Organic Electronics 18 (2015) 89–94

Contents lists available at ScienceDirect

Organic Electronics

journal homepage: www.elsevier.com/locate/orgel

Letter

Inkjet printed flexible electrodes for surface electromyography

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ARTICLE INFO

Article history:
Received 28 August 2014
Accepted 12 December 2014
Available online 12 January 2015

Keywords:
Inkjet printing
Surface electromyography
Silver nanoparticles

ABSTRACT

Recording of surface ElctroMyoGraphic (sEMG) signal represents a challenge, due to the nature both stochastic and deterministic of the biopotential. The sEMG signal results from the superimposed activity of a high number of motor units, artifacts, and background noise. Among the different non-invasive techniques to measure the muscle electrical activity, recent studies showed how High-Density surface EMG (HD-sEMG) constitutes a very effective tool. HD-sEMG uses multiple closely spaced electrodes overlying a restricted area of the skin and provides high definition temporal and spatial information on muscle activity. To optimize the features of the current devices for HD-sEMG signal detection, this paper reports on the realization of the first inkjet printed HD electrode matrix. The matrix was built by inkjet printing of a commercial silver-based ink on a flexible Kapton® substrate. While bringing about all advantages of drop-on-demand inkjet technology, such as rapid prototyping and simple CAD re-desing of customizable devices, the printed matrix produces interesting electrical results in terms of resolution, resistance and electrode–skin contact impedance. Electrode–skin impedance obtained with our measurement setup was indeed always in line, while not better than the one obtained with commercial electrodes.

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ElectroMyoGraphy (EMG) is extensively used as a technique to acquire information about the clinical state of the muscles and as a source of control information for prosthetics [1] and telerobotics. In recent years surface EMG (sEMG) usage has been extended with the aim of bringing it outside of laboratories and into everyday usage. In this direction Thalmic Labs, with the Myo armband, have demonstrated many possible uses of such a technology. Specifically, the EMG biopotential is the result of signal inputs from cortex to neurons. Neurons transmit these electrical signals to muscles, allowing the movement [2]. There exist two techniques for EMG: intramuscular EMG (iEMG) and surface EMG (sEMG): iEMG requires needle electrodes to be implanted inside the muscle; sEMG, instead, requires the electrode to contact the skin over the muscle [3]. While iEMG returns cleaner information about the local behavior of a muscle fiber [3,4], sEMG allows to get a broader look at the instantaneous state of the muscles underneath the electrodes [5–8]. In the clinical domain sEMG cannot usually supersede the precision of iEMG [3,9–11] but, at the same time, many scenarios, such as prosthetics [1,12], biofeedback [13,14], ergonomics [15,16], sport [17], and space medicine [18–20], are not compatible with the inserting of needles in the muscles. In recent years High Density sEMG (HD-sEMG) arose as a non-invasive technique to acquire precise information about the muscle by increase in the cardinality of electrodes and advanced signal processing techniques [11,21–23]. While this technique resulted very promising in laboratories, it still struggles to become a streamline application. The main limit of HD-sEMG is linked to the small size of electrodes. In laboratories the problem is solved by improving the

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http://dx.doi.org/10.1016/j.orgel.2014.12.017
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electrical contact between the electrodes and the skin by application of a conductive gel or cream. Such an operation is unpractical and requires some time and expertise. Moreover, when the gel dries off contact impedance worsten, thus reducing the use of such 'wet' electrodes to relatively short experiments. In this work we present the first steps towards the production of customizable low impedance electrodes for HD-sEMG by means of inkjet printing.

We present here details about the realization of the first electrode matrix by inkjet printing of a commercial silver-based ink [24,25] on a flexible Kapton® substrate [26]. Inkjet printing ensures rapid prototyping of sensors and devices [27,28], so it is one of the most convenient solutions for research laboratories and education institution [29]. Furthermore, this technique allows to change dimensions, cardinality, and disposition of the electrodes by mean of simple CAD re-desing [30]. This flexibility enables to rapidly create new modified prints of the device. The custom design and the methods of realization represent a new frontier for flexible sensors for biomedical applications. In this paper we present results showing how this techniques returns electrodes which results in lower skin–electrode impedances thus opening the way for further studies in this domain.

Besides the deposition technology, some of the authors have previously developed a specific nanocomposite formulation, based on a polymeric matrix and silver nanoparticle dispersion, whose application have been shown in the high frequency range, up to 1 GHz [31].

To manufacture the first printed electrode array we used silver nanoparticle (NP)-based ink on a Kapton® substrate. Silver NP-based ink is basically an aqueous colloidal/copolymer dispersion of silver NPs (InkA-C100, Politronica® Inkjet Printing). Its properties are listed in [32]. The substrate is a polyimide, Kapton® film (Dupont, Wilmington DE) of 75 μm in thickness. Its properties are shown in [33].

For the insulation treatment of the printed matrix, a solder masking agent (Chemask® LF – Lead Free – Chemtronics) and an insulating silicone-based liquid (DCA Modified Silicone Conformal Coating (SCC3)) were chosen.

To test the electrodes in their wet condition, we needed a commercial gel (SpesMedica s.r.l.) for sEMG.

The innovative sEMG recording system was realized using a piezoelectric Jetlab® IV printer from MicroFab Technologies Inc. equipped with a 60-μm nozzle diameter MJ-AT-01 dispenser. In piezoelectrically driven drop-on-demand devices there is a piezoelectric element used to change the volume of the ink reservoir in order to produce the fluid ejection and retraction pressure pulse. The advantage of piezoelectric actuation is that the pressure pulse rise and fall times can be tailored to optimize monodisperse satellite-free drop production and dynamically alter the diameter of the ejected drops [34]. We realized the design of electrode array with AutoCAD™ 2010. The print head was heated at 60 °C and the substrate at 70 °C on a hotplate during the deposition process, to improve deposit quality. The waveform to expel a single droplet was:

- Time: rise 12 μs, dwell 18 μs, fall 4 μs, echo dwell 38 μs, final rise 2 μs.
- Voltage: dwell voltage +28 V, echo dwell voltage −10 V.

DC voltage was 0 V and fly velocity was set at 50 mm s^{-1}. Step size was set to 100 μm and 6 layers were printed on top of each other, to obtain the desired value of resistivity.

After the printing step, a first thermal treatment was performed in order to promote NP coalescence [35], allowing complete solvent removal and electrical percolation. A Laboratory muffle oven was used with this curing profile: from ambient temperature up to 300 °C with a ramp of 0.5 °C min^{-1}; subsequently a plateau of 30 min at 300 °C and then from 300 °C to 50 °C with a slope of 5 °C min^{-1}.

Furthermore, to make the printed electrode array more resistant, we electrically insulated the silver lines. During this step, we worked under hood. Initially, we used a solder masking agent (Chemask® LF – Lead Free – Chemtronics) to protect pitches and connection zones. A syringe allowed the deposition of the masking agent directly on the parts to protect. After about 30 min, when the mask was dry, the printed electrical circuit was ready to insulation treatment. To avoid problems due to cracks, we decided to use insulating silicone-based liquid (DCA Modified Silicone Conformal Coating (SCC3)) rather than acrylic. Before silicone crosslinking, the mask was removed with tweezers.

After that, we applied a new curing treatment, to increase silver lines' resistance to chemical reagents: 2 h at ambient temperature under vacuum, reducing gas bubbles formation, followed by 4 h on the hot-plate, with a temperature increasing gradually until 120 °C.

After every treatment, conductivity measurements were performed using a digital multimeter to evaluate the integrity of electrodes and the resistance between each electrode pitch and its own corresponding silver line at connection zone.

Once the fabrication of the electrode array was finished, an electrical connection for experimental proofs was made. To accomplish this task, a corresponding Printed Circuit Board (PCB) was constructed, matching the exposed, silver lines. Electrical connection between the electrode array and corresponding PCB was obtained by using a FFC connector 738-8912. During connectors' soldering and matching of printed electrode matrix and PCB, we worked under optical microscope.

Therefore, the sEMG sensors were connected to the measurement setup and placed on a limb [36], so that a couple of printed electrodes directly faced a couple of commercial electrodes at a distance of less than 1 cm. Firstly, by the Potentiostat 700D CH Instruments we tested the impedance of the electrodes in their dry conditions [2] on eight different subjects. Indeed, no gel or cream was applied and the contact between the electrodes and the skin was insured by the pressure produced by the sphygmomanometer. The impedance of the electrodes in their dry usage was tested between 10 and 1000 Hz [2], and at three different pressure levels: 13.33, 26.7, 53.3 kPa (limit value). The setup guaranteed a pressure value constant among the electrodes and in time during the whole measure. Secondly, electrodes in their wet condition have been tested. We applied to both matrices a foam designed to allow accommodation of a small quantity of commercial

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