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Fully printed all-polymer tattoo/textile electronics for electromyography

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Keywords

Inkjet printing, organic electronics, tattoo, textile, electrophysiology

Abstract

Driven by the ever-growing needs for developing low cost, easy-to-use, noninvasive diagnostic tools, biomedical devices that can be integrated on human skin or textiles have begun to emerge. These “wearable” devices should couple electronics directly to the human skin and detect a variety of biologically relevant signals such as the neuromuscular activity. In this work, we develop a simple, low cost and customizable device to perform electromyography (EMG) measurements based on electronics

1 fabricated on a tattoo paper. The electrodes are based on the conducting polymer poly(3,4-
2 ethylenedioxythiophene) doped with polystyrene sulfonate (PEDOT:PSS) and inkjet-printed on the
3 conformable tattoo paper. Addressing the integration challenge common for such flexible electronic
4 devices, we connect the tattoo electrodes to the acquisition system through a textile in the form of a
5 wristband comprising of printed PEDOT:PSS contacts. While the textile wristband conforms around the
6 “tattooed” skin, it enables a reliable contact with the electrodes beneath. We show that this tattoo/textile
7 electronics platform is able to monitor the biceps activity of the arm during muscle contractions lasting
8 for a period of seven hours, with comparable performance to conventional biopotential electrodes and
9 without the use of gels or expensive metallic materials. Combining the tattoo electronics with the
10 electronic textile improves the communication of skin-like electrodes with external electronics, renders a
11 reliable and versatile system for detecting biopotential signals critical for myoelectric prosthesis, muscle
12 injury prevention and/or detection.

30 Introduction

33 One of the greatest endeavors of current medical diagnostics research is to develop devices that can record
34 vital signals from the body in real time whilst being minimally invasive [1]. The sensor technologies thus
35 move from lab-on-chip platforms to wearable electronics, indicating that sensors should be integrated with
36 the skin or clothes that surround the skin, allowing for personal health monitoring as well as drug delivery
37 systems [2–8]. Electrically interfacing the skin with electrodes for monitoring the electrical activity related
38 to functioning of cells is a well-established practice in clinic, albeit performed with bulky rigid electronics
39 that are not comfortable to wear. For instance, standard devices to record muscle activity, i.e.
40 electromyography (EMG), are nonpolarizable Ag/AgCl electrodes which are applied on the skin typically
41 with gels that allow for ion/water penetration and therefore reduce skin/electrode impedance. The
42 performance of these electrodes depends on the gel-material and as such they pose several limitations
43 including discomfort, allergies or irritation when in contact with the human skin, as well as limited stability

1 related to gel drying out during long-term applications [9]. New materials and concepts have emerged in
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3 order to address these aforementioned limitations, aiming to develop devices that can adhere seamlessly
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5 to skin as well as adapt to its movements. Conformable electrodes based on smart textiles are prime
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7 devices generated to fulfill these requirements [10–13]. Someya and coworkers, for instance,
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9 demonstrated new conductors made of highly stretchable materials composed of Ag flakes mixed with
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11 fluorines. These materials could be integrated into garments via screen printing and used for recording
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13 electrophysiological signals [12]. Graphene has been another material integrated with nylon fabrics [13].
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15 In parallel to the development of smart textiles, a new generation of devices that can be fully conformable,
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17 known as electronic skin or e-skin, [7,14,15] has emerged. The pioneering work in this field incorporated
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19 electrophysiology, temperature and strain sensors made of gold, thin silicon, gallium arsenide in the form
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21 of nanoribbons and nanomembranes [15]. These materials were deposited directly on the skin by
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23 dissolving a water soluble film of polyvinyl alcohol, leaving the electronics adhered to the skin by van der
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25 Walls interactions [15]. Recently Bao's team proposed a new route for the fabrication of "epidermal"
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27 circuitry comprising transistors arrays, based on stretchable polymer materials such as carbon nanotubes
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29 and conjugated polymer/elastomer mixtures, and developed integrated touch-sensors able to detect the
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31 footprint of an artificial ladybug [8]. These devices were transferred to the skin using a water-soluble
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33 sacrificial layer as well. A promising substrate material for device integration has in fact been the
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35 commercially available tattoo paper. If the electronics can be integrated on the paper, it can be transferred
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37 to the skin by dissolving the water-soluble sacrificial layer (typically made of starch-dextrin) of the paper.
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39 As such, the electronics conforms onto the complex features of the skin and represents an ideal disposable
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41 system for performing noninvasive daily diagnostics of a variety of biomarkers. These tattoo integrated
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43 electronic platforms exhibit excellent skin conformation, resistance to deformation and good
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45 biocompatibility with the human skin [2,16–18].

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47 Organic electronic materials have come to the fore as promising materials in interfacing the tissue due to
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49 the mechanical softness and biocompatibility as well as the low impedance of their films [19]. Amongst
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1 these materials, poly(3,4-ethylenedioxythiophene) doped with polystyrene sulfonate (PEDOT:PSS) has
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3 been the standard organic electronic coating, owing its popularity mostly to its electrochemical stability
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5 (no degradation in aqueous environments rich in oxidative species) and being readily available as a
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7 dispersion in water through commercial vendors [20]. Another reason for the current popularity of
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9 PEDOT:PSS is that its dispersion can be processed through various means, allowing for integration into
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11 a wide range of existing fabrication procedures. Co-solvents, secondary dopants, and crosslinkers are often
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13 added to the dispersion to modulate the film conductivity, mechanical properties as well as viscosity so
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15 that it is ideal for printing and coating over large areas [21–27]. All these properties render PEDOT:PSS
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17 an ideal material to cast on unconventional substrates such as the tattoo paper. Zucca et al reported
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19 different techniques for the fabrication PEDOT:PSS based tattoo nanosheet electrodes. These involve spin
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21 casting the polymer on a commercial tattoo paper and inkjet printing a NaClO solution for the subtractive
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23 patterning of the tattoo [28], as well as coating poly(D,L-lactic acid) and PEDOT:PSS on a polypropylene
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25 substrate with a gravure roll [29]. In both cases, after transferring the tattoo electrode on the skin, EMG
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27 signals could be successfully recorded. The integration of interconnects is however challenging for tattoo
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29 electronics and needs to be considered for stability and use for long term measurements. Bareket et al
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31 developed tattoo electronics based on screen printed carbon electrodes and plasma polymerized
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33 PEDOT:PSS [30]. The device recorded EMG signal for 3 hours, however the overall thickness of the
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35 electrodes (ca. 100 μ m) is a concern for the conformal contact between the skin and the electrode.
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39 For these disposable and organic electronics-based tattoo electrodes to be ever commercialized, the
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41 fabrication should be cost and time effective. Solution processing technologies such as spraying
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43 techniques [31], screen printing [32,33], or inkjet printing [26,34] not only involve a small number of
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45 fabrication steps, but are also compatible with large-area flexible substrates. Inkjet printing, in particular,
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47 allows for the digital fabrication of customizable designs at atmospheric conditions, with patterns that can
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49 be easily modified on demand, ideal for fabricating diagnostic devices on flexible substrates [35,36]. A
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51 recent work of Ferrari et al demonstrated PEDOT:PSS electrodes printed on a tattoo paper which could
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1 measure muscular activity with high precision on limbs and face [37]. The electrode allows for hair growth
2 underneath, a great advantage over conventional systems promising for long-term recordings on areas
3 with high hair density. For practical applications, however, where the measurements are intended to be
4 carried on the skin for longer periods of time, the integration of the tattoo with the acquisition system
5 remains as a challenge that needs to be addressed. Design solutions should exclude adhesives, insulating
6 layers as well as metallic interconnects in order to reduce fabrication costs and involve a user-friendly
7 interface for these disposable systems. It is also important that all the fabrication procedure is
8 accomplished in one printing step.
9

10 In this work, we present a simple process for the fabrication and integration of a fully inkjet-printed e-skin
11 device based on tattoo technology. Using PEDOT:PSS both as the active electrode as well as contact
12 material, we record high quality EMG signals from the arm's biceps over 7 hours. Replacing conventional
13 electrical wires or any metal contact fabricated via evaporation which limit the feasibility of e-skin
14 technologies, we utilize PEDOT:PSS contacts printed on a conformable textile wristband. We characterize
15 this "tattoo/textile electronics" system and investigate the performance of the devices compared to
16 commercial electrodes used for EMG recordings. The thickness and roughness of the tattoo electrodes and
17 electrical properties of the tattoo/textile electronics are evaluated. The tattoo/textile electronics-skin
18 contact impedance at the relevant frequency range are measured. With the textile wristband conforming
19 on the skin and the intimate adhesion of the tattoo, we are able to record stable EMG signals with
20 comparable performance to conventional wet (gelled) Ag/AgCl electrodes. The proposed platform paves
21 the way towards easy-to-fabricate and use, conformable, all-polymer biopotential electrodes that can be
22 utilized in a wide range of diagnostics and as well as therapy applications.
23

24 **Experimental**

25 *Fabrication of PEDOT:PSS based electronics on tattoo and textile:* PEDOT:PSS ink was prepared using
26 a mixture containing PEDOT:PSS dispersion (conducting polymer, Heraeus, Clevis PH1000), 20 wt%
27 ethylene glycol (EG, co-solvent, Sigma Aldrich) and 0.8 wt% glycidoxypropyltrimethoxysilane (GOPS,
28

1 crosslinker, Sigma Aldrich). While EG increases the conductivity of the films deposited, GOPS prevents
2 their delamination during the transfer of the tattoo onto the skin. In this mixture, we also included 0.3wt%
3 of commercially available surfactants and a combination of organic solvents to render rheological
4 properties of the ink suitable for inkjet process without compromising softness and electrical conductivity.
5
6 A Dimatix DMP-2800 inkjet printer was used to fabricate the tattoo/textile electronics. 1cm² square
7 pattern of PEDOT:PSS (single layer) was printed on a commercially available temporary tattoo sheet
8 (Tattoo 2.1, the Magic Touch Ltd., 123 Applications). The sheet is composed of a single decal transfer
9 paper (including 3 layers : a paper carrier, a starch–dextrin water soluble sacrificial layer, and a layer of
10 ethylcellulose (ca. 400nm in thickness). For the wristband, we printed up to 8 layers of PEDOT:PSS ink
11 on a commercially available satin ribbon (Ideatiss). The metallic snaps buttons were installed on the ribbon
12 via a snap plier. The drop spacing for all printing procedures was 15 μm. Once the ink was printed, the
13 samples were placed in a conventional oven to cure the films at 110°C for 60 min. The tattoo electrodes
14 were transferred to the skin (no pre-treatment) by dissolving the water-soluble sacrificial layer and
15 removed by washing the exposed skin with soap and water.
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33 *Electrical characterization of the tattoo electrodes:* A four-point probe system (Jandel) was used to
34 determine the sheet resistance of PEDOT:PSS films. We used a mechanical profilometer (Ambios
35 Technology) to estimate the thickness of a single PEDOT:PSS layer printed on the substrates. The
36 electrical resistance of the PEDOT:PSS contact on the textile as well as the one between the tattoo
37 electrode and the textile contact was measured using a digital Multimeter (Amprobe 37XR-A Tru-rms).
38
39 Electrode–skin contact impedance measurements were performed as shown in Figure 2. The working
40 electrode (e.g. the tattoo/textile electronics) and the counter electrode (wet Ag/AgCl, Ambu blue sensor)
41 were placed on the forearm of a volunteer, keeping 3 cm distance from each other. The reference electrode
42 (wet Ag/AgCl, Ambu blue sensor) was placed on the elbow. A sinusoidal potential of 10 mV was applied
43 at the working electrode using a potentiostat (Metrohm Autolab B.V.) with no constant differential voltage
44 on the skin. The impedance was screened for a frequency range between 0.1 and 1000 Hz and the data
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1 were analyzed using NOVA software. We evaluated the electrical stability of the tattoo electrode upon
2 mechanical deformation. We transferred the tattoo across the distal phalange of the index finger that is
3 exposed to multiple flexions/extensions. Two copper tapes adhering to the printed PEDOT:PSS electrode
4 on opposite edges were used to establish electrical connection with the acquisition system. The
5 flexion/extension was repeated 250 times at 30° and at 90° angle of closure. The resistance of the electrode
6 was measured at every 10 repetitions.

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15 *EMG recordings and data acquisition:* Three able-bodied volunteers (2 males, 1 female, ages between 25
16 and 40 years old) participated in this study with informed consent acquired before conducting the
17 experiments. All participants took part for the long-term EMG recordings and one volunteered for the
18 measurement of the electrode-skin impedance. All protocols and procedures were approved by the
19 direction of research of the Ecole des Mines de St. Etienne. Two tattoo electrodes were placed 2 cm apart
20 from each other at the center of the biceps of the left arm. The contacts printed on the textile wristband
21 were placed on top of the tattooed areas of the skin in order to connect the recording and counter electrodes
22 to the acquisition system. 2 wet Ag/AgCl electrodes (Metrohm Autolab B.V.) were placed on the biceps
23 of the right arm. In order to have comparable signals, the volunteer was asked to lift 1 kg of a standard
24 weight with each arm simultaneously. A ground electrode was placed on the right leg. For each recording,
25 the experiment has been repeated at least 6 times, and the data were collected once in every 3 hours over
26 7 hours ($T_0=1h$, T_0+3h and T_0+6h). EMG data were acquired using A RHD2216 chip from Intan
27 Technologies. The data were recorded with 16-bit resolution and a sampling rate of 1.1kHz per channel
28 and then transferred to a PC through an USB cable. More information can be found in the datasheet of the
29 RHD2216 chip from Intan Technologies. The on-chip filters were set as follows: the first order high pass
30 was set at 10 Hz and the third order low pass filter was set at 300 Hz.

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52 *Processing of EMG data:* We used a customized Labview software to filter the signal and calculate the
53 signal-to-noise-ratio (SNR). All recordings were treated with a first order Butterworth band-pass (10-300
54 Hz). The SNR value of each recording was then calculated based on the following equation:

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3 $SNR(S_r) = 20 * \log \left(\sqrt{\frac{\sum_i S_{c_i}^2}{\sum_i S_{r_i}^2}} \right)$, where S_c is a portion of 5 s of the signal during a contraction and S_r is a
4
5
6 portion of 5 s of the signal during a resting position.
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11 Results and Discussion

14 Figure 1a shows a layer schematic of the tattoo electronics integrated with the textile contacts. The tattoo
15 includes one layer of PEDOT:PSS ink inkjet-printed onto the temporary tattoo paper. PEDOT:PSS coating
16 increases the roughness (RMS) of the substrate from 1.35nm to 3.77nm (Figure S1). The electrical
17 conductivity of a 95 ± 5 nm thick printed PEDOT:PSS electrode (single layer) is 240 ± 10 S/cm. The
18 thickness and the associated sheet resistance can be typically controlled by varying the number of printed
19 layers of PEDOT:PSS. However, we find the tattoo substrate not compatible with printing multiple layers
20 of aqueous ink due to instability of the sacrificial layer in aqueous. We estimated the conductivity of a
21 single layer of the film to be adequate to acquire EMG signals. Following the fabrication, the electronics
22 printed on the paper was transferred onto the skin by simply wetting the tattoo transfer paper and gently
23 pressing the tattoo against the skin. As the transfer paper is wetted, it dissolves and the conducting polymer
24 pattern remains on the skin. As inkjet offers freedom of form and design, different electrode patterns can
25 be achieved (Figure 1b). The electrode conforms well to the skin and is in fact reported to enter deeply
26 inside the creases of the skin [37] (Figure 1c). As the skin is compressed or stretched due to muscle
27 movements, the electrodes move with the skin, acting as its natural component (Figure 1c). The electrodes
28 tattooed on the skin have an intimate contact with the tissue and by using the conducting polymer which
29 has a Young's modulus of ca. 150MPa in wet state [25], similar to that of skin ca. 2kPa-500 MPa [38],
30 we overcome the inherent mechanical mismatch (Figure 1d).
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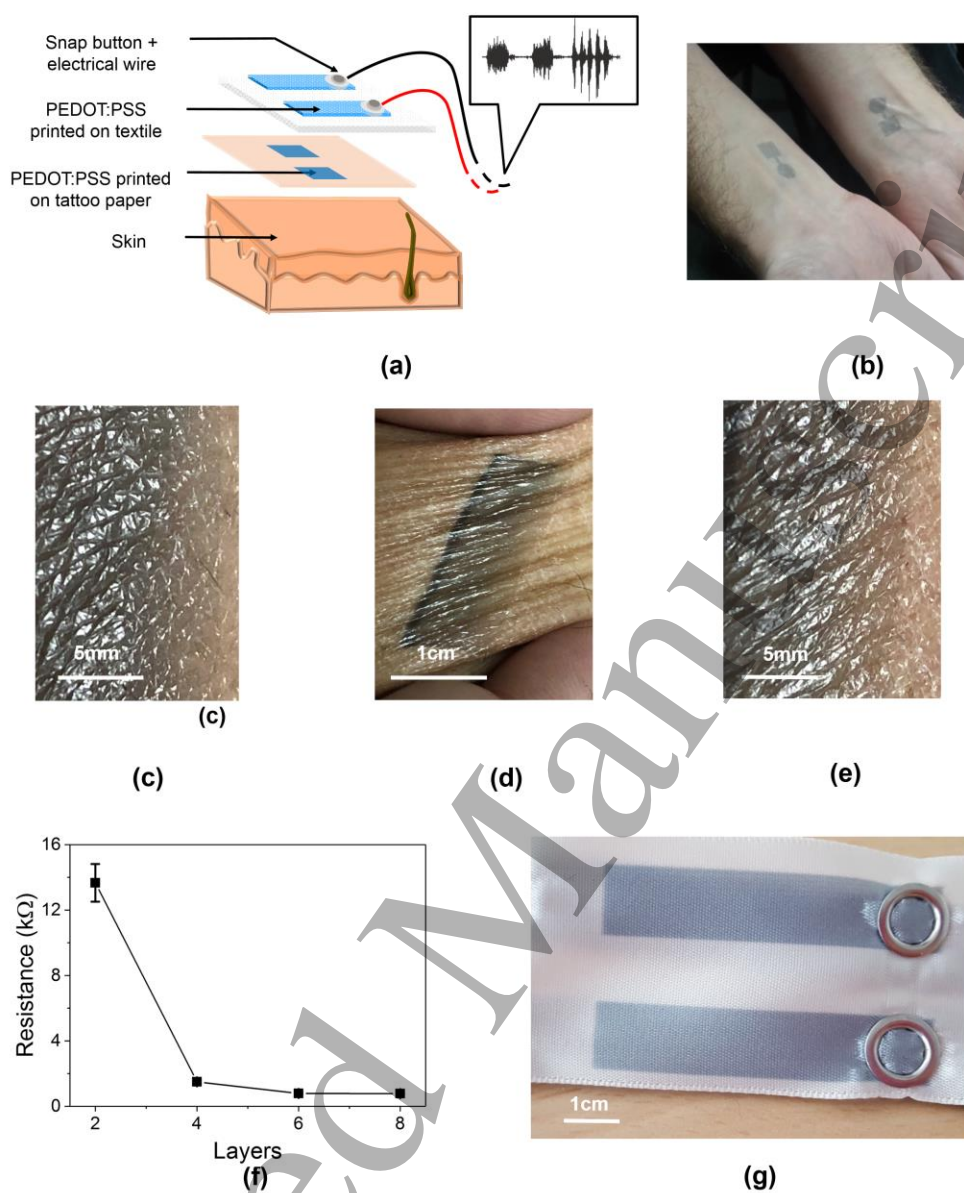


Figure 1. The architecture of tattoo-textile electronics (a) Schematic of the tattoo/textile electronics. The PEDOT:PSS electrode printed on the tattoo paper is transferred to the skin. The textile containing the printed conducting polymer contacts is wrapped around the tattooed electrode on the skin, connecting the tattoo to the acquisition system. (b) Photograph of customized printed tattoo electrodes deposited on the skin. (c, d, and e) Photographs of printed tattoo electrodes on the skin: undeformed (c), compressed several times (d), and after compression (e). The tattoo electrode shows a good resistance to mechanical deformation, with no cracks observed at the surface of the electrode. (f) Photograph of the PEDOT:PSS printed on the textile (6 layers) with the snap buttons integrated. (g) Electrical resistance of the printed PEDOT:PSS patterns on the textile ($1.5 \times 1.5 \text{ cm}^2$).

The conducting polymer was also printed on the textile wristband made of knitted soft fabric (i.e. satin) using the same printing protocol as for the tattoo paper (Figure 1f). In order to optimize the electrical performance of the contacts on the textile, we varied the number of the printed PEDOT:PSS layers

(1.5x1.5 cm²). The electrical resistance of the contacts decreased with the number of printed layers and reached a plateau with 6 layers of PEDOT:PSS (Figure 1g). The saturation is due to complete impregnation of the fabric with the polymer. Following the fabrication of electrodes, we inserted metal snap buttons to the textile on top of the conducting pattern. With the buttons attached, the textile was in the form of a wristband which could comfortably be wrapped around the target skin area where the tattoo electrodes were transferred and fixed using the snap button closure (Figure 1f). As such, we directly interface the tattooed electrodes with the contacts on the textile. The metal snaps enable to connect the tattoo electrodes to the acquisition system in order to record the electrophysiological activity. The electrical resistance between the tattooed electrodes on the skin and the metal snap on the textile is measured to be ca. 52 ± 5 k Ω (Figure S2). This configuration obviates the need to use external wires in contrast to other studies with EMG electrodes relying on external connectors such as silver cotton wires [28], gold coated plastic foils [29], or polyimide films [30]. These contacts tend to cause unstable measurements due to motion artefacts and discomfort to the user, as well as their integration requires additional fabrication steps. In the tattoo/textile electronics, there is only one main electronic material, PEDOT:PSS, and it is deposited in a single step using one fabrication protocol.

We compared the electrode-skin contact impedance of the tattoo/textile electronics with the conventional wet (gelled) Ag/AgCl electrodes, one hour after they have been in contact with skin (Figure 2a). Figure 2b shows the configuration of the measurements on the forearm of a volunteer. The commercial electrodes exhibited impedance values two orders of magnitude lower than the tattoo/textile electronics at low frequency range (from 0.1 Hz to 100 Hz). For frequencies above 100 Hz, the impedance of the tattoo/textile electronics was one order of magnitude higher. The lower impedance of the commercial electrodes is due to the presence of the gel electrolyte applied between the Ag/AgCl and the skin [39]. The tattoo/textile system, on the other hand, shows reduced impedance values for all the frequencies explored, as well as a smaller sample-to-sample deviation compared to the printed textile electrode only. At the frequency range where the significant EMG signals occur (5-450 Hz) [40], the tattoo/textile electronics

display an impedance of ca. $1.83 \pm 0.01 \text{ M}\Omega$ at 5 Hz and $225 \pm 10 \text{ k}\Omega$ at 450 Hz. Albeit with impedance higher than conventional electrodes, we and others postulate that the dry electrodes have the advantage of being more stable [41].

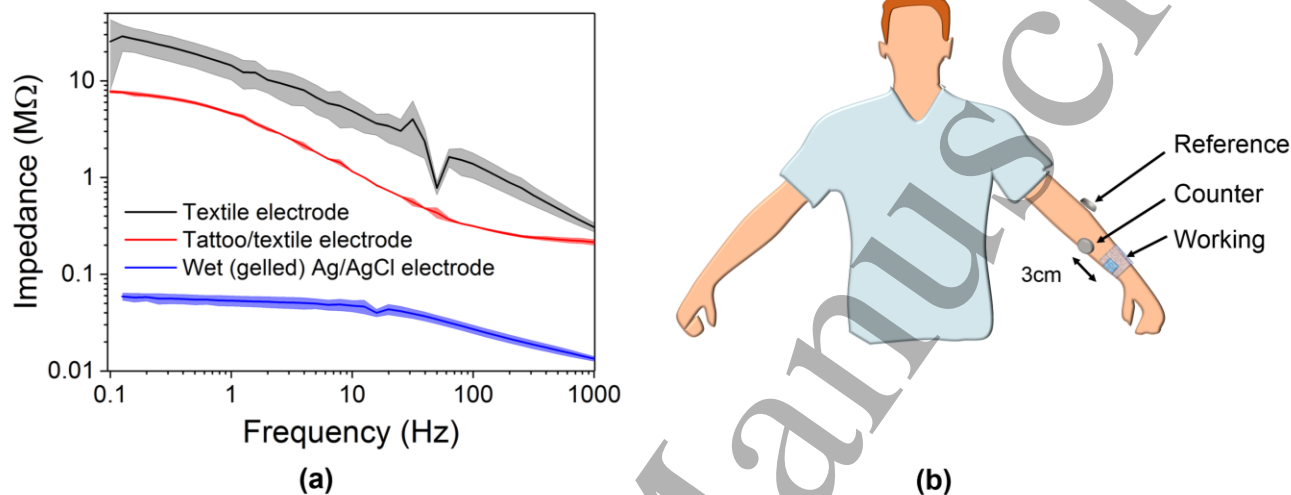


Figure 2. Electrical characterization of the electrodes (a) Electrochemical impedance spectra of the textile, tattoo/textile electronics and wet Ag/AgCl electrodes after one hour of contact on the skin. The evaluation includes three devices and the shaded areas represent the standard deviation. (b) Schematic representation of the experimental set-up showing electrode positions on the forearm: working electrode is the textile, tattoo/textile electronics or wet Ag/AgCl electrodes, while counter and reference electrodes are wet Ag/AgCl.

In contrast to the rigid conventional electrodes, tattoo electrodes conform with the skin. In order to demonstrate the stability of the tattoo electrode upon bending, we measured its electrical resistance under deformation. The tattoo was deposited between the distal and intermediate phalanges of the index finger of the volunteer who was asked to perform repetitive flexions/extensions (Figure 3a and b). We recorded the electrical resistance of the electrode as the finger was subject to different bending angles (30° and 90°) up to 250 cycles. While we do not observe a significant variation in the electrical resistance at a flexion to 30° , resistance increases slightly for flexion to 90° . This is attributed to the failure of the thick metallic adhesives used to contact the tattoo, proposed also by Zucca and colleagues [29]. In fact, we observe such cracks observed at the adhesive-tattoo interface after the experiments (Figure S3). In parallel, we examined the electrical stability of the contact printed on the textile electrode worn for a period of 7 hours on the

arm wrist. We observed no changes in the polymer conductivity for this period, evidencing the stability of the textile contact.

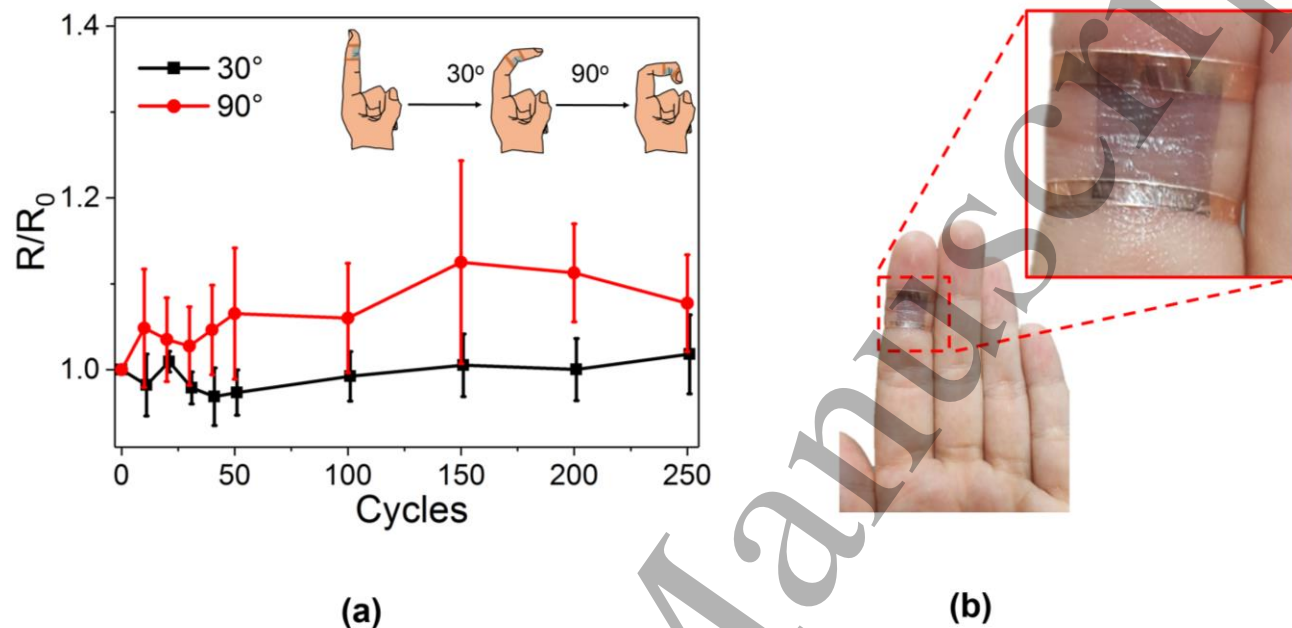


Figure 3. Evaluation of the resistance of the tattoo electronics on the skin towards bending (a) The change in the electrical resistance of the printed tattoo electrode upon repeated contractions of the index finger at 30° and 90° bending angle, **(b)** Photograph of the tattoo on the index finger of the volunteer and the bending experiment. The tattoo electrode is connected to the acquisition system via copper adhesives which were attached underneath the edges of the conducting pattern.

To test the performance of our tattoo/textile electronics in acquiring EMG signals, we transferred the tattoo onto the forearm of a volunteer and connected the tattoo electrode to the acquisition system using the electronic wristband as shown in Figure 4a. The tattoo electrodes (working and counter) were positioned 2 cm apart from each other on the lower part of the left arm's biceps and EMG signals were recorded over 6 hours once in every 3 hours. The signals acquired with the tattoo/textile electronics were compared to those acquired with wet Ag/AgCl electrodes. These electrodes were positioned at the exact same locations, this time on the right arm which was exposed to the same experimental conditions.

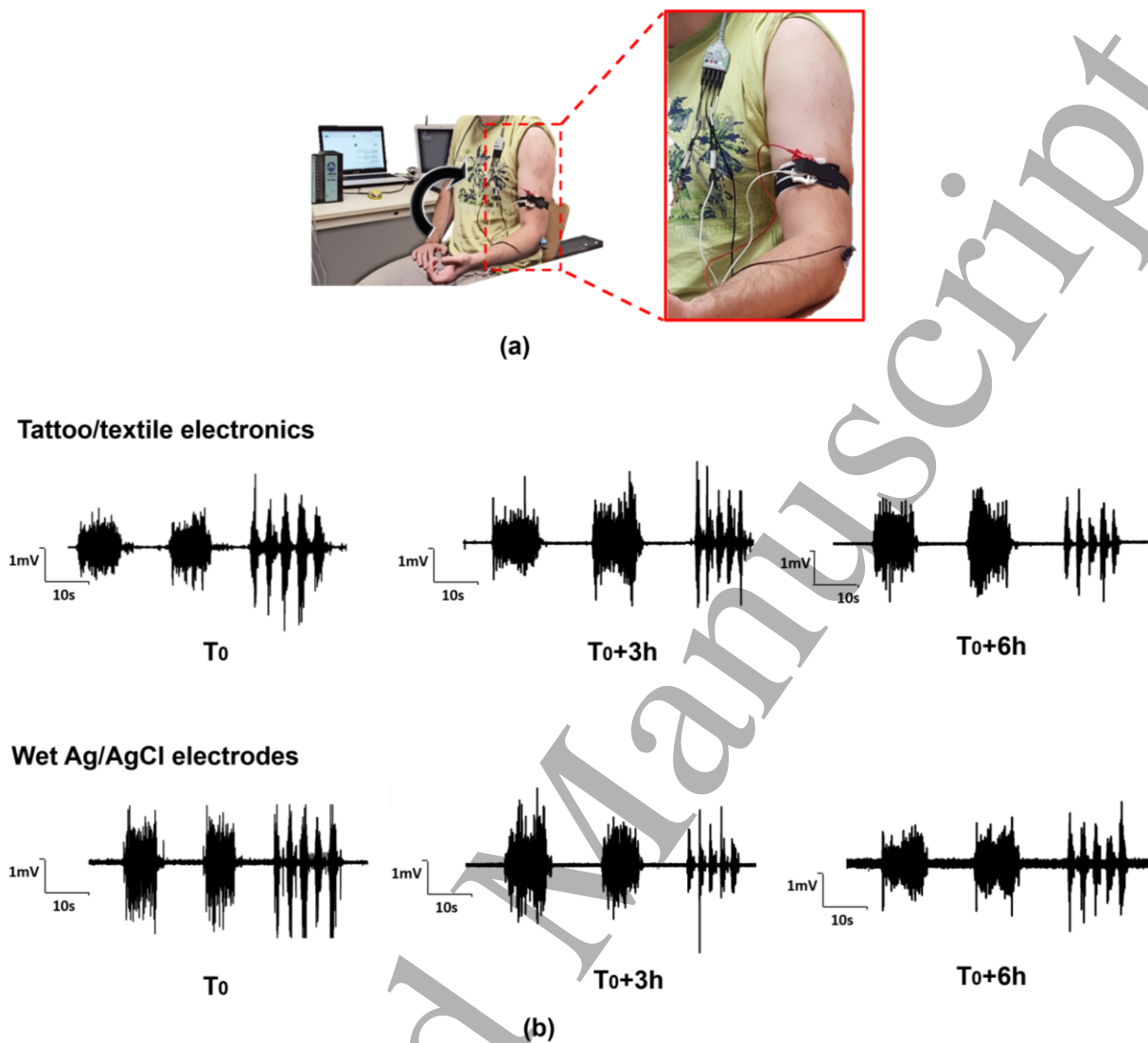
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Figure 4. EMG recordings of the biceps contraction (a) Photograph of the EMG experiment with a loop on the arm of the volunteer: the tattoo electrodes were deposited on the arm and covered by the textile electrode which was placed underneath a black adjustable sweatband, (b) EMG signals recorded with tattoo/textile electronics after 2 long contractions (10 s) and 5 short contractions (1 s) at T₀+1h, T₀+3h and T₀+6h as well as with wet Ag/AgCl electrodes. Each experiment was repeated 6 times.

Figure 4b shows the EMG signals recorded with tattoo/textile electronics as well as Ag/AgCl electrodes during the biceps activity for both short (1 s) and long (10 s) voluntary muscle contractions over 7 hours. The recordings used to extract the signal to noise ratio (SNR) consisted of 10 seconds-long contractions during which the volunteer was asked to lift a 1 kg of standard weight with an elbow angle of ca. 90°, with a resting period of 10 seconds. As the electrodes were placed on the skin (T₀=1h), the tattoo/textile electronics exhibited a mean SNR of 11±2 dB compared to a mean SNR of 29±1 dB for the wet Ag/AgCl

1 electrodes. The lower quality of the recordings with the dry tattoo after its adherence with the skin is
2 attributed to the high electrode-skin contact impedance of these electrodes (Figure 2a). Despite the lower
3 SNR, the tattoo/textiles electronics could measure the EMG activity. Interestingly, after 4 hours of contact
4 with the skin ($t=T_0+3h$), the performance of the electronics was substantially improved: SNR increased to
5 26 ± 2 dB while the SNR of Ag/AgCl electrodes decreased to 25 ± 1 dB. Note that the signal-to-noise ratio
6 is expected to be slightly higher at the non-dominant side (i.e., left arm for the right-handed volunteers),
7 validated by the tattoo/textile electronics upon $t=T_0+3h$. We postulate that the quality of the contact of the
8 tattoo/textile electronics with the skin increased due to the sweat-gland activity. The sweat produced by
9 the body wets the tattoo, thus enhances the ionic conductivity at the skin-electrode interface and leads to
10 a decrease of the contact impedance between the skin and PEDOT:PSS [42]. Another factor that
11 contributed to the reduced impedance is the improved quality of contact between the tattoo and the textile.
12 As shown in Figure 1a, there is a thin layer of the ionic conductor, ethyl cellulose, between the two
13 PEDOT:PSS layers (the tattoo and the textile). As the skin gets wet, this layer becomes more permeable
14 to ions allowing the current to pass easier from the skin to the textile contact. Finally, 7 hours after the
15 first recordings ($t=T_0+6h$), the quality of the recordings are similar, i.e., SNR corresponds to 21 ± 1 dB and
16 20 ± 5 dB, for tattoo/textile electronics and Ag/AgCl electrodes, respectively. To demonstrate the
17 reproducibility of the platform, we recorded similar type of EMG signals from 2 other volunteers using
18 new pairs of printed electrodes as well as at different contraction levels. At $t=T_0+6h$, the SNR values of
19 tattoo/textile electronics and wet Ag/AgCl electrodes were 31 ± 6 dB vs 30 ± 4 dB for one volunteer and
20 23 ± 1 dB vs 21 ± 1 dB for the other, respectively (Figure S4). We thus conclude that for recording cutaneous
21 EMG signals, the tattoo/textile electronics show similar performance to clinical skin-contact electrodes.

22 In order to decouple the effect of the tattoo electrodes from the textile, the EMG signals were recorded
23 using the textile electrodes (Figure S5). Although at T_0 , the SNR of the textile electrodes was higher than
24 the tattoo/textile electronics, and over time, the textile recordings became less stable (e.g. 18 ± 2 dB at T_0

1 and $4\pm 3\text{dB}$ at $T_0+6\text{h}$). Note that to be able to acquire these signals, the textile band has to be strongly tied
2
3 around the arm to assure contact with the skin. We, therefore attribute the decrease in the SNR (or its
4
5 instability) to the low skin-electrode contact quality rather than a degradation of the electrode. These
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7 results demonstrate the ability of the tattoo/textile electronics to record stable EMG signals over time
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9 alongside the simplicity of device fabrication/integration and gel electrolyte-free use.
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15 **Conclusions**

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17 We designed an inkjet-printed tattoo/textile electronics system utilizing the biocompatible conducting
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19 polymer PEDOT:PSS both as the transducer and the contact to the acquisition system. The polymer printed
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21 on the tattoo showed excellent adherence to the skin with no evident side effects such as skin irritation or
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23 allergies. The electrodes exhibited sufficient conductivity for acquiring muscle activities from the skin.
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25 The same conducting polymer was printed on a textile used as a wristband around the skin and acted as a
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27 bridge between the tattoo electrode and the external electronics. We show that over 7 hours of EMG
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29 recordings (biceps contraction), the tattoo/textile electronics showed similar performance to conventional
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31 wet Ag/AgCl electrodes without the use of gel electrolytes. The integration of the conducting polymer
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33 both as the electrode and contact material minimizes the cost while improving the feasibility of such skin-
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35 adhering electronics envisaged to be applied for diagnostics and/or therapy. The intimate contact of the
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37 tattooed electronics with the skin, its conformability and bending with the movements of the limbs is a
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39 great advantage over other cutaneous systems. Our approach for interfacing the external electronics avoids
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41 the use of inorganic rigid wires and presents a wearable solution. Due to the versatility of the materials
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43 and the processes employed, these systems can encompass a variety of applications such as myoelectric
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45 prosthesis or detection of muscle functioning.
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