Organic Field Effect Transistors for Neural Stimulation - In Vivo Tests

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Abstract

Pentacene is an organic semiconductor material, which can be integrated in micromachined polyimide-based high channel neural stimulation electrode arrays in order to build active switching matrices. Different to conventional direct one-to-one wiring between electrode contacts and stimulator output stages, a matrix of n x m electrode contacts requires only n + m interconnects to a stimulator, when active switching elements are used. In this paper, we demonstrate that Pentacene transistors can be used as switching elements and are capable of driving currents suitable for electrical stimulation of nervous tissue.

1. INTRODUCTION

Some neuroprosthetic applications, e.g. retina prosthesis, require a high number of individual electrode contacts arranged on a small geometrical area. Assuming an electrode array of m x n contacts, direct wiring of the contacts to a stimulator would require m x n individual wires and – in case of a hermetic housing of the stimulator electronics – m x n feedthroughs through the implant package. The feasibility is compromised by the restricted space and the high number of channels, resulting in a high integration density.
An alternative approach to direct wiring is an active switching matrix located at the actual electrode array which reduces the number of wires dramatically to m + n.
We have shown in previous publications that the organic semiconductor Pentacene can patterend to build an organic field effect transistor (OFET), which can be integrated into a micro machined polyimide-based micro electrode arrays [1]. In order to evaluate the potential of Pentacene OFETs for use in active electrode matrices, their performance as addressable switching elements as well as voltage controlled current sources for electrical stimulation of peripheral nerves had to be investigated.

2. METHODS

2.1. Transistor Technology

In the work presented here, we focussed on the properties of the semiconductor material Pentacene rather looking into its characteristics when integrated into a polyimide electrode array.

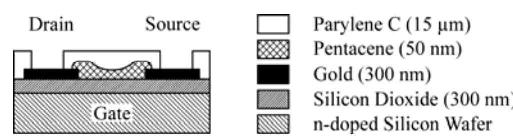

Figure 1: Cross section of the OFET.

For our experiments, an OFET was build which cross sectional structure is shown in figure 1. The active channel of the OFET had the length of 20 µm (distance between drain and source contact). Source and drain contacts were arranged inter-digitally with a total channel width of 72 mm (length for which the contacts run in parallel).

2.2. Driving the Transistor

Two programmable voltage sources were used to drive the OFET as a current source. One voltage source controlled the gate potential V_G, the other set the drain potential V_D. The source terminal of the OFET was used as output for the stimulation current i_{stim} (see figure 2). The voltage sources were capable of delivering pulses with amplitudes up to ±80 V of arbitrary width.

2.3. Neural Interface

A self-spiralling cuff electrode was used to interface with the nervous tissue. It featured three ring-shaped platinum contacts (width: 0.32 mm), arranged at a longitudinal pitch of 5 mm on the inside of an insulating polyimide cuff (inner diameter: 1.6 mm). Fabrication of the cuff electrode, its features and performance in-vivo are described in detail in [2].

2.4. In-Vitro Set Up

A sciatic nerve was explanted from an adult Xenopus Laevis frog and used as a representation of excitable tissue. The nerve was kept at room temperature in amphibian ringer's solution. The cuff was wrapped around the nerve. One end of the nerve was elevated from the ringer's solution and rested on bipolar platinum hook electrode, which was connected to a low-noise amplifier with a gain of 30k and a bandwidth of 0.5 to 5 kHz. Triggered by the voltage sources V_D and V_G, an oscilloscope sampled the output of the amplifier at 25 kHz, 8 bit and transferred the data to a PC.

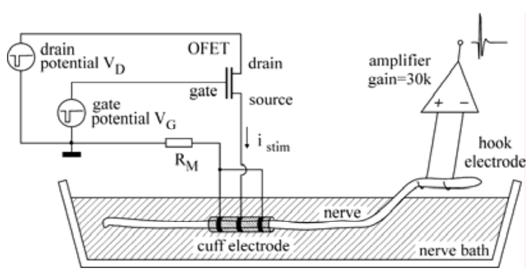

Figure 2: The setup for nerve stimulation and recording. The resistor R_M was used to measure the current through the electrode.

2.5 Experimental Protocol

In order to investigate the current output characteristics of the OFET, the drain potential V_D was set to -10 V, while the amplitude of the gate potential was increased from 0 to -50 V in steps of -10 V. Instead of an electrode, a 1 kΩ resistor was used as load. The current through the resistor was measured as stimulation current i_{stim}. The experiment was repeated for V_D=-20 V and V_D=-30 V.

The first neural stimulation experiment was the synchronous pulsing (0.3 ms pulse width) of V_D and V_G. V_D was kept at constant pulse amplitude while the pulse amplitude of V_G was increased from 0 V to -50 V in steps of -5 V.

The OFET was connected to the cuff electrode as shown in figure 2. The activity of the nerve was recorded.

The OFET was intended to be used as an switching element in an active electrode matrix. In order to investigate the feasibility of matrix-addressing, asynchronous pulsing of V_D and V_G was performed. If matrix addressing is possible, i_{stim} should be zero in all cases except for synchronous pulsing of V_D and V_G.

3. RESULTS

3.1. Current Output Characteristics

In order to drive a current of an amplitude beyond 0.1 mA, voltages of several 10 Volts were required. A combination of V_D=-10 V and V_G=-50 V, e.g., led to a current of 1.2 mA (see figure 3).

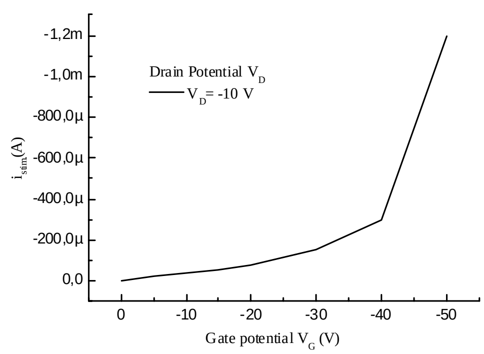

Figure 3: OFET Current output characteristics.

3.2. Recruitment Curve

A stimulation current i_{stim} was driven by synchronous voltage pulses (0.3 ms) at the gate and drain contacts, which caused compound action potentials (CAP) to be propagated along the nerve. Keeping the amplitude of the V_D pulse at -10 V and increasing the amplitude of the V_G pulse, first led to a recruitment of fast fibres (V_G=-30 V). At higher amplitudes (V_G=-40 V) the activity of slower fibres contributed to the CAP (see figure 4).

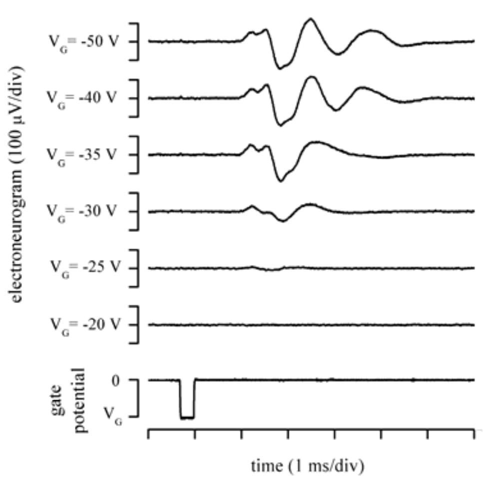

Figure 4: Neural recruitment curves obtained by gradually increasing the gate potential V_G, while the drain potential V_D was set to -10 V.

3.3. Matrix Addressing Characteristics

No neural activity was recorded when only one of the contacts were pulsed. Synchronous pulses at both contacts, led to excitation of the nerve and cause a CAP to be propagated (figure 4).

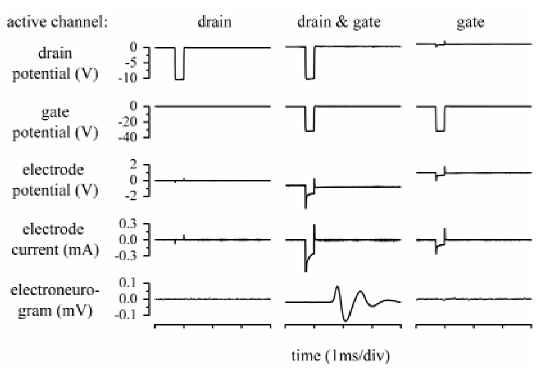

Figure 5: Activating gate and drain separately did not result in nerve excitation. Synchronous activation stimulates the nerve.

4. DISCUSSION AND CONCLUSIONS

Figure 3 shows, that relatively high voltages have to be applied to gate (several -10 V) and drain (-10 V) contacts of the OFET to obtain currents up to 1 mA. However, taking retina stimulation as one of the potential application of an active electrode matrix, much smaller currents are expected to be sufficient. Literature reports of current amplitudes between 2 µA [3] and 100 µA [4] at pulse widths between 100 µs and 500 µs. This amplitudes required a driving voltages of about -10 V when our OFET switch is used, which seems to be within practical limits.

Figure 4 shows a gradual recruitment of nerve fibres is possible. Because of high on-resistance the OFET does not work as a pure switch but as a voltage controlled current source, which can used to provide a range of different levels of neural excitation.

Figure 5 demonstrates that matrix addressing of the OFETs is possible. Activation of one OFET in a matrix will not result in activation of another OFET. However, some crosstalk was found: When the gate potential V_G was switched to -30 V while the drain potential V_D was kept at 0 V, a small current flowed through the electrode. We assume, that this current could be caused by a pinhole in the insulation layer between gate and source contact of the OFET.

Future work will imply further miniaturization of the OFETs (shorter channel length) and the integration into polyimide electrode arrays. The brittle silicon dioxide layer will be replaced by a flexible ultra-thin polymer layer. Furthermore, the long-term stability of Pentacene has to be investigated, especially when voltages applied over a long time span in aqueous environment.

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