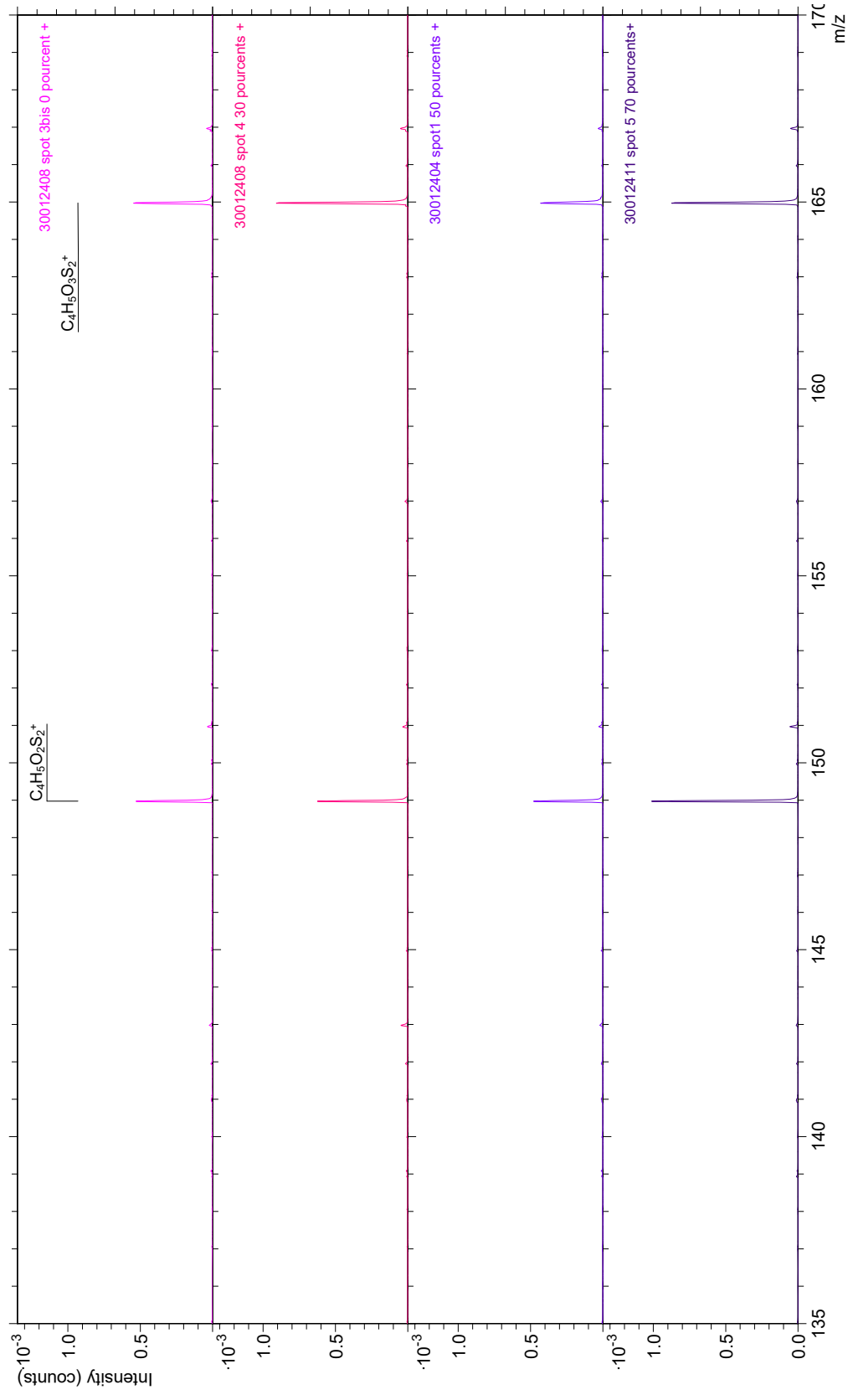
Spectres overlay pola + range 70-90



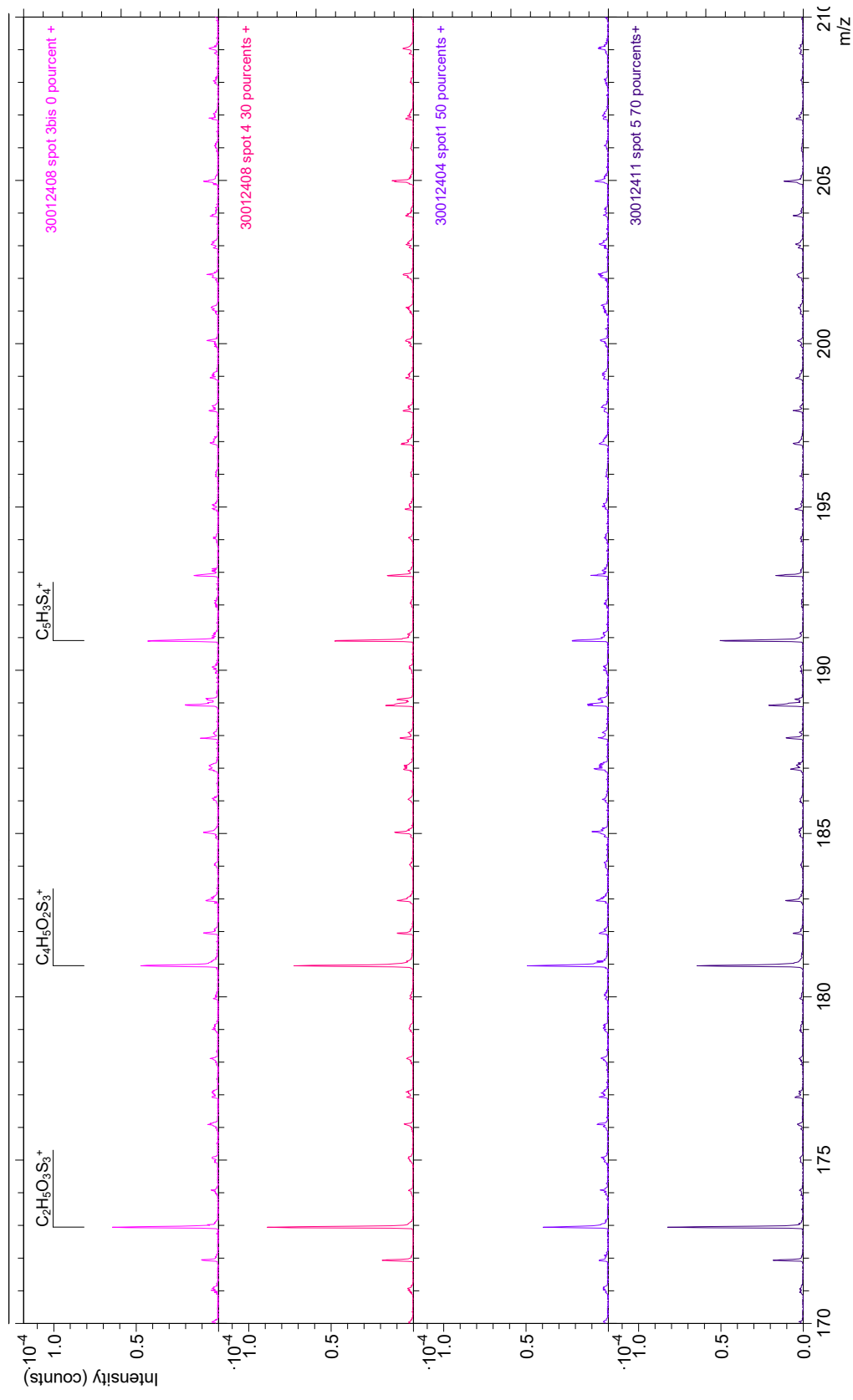
Spectres overlay pola + range 90-130



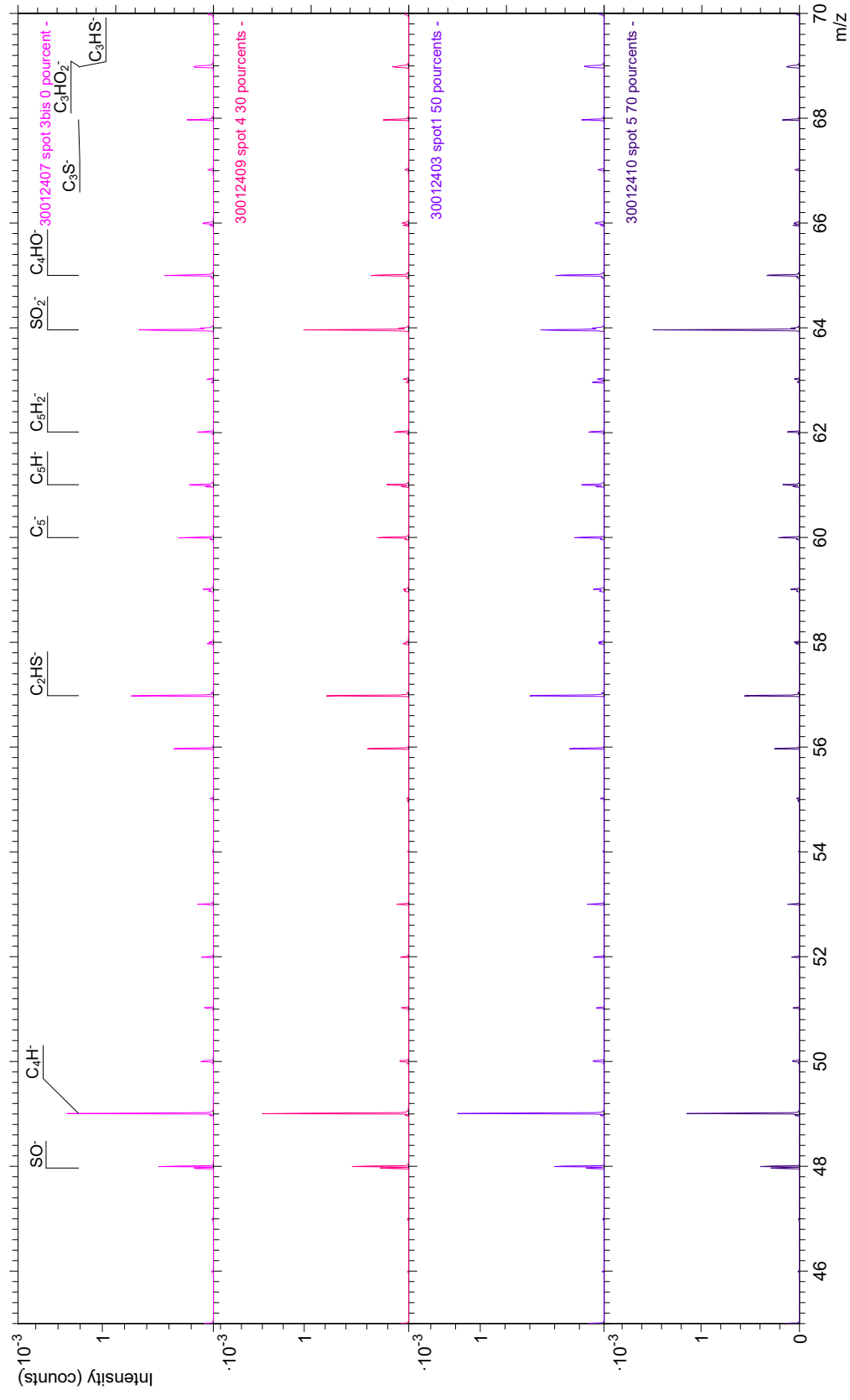
Spectres overlay pola + range 135-170



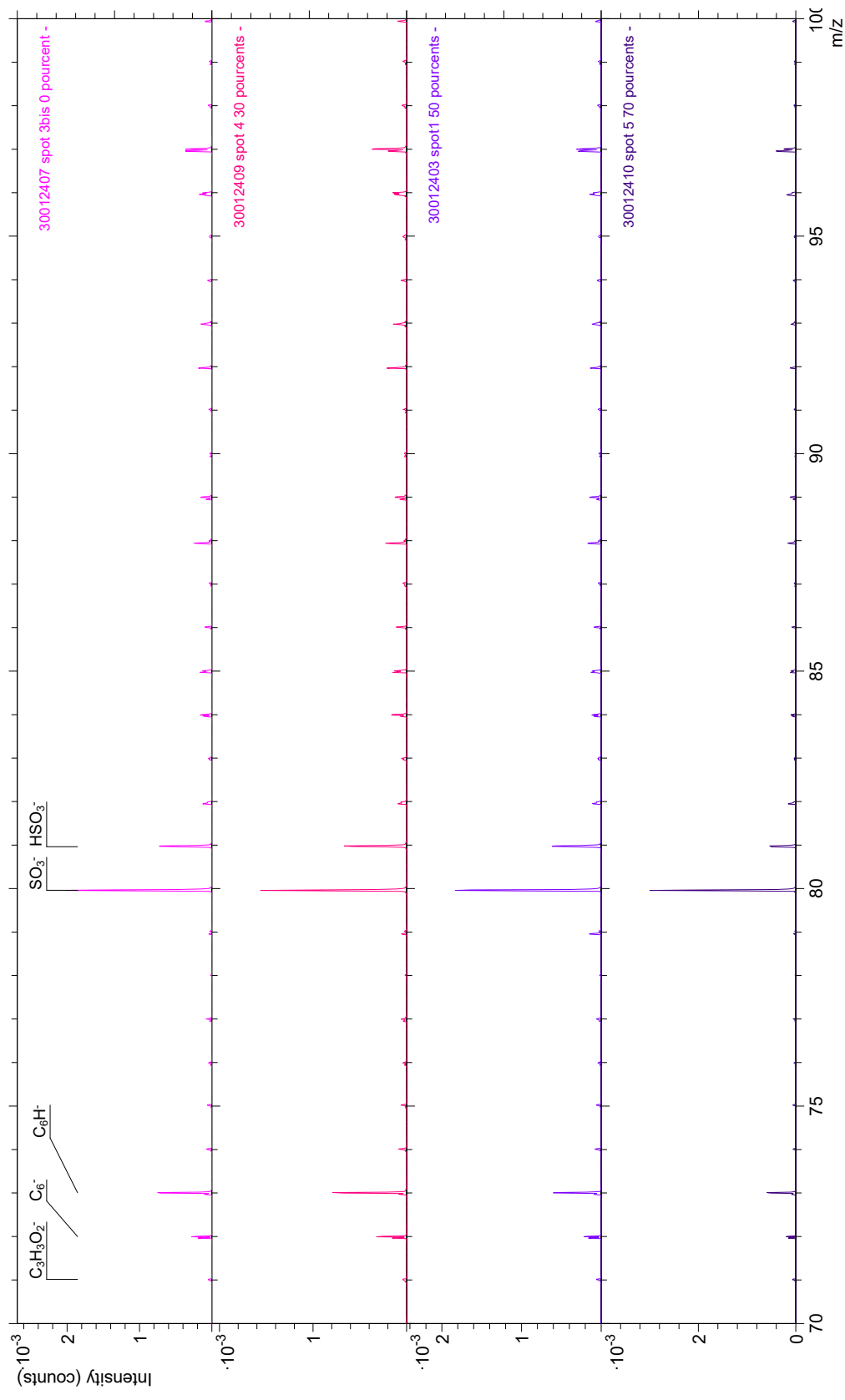
Spectres overlay pola + range 170-210



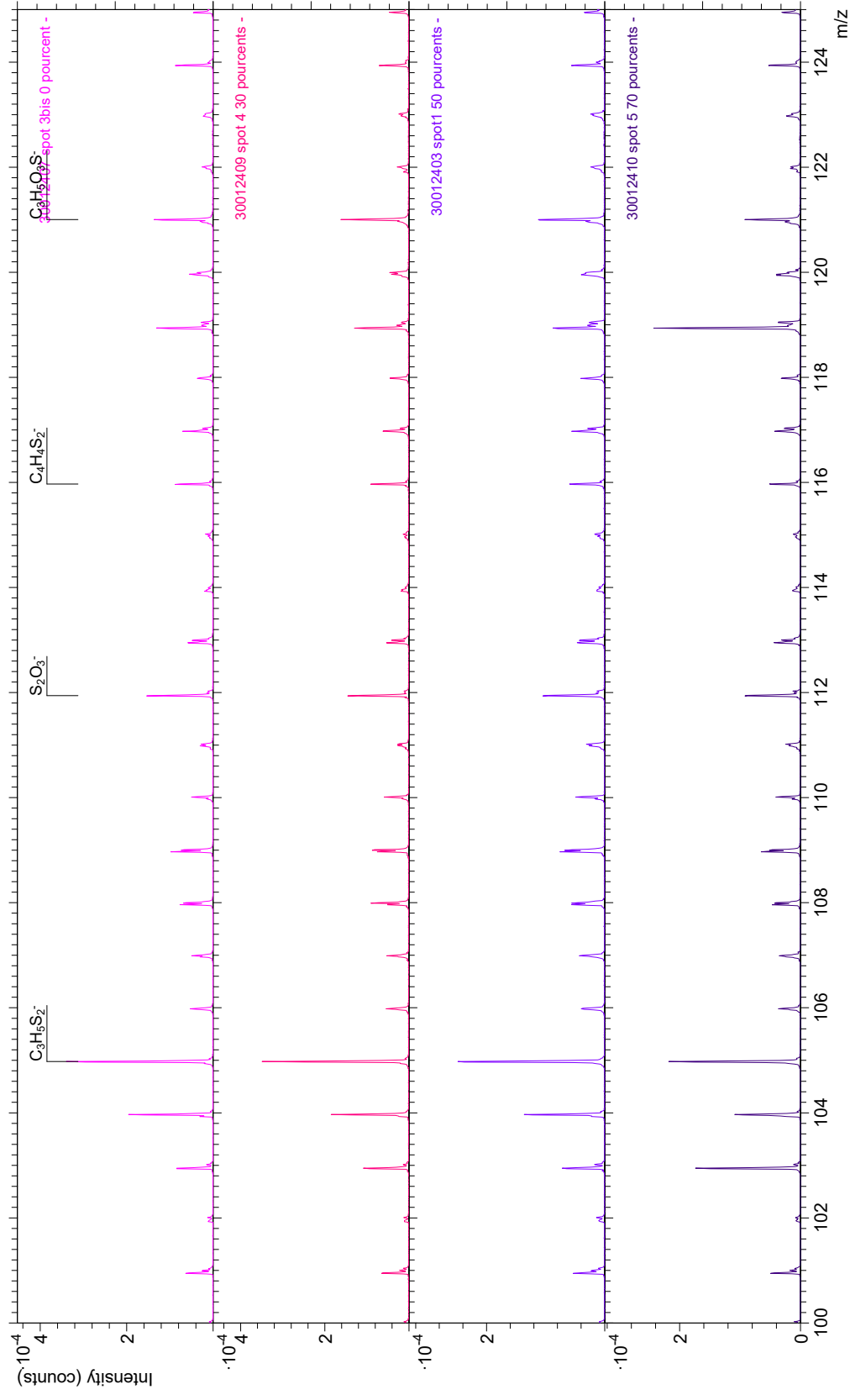
Spectres overlay pola - range 45-70



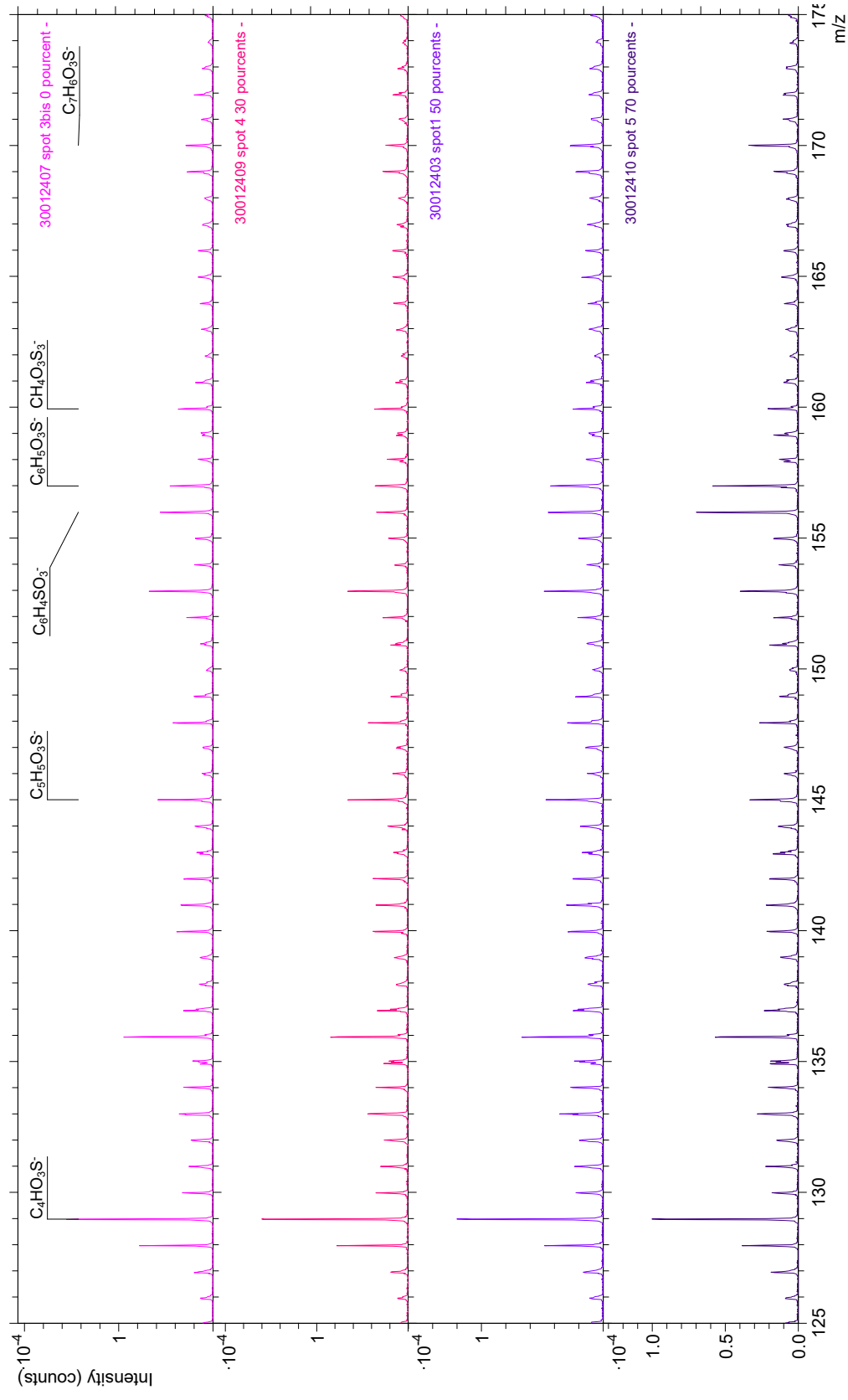
Spectres overlay pola - range 70-100



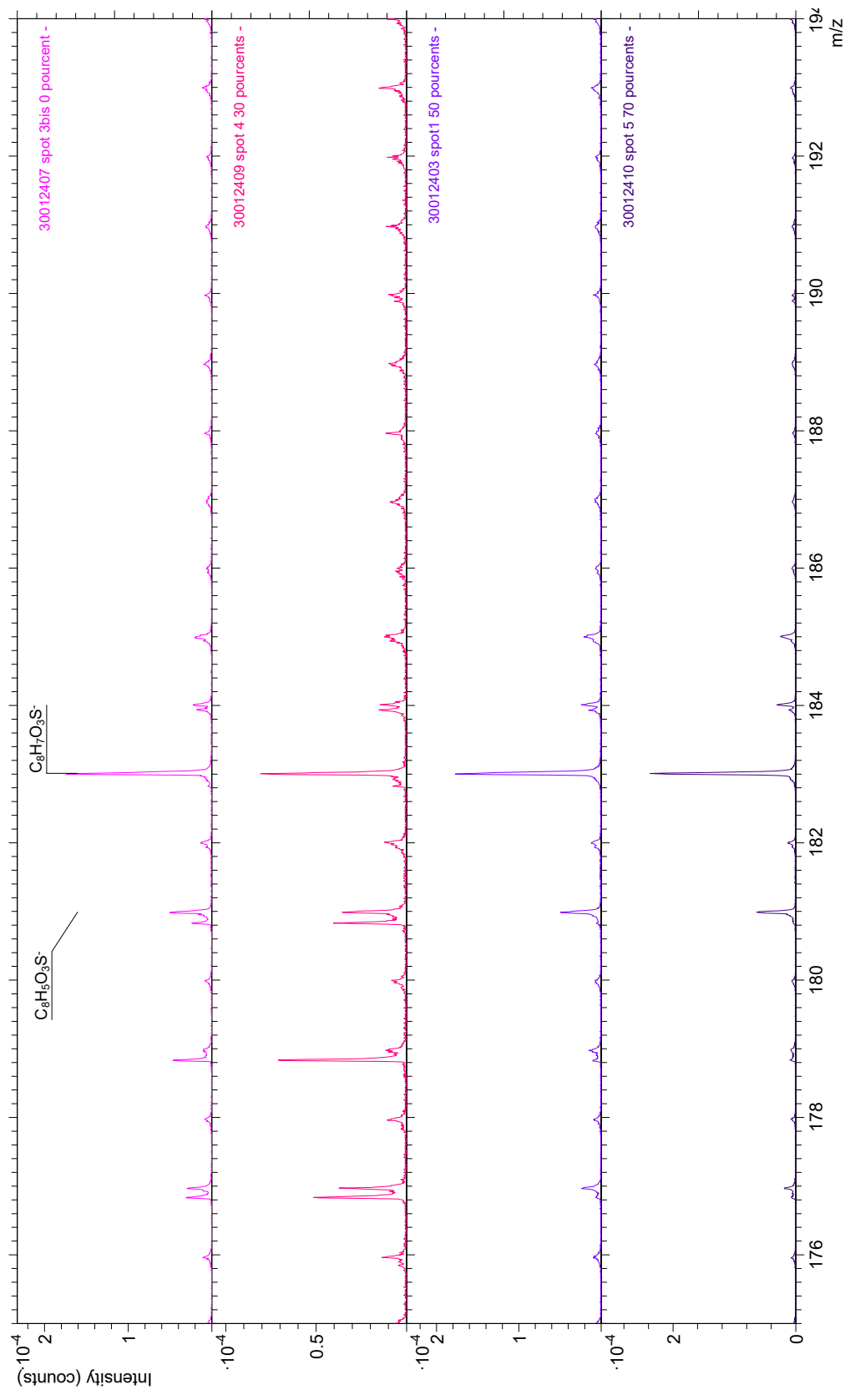
Spectres overlay pola - range 100-125



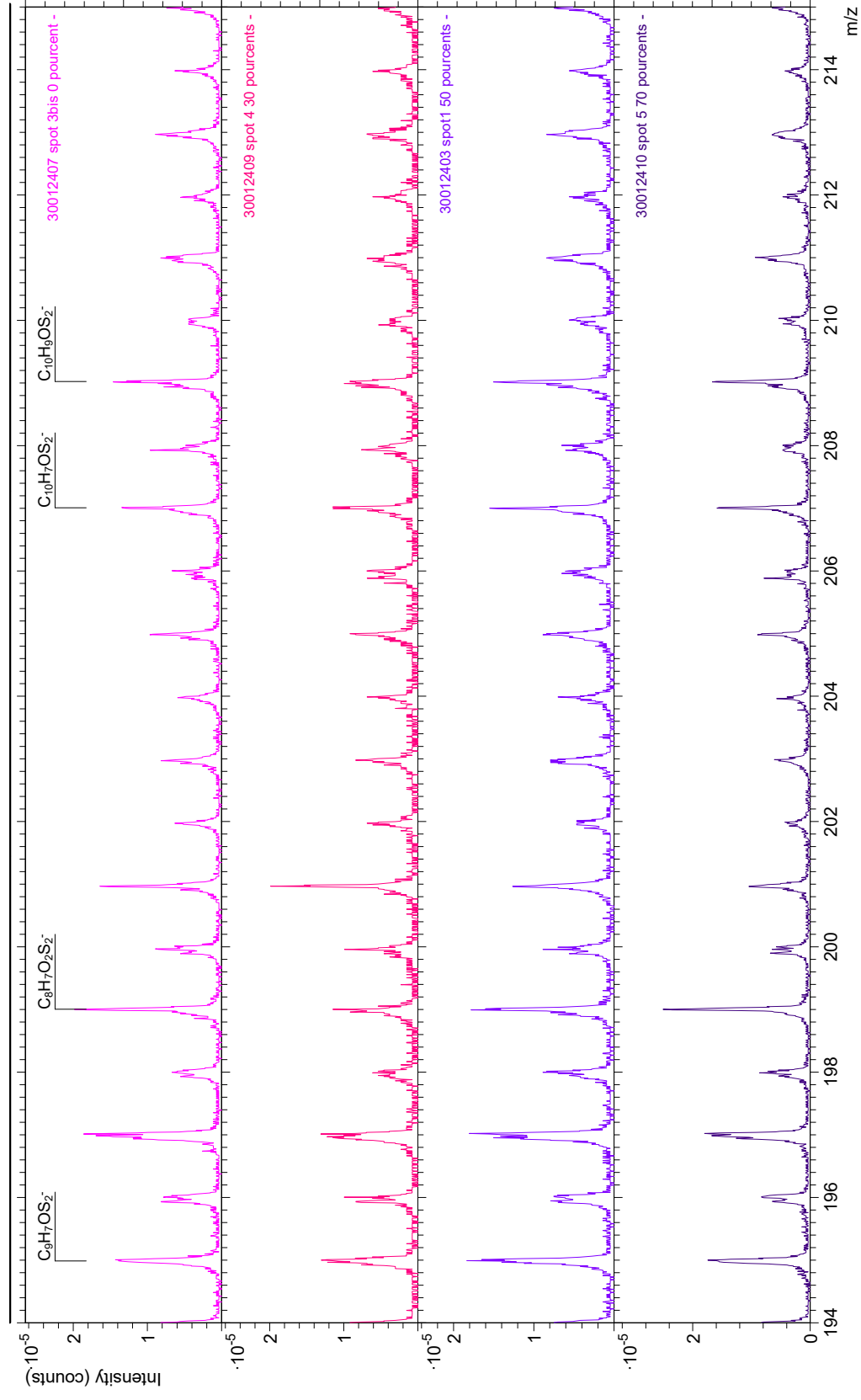
Spectres overlay pola - range 125-175



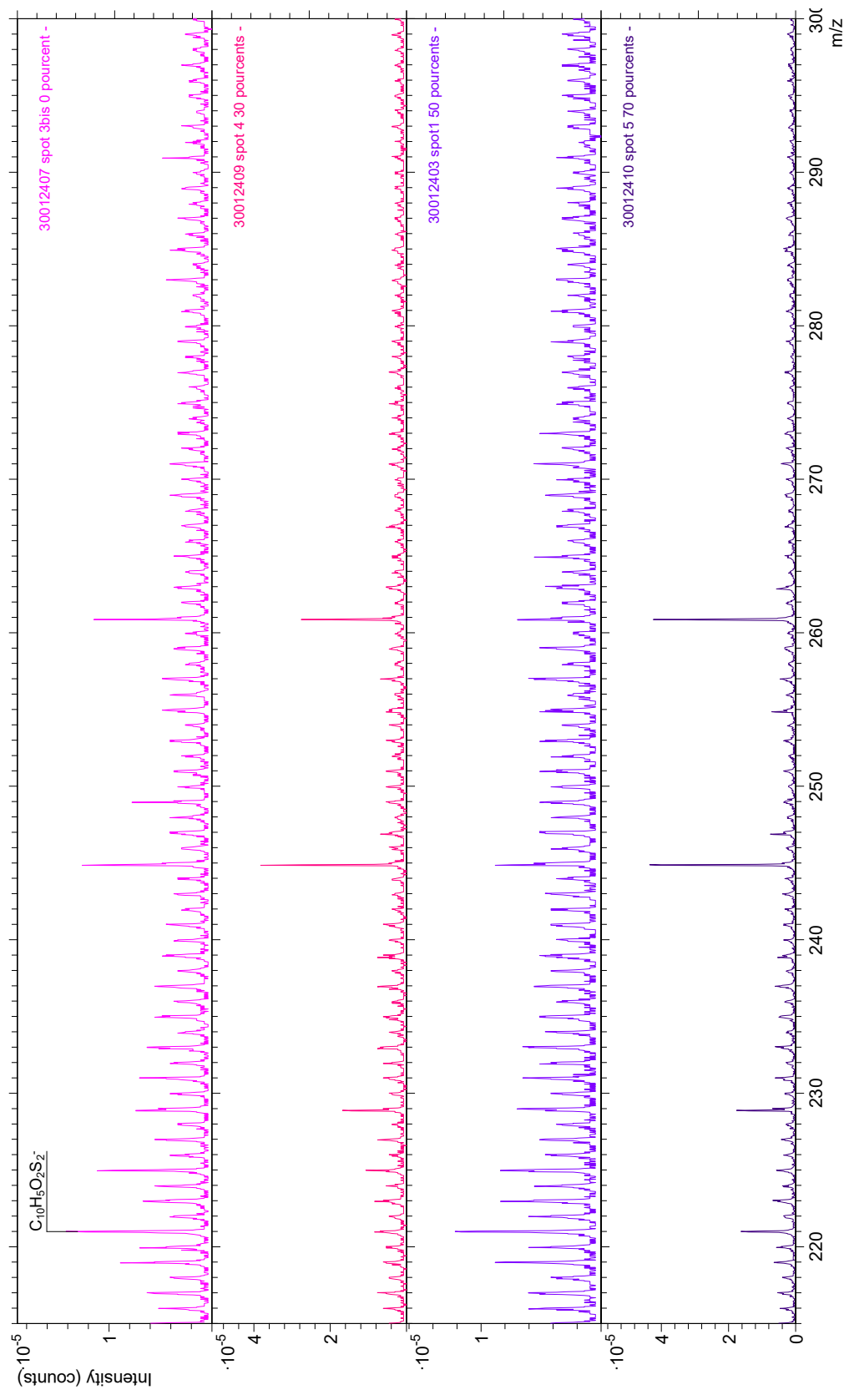
Spectres overlay pola - range 175-194



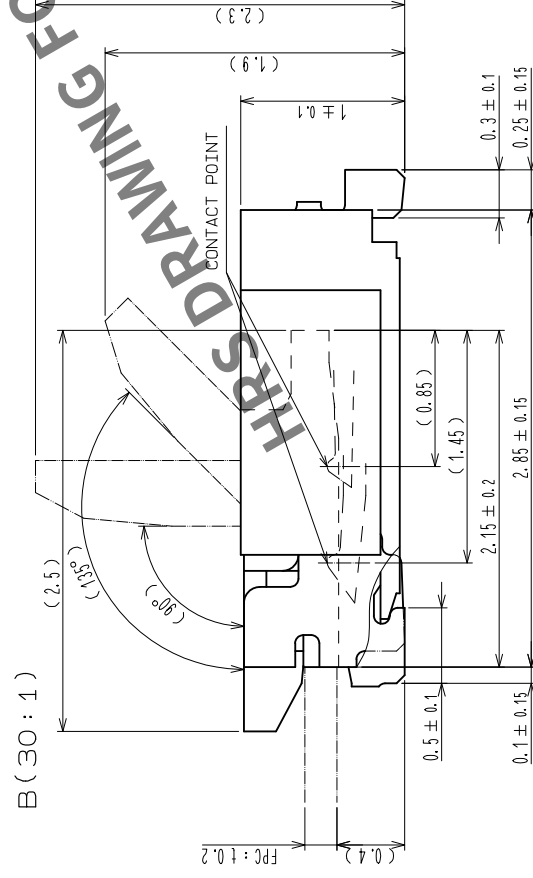
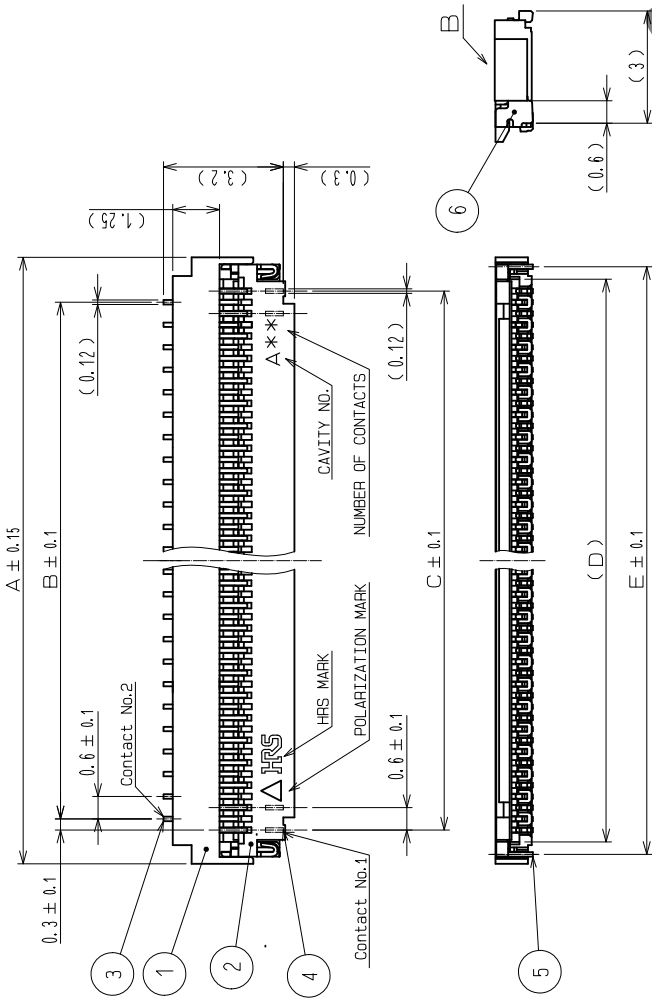
Spectres overlay pola - range 194-215



Spectres overlay pola - range 215-300



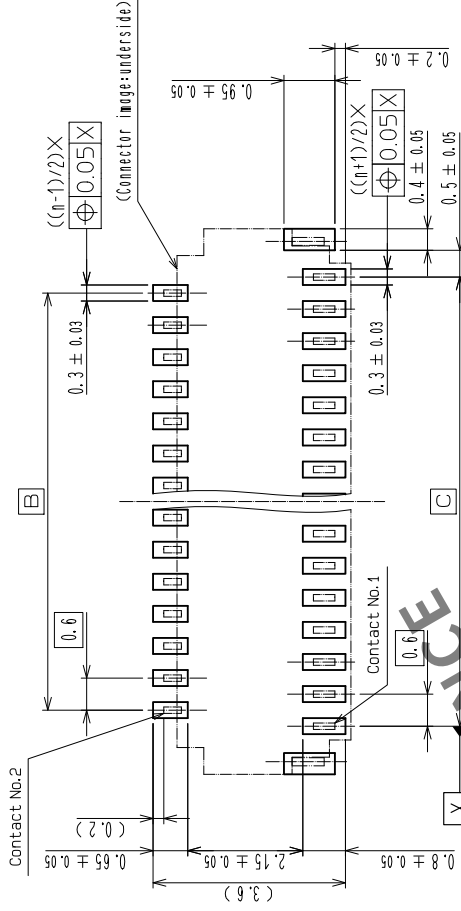
**Technical documentation for the
FH26W-17S-0.3SHW60 HiRose
ZIF connector**



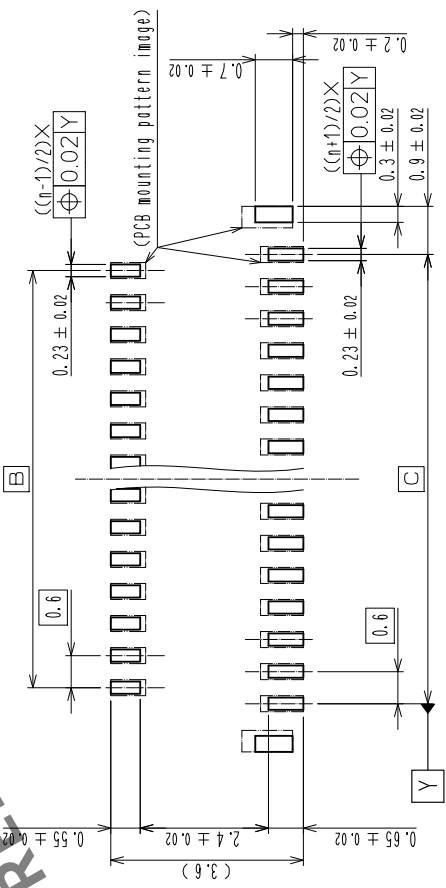
NOTE

1. THE DIMENSIONS IN PARENTHESES ARE FOR REFERENCE.
2. LEAD CO-PLANARITY INCLUDING REINFORCED METAL FITTINGS SHALL BE 0.1 MAX.
3. TO BE DELIVERED WITH TAPE AND REEL PACKAGES.
SEE ATTACHED PACKAGING SPECIFICATIONS FOR DETAILS.
4. NOTE THAT PREVENTIVE HOLE FOR SINK MARK COULD BE ADDED FOR IMPROVEMENT.
THE QUALITY REMAINS GOOD, EVEN WITH THE DARK SPOTS, WHICH COULD OCCASIONALLY OCCUR ON MOLDED PLASTIC.
5. THIS PRODUCT SATISFIES HALOGEN FREE REQUIREMENTS DEFINED AS 900ppm MAXIMUM CHLORINE, 900ppm MAXIMUM BROMINE, AND 1500ppm MAXIMUM TOTAL OF CHLORINE AND BROMINE.
6. 'n' REPRESENTS THE NUMBER OF CONTACTS.

RECOMMENDED PCB MOUNTING PATTERN (10:1)

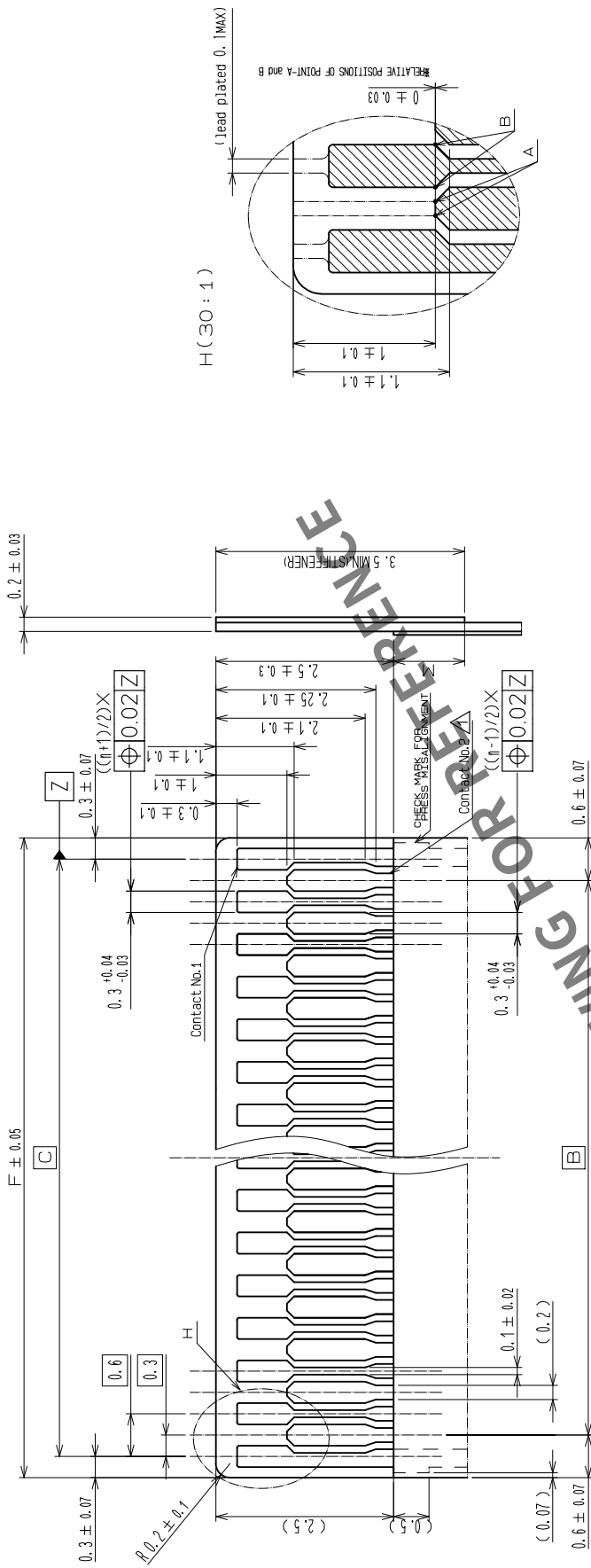


RECOMMENDED STENCIL PATTERN (10:1)

[illegible]

RECOMMENDED FPC(FREE)

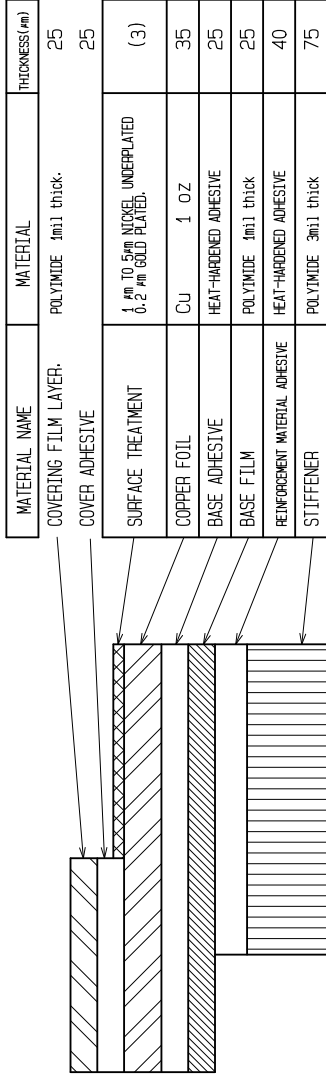
(SCALE:FREE)



DIMENSION MUST BE 0.5mm MINIMUM WHEN THE STIFFENER IS SHORTER THAN 3.5mm.

FPC CONFIGURATION (REFERENCE EXAMPLE)

(SCALE:FREE)



MATERIAL NAME	MATERIAL	THICKNESS(mm)
COVERING FILM LAYER.	POLYIMIDE 1mil thick.	25
COVER ADHESIVE		25
SURFACE TREATMENT	1 μm TO 5μm NICKEL UNDERPLATED 0.2 μm GOLD PLATED.	(3)
COPPER FOIL	CU 1 OZ	35
BASE ADHESIVE	HEAT-HARDENED ADHESIVE	25
BASE FILM	POLYIMIDE 1mil thick	25
REINFORCEMENT MATERIAL ADHESIVE	HEAT-HARDENED ADHESIVE	40
STIFFENER	POLYIMIDE 3mil thick	75



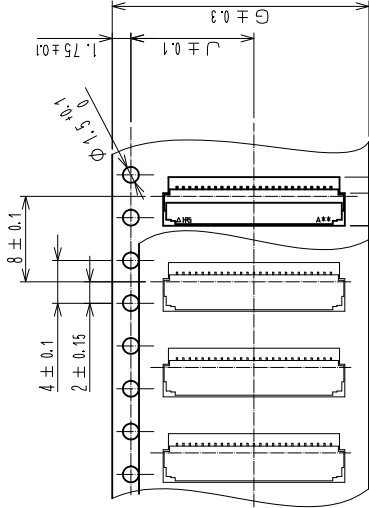
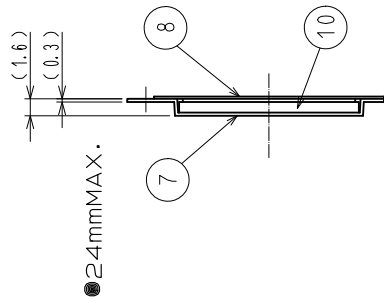
DRAWING NO. EDC3-323714-05

PART NO. FH26W-**-S-O. 3SHW<60>

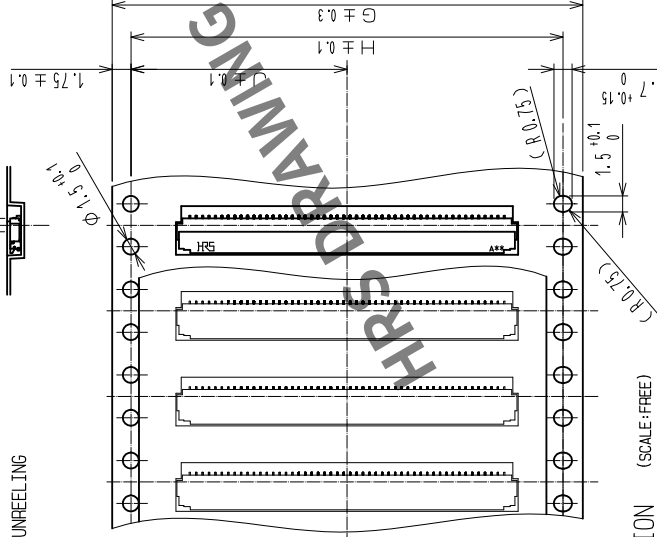
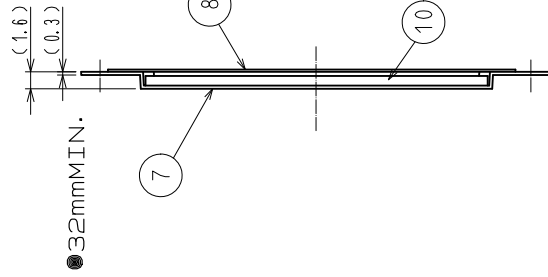
CODE NO. CL580

2/6

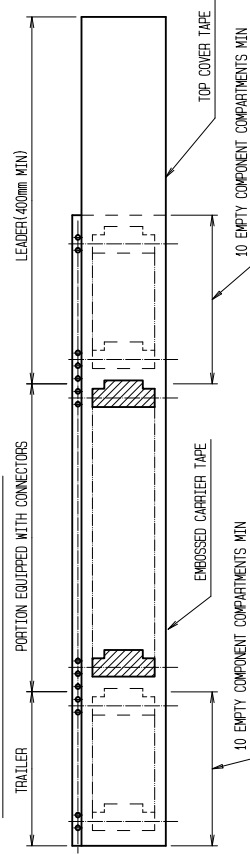
EMBOSSED CARRIER TAPE DIMENSION (2:1)



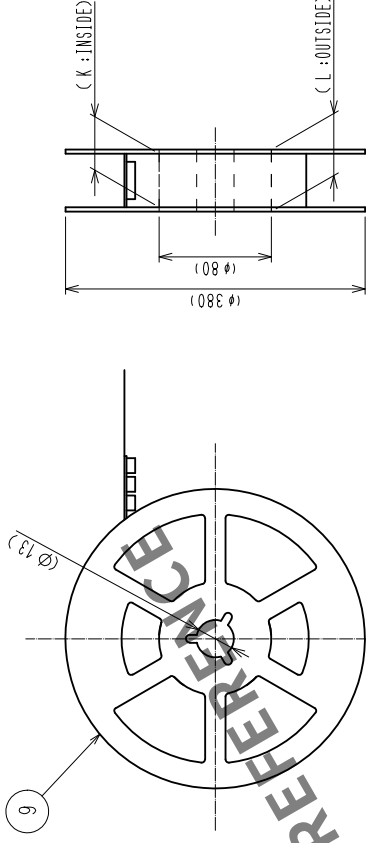
DIRECTION OF UNREELING



LEADER, TRAILER DIMENSION (SCALE:FREE)



REEL DIMENSION (FREE)



NOTE 7 PER REEL : 5000 CONNECTORS.

△ 8 REFER TO JTS C 0806 and IEC 60286-3 (PACKAGING OF COMPONENTS FOR AUTOMATIC HANDLING.)

HRS	DRAWING NO.	EDC3-323714-05
	PART NO.	FH26W-**-S-0.3SHW(60)
	CODE NO.	CL580
	NO.	3/6

PART NUMBER	CL	NUMBER OF CONTACT	DIMENSION OF CONNECTOR, FPC PATTERN AND METAL MASK						DIMENSION OF DRAWING FOR PACKING					
			DIMENSION OF CONNECTOR, FPC PATTERN AND METAL MASK						DIMENSION OF DRAWING FOR PACKING					
			A	B	C	D	E	F	G	H	J	K	L	
FH26W-13S-O. 3SHW(60)	CL580-2401-1-60	13	5.4	3.0	3.6	4.23	4.9	4.2	16	—	7.5	17.4	21.4	A
FH26W-15S-O. 3SHW(60)	CL580-2402-4-60	15	6.0	3.6	4.2	4.83	5.5	4.8	16	—	7.5	17.4	21.4	B
FH26W-17S-O. 3SHW(60)	CL580-2403-7-60	17	6.6	4.2	4.8	5.43	6.1	5.4	16	—	7.5	17.4	21.4	
FH26W-19S-O. 3SHW(60)	CL580-2437-9-60	19	7.2	4.8	5.4	6.03	6.7	6.0	16	—	7.5	17.4	21.4	
FH26W-21S-O. 3SHW(60)	CL580-2404-0-60	21	7.8	5.4	6.0	6.63	7.3	6.6	16	—	7.5	17.4	21.4	
FH26W-23S-O. 3SHW(60)	CL580-2405-2-60	23	8.4	6.0	6.6	7.23	7.9	7.2	16	—	7.5	17.4	21.4	
FH26W-25S-O. 3SHW(60)	CL580-2406-5-60	25	9.0	6.6	7.2	7.83	8.5	7.8	16	—	7.5	17.4	21.4	
FH26W-27S-O. 3SHW(60)	CL580-2400-9-60	27	9.6	7.2	7.8	8.43	9.1	8.4	16	—	7.5	17.4	21.4	
FH26W-29S-O. 3SHW(60)	CL580-2407-8-60	29	10.2	7.8	8.4	9.03	9.7	9.0	24	—	11.5	25.4	29.4	
FH26W-31S-O. 3SHW(60)	CL580-2408-0-60	31	10.8	8.4	9.0	9.63	10.3	9.6	24	—	11.5	25.4	29.4	C
FH26W-33S-O. 3SHW(60)	CL580-2409-3-60	33	11.4	9.0	9.6	10.23	10.9	10.2	24	—	11.5	25.4	29.4	
FH26W-35S-O. 3SHW(60)	CL580-2410-2-60	35	12.0	9.6	10.2	10.83	11.5	10.8	24	—	11.5	25.4	29.4	
FH26W-37S-O. 3SHW(60)	CL580-2411-5-60	37	12.6	10.2	10.8	11.43	12.1	11.4	24	—	11.5	25.4	29.4	
FH26W-39S-O. 3SHW(60)	CL580-2412-8-60	39	13.2	10.8	11.4	12.03	12.7	12.0	24	—	11.5	25.4	29.4	D
FH26W-41S-O. 3SHW(60)	CL580-2413-0-60	41	13.8	11.4	12.0	12.63	13.3	12.6	24	—	11.5	25.4	29.4	
FH26W-45S-O. 3SHW(60)	CL580-2414-3-60	45	15.0	12.6	13.2	13.83	14.5	13.8	24	—	11.5	25.4	29.4	
FH26W-51S-O. 3SHW(60)	CL580-2415-6-60	51	16.8	14.4	15.0	15.63	16.3	15.6	24	—	11.5	25.4	29.4	
FH26W-55S-O. 3SHW(60)	CL580-2416-9-60	55	18.0	15.6	16.2	16.83	17.5	16.8	32	28.4	14.2	33.4	37.4	
FH26W-57S-O. 3SHW(60)	CL580-2417-1-60	57	18.6	16.2	16.8	17.43	18.1	17.4	32	28.4	14.2	33.4	37.4	
FH26W-61S-O. 3SHW(60)	CL580-2418-4-60	61	19.8	17.4	18.0	18.63	19.3	18.6	32	28.4	14.2	33.4	37.4	
FH26W-71S-O. 3SHW(60)	CL580-2419-7-60	71	22.8	20.4	21.0	21.63	22.3	21.6	44	40.4	20.2	45.4	49.4	E



DRAWING NO.
EDC3-323714-05

PART NO.
FH26W- * S-O. 3SHW(60)

CODE NO.
CL580

FORM HC0011-5-8 1

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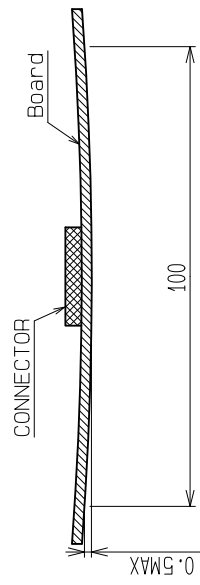
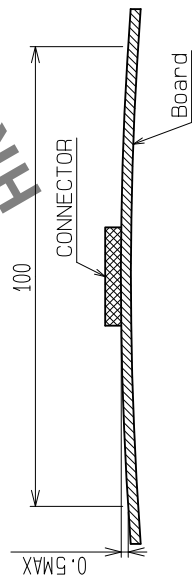
This connector is small and thin and requires delicate and careful handling.
Read through the instructions shown below and handle the connector properly.
Each value indicating here are for reference and may differ from standard value.

[INSTRUCTIONS FOR MOUNTING ON THE BOARD]

- ◆Warp of Board
 - Minimize warp of the board as much as possible.
 - Lead co-planarity including reinforced metal fittings is 0.1 mm or less.
 - Too much warp of the board may result in a soldering failure.
- ◆Flexible board design
 - Please make sure to put a stiffener on the backside of the flexible board.
 - We recommend a glass epoxy material with the thickness of 0.3mm MIN.
- ◆Load to Connector
 - Do not add 0.5N or greater external force when unreeel or pick and place the connector etc. or it may get broken.
 - In addition, do not insert the FPC or operate the connector before mounting it.

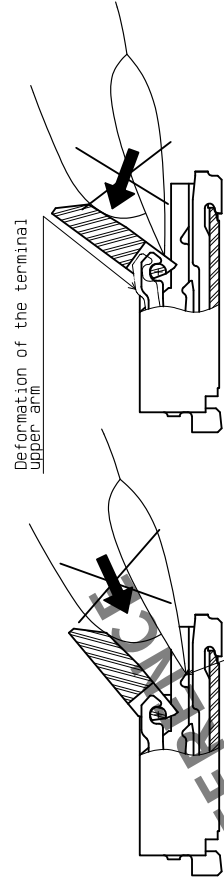
[INSTRUCTIONS FOR PCB HANDLING AFTER MOUNTING THE CONNECTOR]

- ◆Load to Board
 - Splitting a large board into several pieces
 - Screwing the board
 - Avoid the handling described above so that no force is exerted on the board during the assembly process.
 - Otherwise, the connector may become defective.
- ◆Amount of Warp
 - The warp of a 100-mm wide board should be 0.5 mm or less.
 - The warp of board suffers stress on connector and the connector may become defective.



[INSTRUCTIONS ON INSERTING FPC AND CONNECTION]

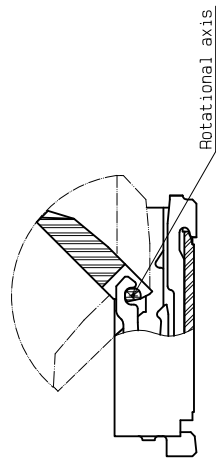
- ◆Use of the actuator
 - 1. Be very careful not to apply excessive force when releasing the actuator in the initial position (with no FPC inserted).
 - If you use your nail or finger as shown below, the terminals may be deformed.



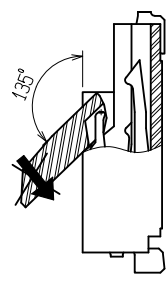
Example1.

Example2.

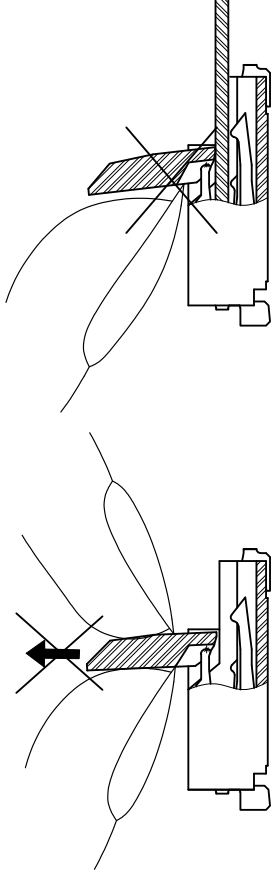
- 2. The actuator rotates around the rotational axis as shown below.
Rotate the actuator.



- 3. The actuator will not open more than 135°. Do not apply any force backward beyond this point. Otherwise, the actuator may come off or break.



4. Move the actuator at approximately the center.
5. Do not pinch or pick the actuator to lift it as shown below. Otherwise, it may break.
(Do not carry out any operation other than rotating the actuator as shown in 2 above.)



Direction of Contacts

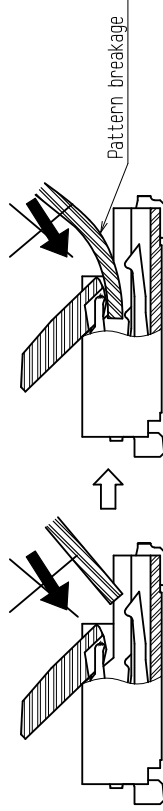
This connector has contacts on the bottom. Thus, insert the FPC with the exposed conductors face down.

Inserting the FPC

1. Insert the FPC horizontally along the surface and at a right angle to the connector.
Insert it properly to the very end.
If the FPC is inserted at a slant (incorrectly), the conductors may short-circuit due to pitch shift or the edge of the FPC may catch in the terminals, resulting in deformation of the terminals.
This connector has a ZIF structure, and its effective engagement length is 0.35 mm (when the recommended FPC nominal is used).
Use the actuator carefully to prevent the FPC from dislocating after inserting it.
2. Do not insert the FPC diagonally from above.
If the FPC is inserted at a slant (incorrectly) as shown below in the FPC insertion process, the FPC may bend and patterns may break or the FPC may not insert completely, resulting in improper conduction.

❗Keep a sufficient FPC insertion space in the stage of the layout in order to avoid incorrect FPC insertion.
Besides, it is not difficult to insert FPC correctly all the way to the end.
Design the proper layout of parts.

❗Make adjustments with the FPC manufacturer for FPC bending performance and wire breakage.



Checking the Locking Condition

In the locked condition, make sure that the actuator is horizontal on the board surface.
Do not apply excessive force to it near the 0° position of the actuator.
Otherwise, the terminals may be deformed. (Allowable force: 1 N or less)

INSTRUCTIONS ON FPC LAYOUT AFTER CONNECTION

Load to FPC

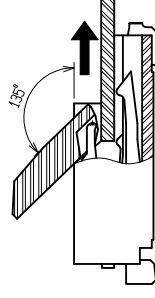
Be very careful not to apply any force to the FPC after inserting it.
Otherwise, the connector may become unlocked or the FPC may break.
Fix the FPC, in particular, when loads are applied to it continuously.
Design the FPC layout with care not to bend it sharply near the insertion opening.

Load: $0.05Xn$ N MAX
(n: number of contacts)



INSTRUCTIONS ON REMOVING FPC

Release the actuator to remove the FPC.



Instructions on Manual Soldering

Follow the instructions shown below when soldering the connector manually during repair work, etc.
1. Do not perform reflow soldering or manual soldering with the FPC inserted into the connector.

2. Do not heat the connector excessively. Be very careful not to let the soldering iron contact any parts other than connector leads. Otherwise, the connector may be deformed or melt.

3. Do not use excessive solder (or flux).

If excessive solder (or flux) is used on the terminals, solder or flux may adhere to the contacts or rotating parts of the actuator, resulting in poor contact or a rotation failure of the actuator.

Supplying excessive solder to the reinforcing bracket may hinder actuator rotation, resulting in breakage of the connector.

	DRAWING NO.	EDC3-323714-05
	PART NO.	FH26W-**-S-0.3SHWC60
	CODE NO.	CL580