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PDMS electrodes for recording and stimulation

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Abstract: The usability of flexible electrodes in moving environment is limited due to different mechanical characteristics of their metallic and polymeric components. To achieve structure compatible electrodes, all used materials need to have similar Young's moduli as the surrounding tissue. This paper describes the characterization of macroscopic as well as miniaturized electrodes entirely made out of modified silicone (PDMS). Electrochemical, mechanical, biological, optical, and applicative methods were used. It could be shown, that PDMS electrodes are capable to be used for recording electrocardiograms with similar form and amplitude as with standard electrodes.

Keywords: cardiomyocytes; characterization; charge injection capacity; electrocardiogram; impedance; silicone electrodes; structure compatibility; Young's moduli.

1 Introduction

Electrodes serve as electrical interface between biological tissues and technical systems. They are capable for recording bioelectrical potentials or stimulation of tissue. Surface electrodes as well as implantable ones are used in many sectors of medicine and medical engineering. Nevertheless, an application of flexible electrodes in moving muscular environment causes mechanical stress within the structures and to the tissue. Cardiac muscles changes their length up to about 10% [1], other types even more [2]. The Young's moduli (YM) of the tissue and the materials used for flexible electrodes often distinguish from each other. The conductive structures of electrodes are usually made out of metals (e.g. gold, YM 78 GPa [3]) or intrinsic conductive polymers (e.g. PEDOT:PSS, YM 1 GPa [4, 5]). The use of polymeric substrates like polyimide (PI, YM 10 GPa [6]) or

polydimethylsiloxan (PDMS, YM 9 MPa [7]) provides the flexibility [5, 8]. The Young's modulus of muscle tissue is in the range of 10^{-3} Pa (not active) to 20 MPa (tetanic) [9]. These differences limit the long-term stability of micro electrodes used in moving tissue. Hence, the development of stretchable and thus structure compatible electrodes is of high interest.

This work was aimed to characterize electrodes entirely made out of PDMS as monomorphic material as well as the used raw materials.

2 Methods

The conducting paths of the electrode were made of a PDMS composite (MED-6015, Nusil) including carbon nanotubes (CNT, Sigma-Aldrich). Using them to form the composite material, arose in an electrical resistance, that was independent of elongation [10]. Different amounts of CNT were compared regarding the specific conductivity of the resulting composite. For the electrode contact, an established material for dry surface electrodes was used [11]. To reduce the YM of the materials, a silicone thinner (ST, Smooth-ON) was added. A soft lithographic fabrication process was used to connect all PDMS composites as different components of the electrode. For structuring the conduction paths, a screen printing based technique was used. Detailed information for this method can be found elsewhere [12]. Two electrode designs – a macroscopic monopolar and a miniaturized bipolar version – were investigated (see Figure 1).

To characterize the electrodes and base materials electrical, electrochemical, mechanical, optical, biological and applicative methods were used. The specific conductivity of the CNT-PDMS material was measured by using a four point setup. For mechanical characterization, i.e. to determine the YM, tensile tests were carried out. The homogeneity of the particles within the polymer matrix was visualized with confocal Raman microscopy [13]. To evaluate the technical characteristics of the electrodes, electrochemical impedance spectroscopy, cyclic voltammetry (CV), and pulse tests (PT) were performed. Therefore, a three electrode setup in 0.9% sodium chloride solution was used. Using CV the voltage boundaries (VB) of both macroscopic

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and miniaturized electrodes were determined. Based on these values, the charge injection capacity (CIC) could be evaluated by analysis of the voltage response to current pulses (details in [14] and [15]). The current was measured to calculate the current density at the electrode surface. It was determined by summation of the discrete values of the current pulses and dividing it by the sampling frequency (2.5 Hz). The tolerance to cells could be shown by direct cell contact tests. Therefor, human lung fibroblasts (LGC Standards) were used. Additionally the electrodes were

colonized with human induced pluripotent stem cell derived cardiomyocytes (hiPS-CM, AxioGenesis) to track their behavior over time. This required a preliminary coating of the electrodes with matrigel (BD Biosciences). Electrocardiograms (ECG) were recorded over extremity acquisition (Einthoven I method). For comparison Ag/AgCl-electrodes were used.

3 Results

The specific conductivities of the CNT-PDMS composites are shown in Figure 2. Due to the addition of CNT into the PDMS a specific conductivity of (31.5 ± 0.9) S/m could be achieved. In wide ranges the CNT were homogeneously dispersed, but in the inner of the material little aggregations of silicone could be observed (see Figure 3). The YM of the different materials are listed in Table 1.

Figure 1: PDMS electrodes (top: macroscopic electrode, bottom: miniaturized electrode, also shown in [12]).

Figure 2: Specific conductivity of CNT-PDMS at different particle concentrations in the range of good processability.

Table 1: Young’s moduli (YM) of the electrode materials and standard deviation; also presented in [12].

Material	YM in MPa
Substrate (N = 6)	0.99 ± 0.02
Conductor (N = 13)	1.00 ± 0.10
Electrode contact (N = 9)	4.57 ± 0.17

The impedance spectra associated to the electrodes are shown in Figure 4. For frequencies between 10 Hz and 25 kHz, an ohmic (i.e. frequency independent) behavior was observed for both electrode sizes. The impedance

Figure 3: Raman microscopic images (left, red: CNT signal, green: PDMS signal; scale bar: 50 μm) and correlations graph (right) of a cross section at the border of the conductor material.

Figure 4: Impedance spectra (top: macroscopic electrode, bottom: miniaturized electrode).

magnitude of the macroscopic electrodes (diameter 1 cm) lay between 3 kΩ and 35 kΩ (median 5 kΩ). The electrodes with a diameter of 0.4 cm had impedances in the range between 6 kΩ and 102 kΩ (median 37 kΩ). The CV did not provide a clear water window as for metallic electrodes. Hence, as VB values were chosen, at which the CV did not possess peaks, and thus no signs for chemical reactions were apparent. The negative boundary was defined as the particular voltage turning point, at which in case it was overrun, the first overlap of the curves occurred (see Figure 5, left). The positive potential turning point whereby exceeding a maximum arises within the curve was defined as positive VB (see Figure 5, right). Potentials larger than the VB may firstly result in reversible processes. Visible hydrolysis could not be observed, even at potentials up to ± 3 V. The PT on exemplary demonstrator electrodes yielded to electrode potentials shown in Figure 6. The resulting VB and CIC are summarized in Table 2.

Figure 5: Cyclic voltammograms for evaluation of the voltage boundaries (VB) of a macroscopic electrode; left: negative VB (-0.5 V), right: positive VB (1.1 V); effects occurring at overrun are marked by arrows.

Figure 6: Minimal and maximal electrode potentials measured as voltage response to a biphasic current impulse vs. the related current densities at the electrode surface; left: macroscopic electrode, right: miniaturized electrode.

Table 2: Voltage boundaries (VB) and charge injection capacities (CIC, $\pm$ standard deviation; also in [12]) of macroscopic (diameter 1 cm) and miniaturized (diameter 0.4 cm) electrodes.

Electrode	VB in V	CIC in $\mu\text{C}/\text{cm}^3$
Macroscopic	-0.5 to 1.1	0.13 ± 0.02
Miniaturized	-0.6 to 1.3	1.40 ± 0.34

In direct cell contact non of the used materials caused any visible cell damage or alteration. Furthermore, hIPS-CM attached successfully to the electrode surface. In Figure 7, a dense cell layer can be seen. Videos of beating cells were recorded and analyzed over 24 days. The cells grown on the electrodes regularly beat with frequencies between 2.8 ± 0.2 bpm and 18.9 ± 4.7 bpm. After 10 days no beating could be observed any more. At day 13 and day 24 live/dead staining of cells cultured on the electrodes were performed. For comparison all investigations were also performed with hIPS-CM cultured on conventional polystyrene substrates. The mean beating

Figure 7: Transmission microscopic view of hIPS-CM grown directly next to a PDMS electrode (day 7, electrode contact in the lower right corner).

Figure 8: ECG (Einthoven I) comparison between macroscopic and miniaturized electrodes (PDMSmacro and PDMSmini) and standard electrodes (Ag/AgCl).

frequency of this cell control was 2 times higher than that of the cells at the electrodes. After 13 days approximately half of the cells were still alive. At day 24 all cells on the electrodes were dead, while the cells of the control entirely were alive. The physiological ECG, shown in Figure 8, visualize the comparability of the PDMS electrodes to standard Ag/AgCl electrodes. The signals of all three electrode types had a similar form and amplitude.

4 Conclusion

Macroscopic and miniaturized PDMS electrodes were characterized. Electrically conductive and non-conductive silicone composites were used as materials for all electrode components. The YM of the polymers and (tetanic) muscle tissue were in the same range. The cell compatibility of the components to lung fibroblasts could be shown. Also, high sensitive hIPS-CM retained their vitality for more than 10 days in direct contact to the electrodes. Hence, for the electrodes a structure and surface compatibility has been achieved. The beating frequency of hIPS-CM has been in the same range as published in literature [16], but also lower beat rates were observed. This may be due to a non optimal distribution or absorption on the matrigel layer. Compared to metal electrodes of similar size (area 1 cm^2 [15]) the PDMS electrodes (area 0.79 cm^2) show a much higher impedance for frequencies larger than 1 Hz (three orders of magnitude at 10 kHz). Although the median values of the macroscopic and the miniaturized electrodes were 5 k Ω and 37 k Ω , respectively, the acquired ECG signals had a similar form and amplitude compared to the recordings of standard

electrodes. The determined CIC values are much lower than that of metallic micro electrodes (Pt electrodes, diameter 300 μm , CIC 64 $\mu\text{C}/\text{cm}^2$ [15]). However, 2008 Cogan described in his review the CIC as indirect proportional to the size electrode [17].

It could be shown, that PDMS electrodes can be used for recording physiological signals and for stimulation purposes. The fabrication process [12] is capable for production of functional miniaturized and macroscopic electrodes, both as bipolar and monopolar structures.

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